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Microprocessor Controlled Gamma-Ray CT Scanner For Measurement Of Bone Density

by

© Keith Daryl Whitmore

A THESIS

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The undersigned certify that they have read, and recommend to the Faculty of Graduate Studies and Research, for acceptance, a thesis entitled Microprocessor Controlled Gamma-Ray CT Scanner For Measurement Of Bone Density submitted by Keith Daryl Whitmore in partial fulfilment of the requirements for the degree of Master of Science.

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Abstract

The theory, design and construction of a microprocessor controlled gamma-ray CT bone scanner are described. This scanner, with its ability to determine the radiological density of trabecular bone at appendicular sites in the skeleton, provides an order of magnitude increase in sensitivity to changes in bone mineralization over existing techniques such as absorptiometry and radiographic analysis. This technological development could be invaluable to the clinician for diagnosis and in the evaluation of treatments of osteopenia. The design and implementation of both a single and multi-detector rotation-translation scanning system are detailed along with future plans for a fan-beam rotational system.

Acknowledgements

I would like to express my appreciation to my supervisor, Dr. T.R. Overton for his efforts (in terms of administrative assistance and moral encouragement) which have increased both the value and personal satisfaction of my education. This project has been a team effort and hence there are many people whose efforts I would like to recognize. These include: Mr. I. Yamamoto (for his aid in design and construction of the mechanical components), Mr. R. Heath (for the design, testing, and implementation of the HP programs), Mr. J.D. Ridley (for his aid in evaluation of the scanner), Dr. D. Menon (for his assistance in helping me to understand some of the theoretical aspects of CT and his review of this manuscript) and Dr. T.N. Hangartner (for his efforts in upgrading the CT system and his willingness to share his expertise). Lastly, I would like to express my thanks to my many friends in the Division of Biomedical Engineering who have made my educational experience truly enjoyable.

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I. Historical Aspects

It has long been the goal of medical radiology to obtain clear cross-sectional views of the body. As early as 1921 (1) conventional or focal plane tomography was being used. Conventional tomography involves the movement of both the X-ray tube and film (or even the body itself) in a trajectory (Figure 1) so that only the desired layer remains in focus while all other layers are blurred. This type of tomography which is analogous to a narrow-depth-of-field in photography has the disadvantage of the blurred presence of unwanted fields.

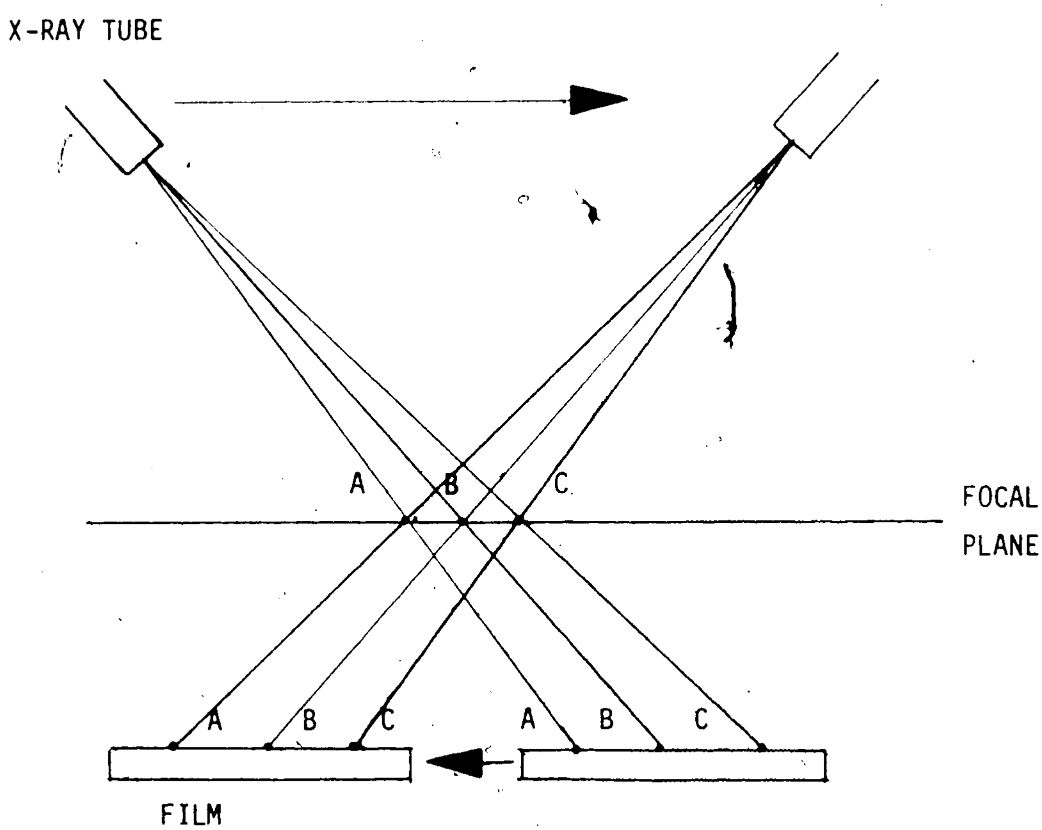
Computed tomography (CT), which is the latest evolution in a series of techniques used in image reconstruction, eliminates these unwanted planes by limiting radiation to the desired plane. CT is based on the principle that if radiological attenuation is measured along a sufficient number of lines passing through a given plane, then the radiological densities throughout that plane can be determined.

The mathematical basis for computed tomography was first described by Radon (2) in 1917. Given that

1. $f(x,y)$ denotes the attenuation coefficient at a point (x,y) in the plane, and
2. that L is any line in the plane:

then, the logarithm of attenuation is equal to the integral of f along L; $P(L) = \int f(x,y)\, ds$

where s indicates a length along L. The problem then is to



ONLY POINTS ON THE FOCAL PLANE ARE CONTINUALLY REGISTERED ON THE SAME PART OF THE FILM. POINTS IN OTHER PLANES ARE IMAGED ONTO VARYING PARTS OF THE FILM AND HENCE ARE BLURRED.

FIGURE 1. CONVENTIONAL TOMOGRAPHY

invert the equation: t· ·t is, given P(L), to determine f(x,y). Assuming P(L) is given for all lines L and that f(x,y) is continuous with compact support, Radon showed that f(x,y) could be calculated for all points in the plane. At the time of its development this mathematical basis for image reconstruction was purely theoretical in nature and in actual fact was derived by Radon while he was working with equations dealing with gravitational fields.

It was not until 1956 that this theory was made use of by Bracewell (3) in the field of radioastronomy. Bracewell was able to identify regions on the sun which emitted microwave radiation. However, the antennas being used could only focus on thin strips criss-crossing the solar surface. In order to map out the solar regions of microwave activity it was necessary to convert these strips into a representative plane of solar activity - the exact problem shown by Radon to be mathematically solvable. Bracewell developed both an analytical technique (equivalent to Radon's) and an iterative technique for image reconstruction.

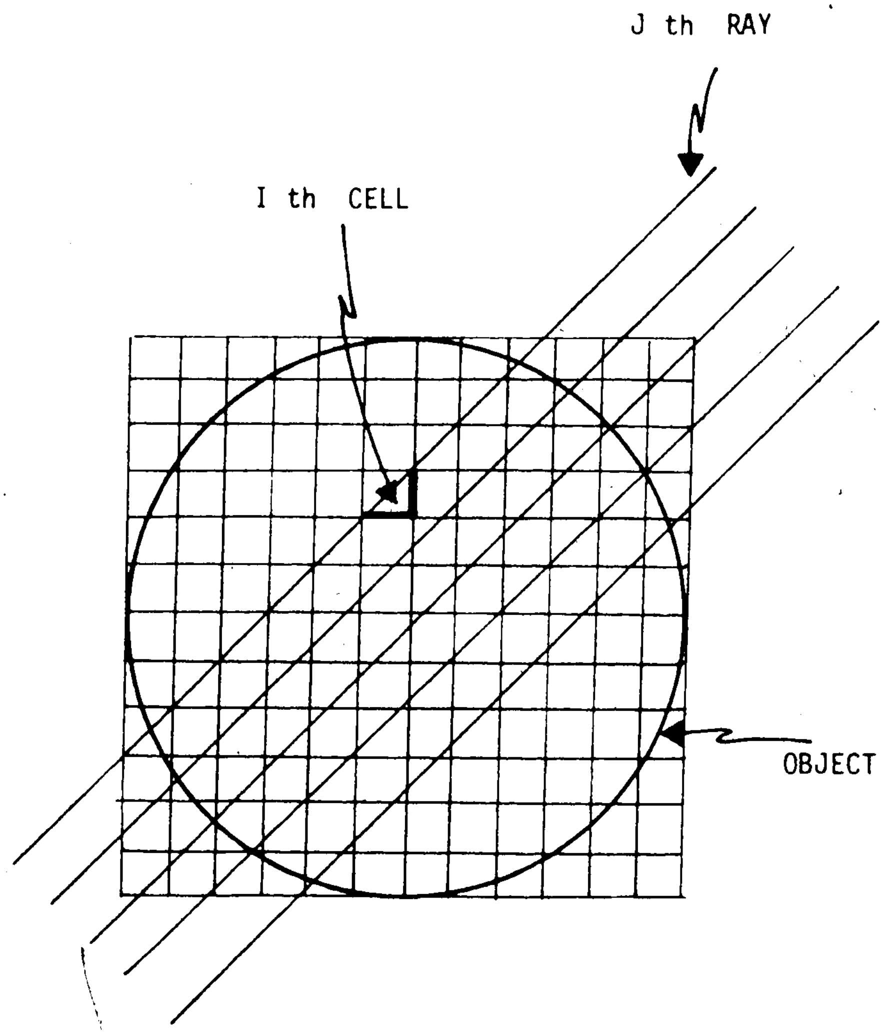
The first medical application appeared 7 years later when Cormack (4) published his first experimental results. Cormack's work involved development of a mathematical tech ique for the accurate reconstruction of an image from x-ray projections and applications based on measurements of simple phantoms. This technique for reconstruction involved the expansion of P and f into Fourier series and the subsequent matching of Fourier coefficients.

The first commercial scanner was the E.M.I. Head Scanner described in 1973 by G. Hounsfield (5). Having realized that the ray-integrals essentially composed a discrete rather than a continuous number of equations, Hounsfield's approach to reconstruction involved an iterative solution to inversion of the equation defined by P_j = the summation over i of W_{ij} f_i (Figure 2) where W_{ij} is a weighting function of the radiological density, f_i. This weighting function represents the contribution of the ith cell to the jth ray.

From these early days CT has evolved from the point where only scans of the head could be performed to today where scans of the entire body can be made. A 1977 study of the U.S. Department of Commerce (6) predicted that by 1980 there would be some 2500 CT scanners in the U.S. and annual related expenditures of $1.7 billion. Over a dozen different companies were involved in marketing CT machines in 1976: machines able to delineate differences in soft tissue density as low as 0.5 % (7).

A. Measurement of Bone at the University of Alberta

The use of CT for the measurement of bone density, as developed in Zurich by Rüegsegger and Elsasser (8), is a relatively new phenomenon. Since changes in bone mineralization occur much more rapidly in trabecular bone than in cortical bone (9), detection of small changes in



THE RECONSTRUCTION MATRIX IS AN N BY N ARRAY OF CELLS.
THE ACTUAL RAYS ARE STRIPS OF FINITE WIDTH. THE CONTRIBUTION
OF THE I th CELL TO THE J th RAY IS DENOTED BY Wij.

FIGURE 2. RECONSTRUCTION MATRIX

mineral content are only possible if trabecular bone density can be quantified. It is for this reason that CT scans which can be used to produce images of bone cross section are so valuable.

Although the principles involved in reconstruction of an image of bone are the same as in commercial whole body scanners, there are a sufficient number of differences to merit a specially designed CT scanner.

The Division of Biomedical Engineering at the U. of A. has been involved in the clinical determination of bone mineral content using absorptiometric techniques for the past 8 years (10). This clinical experience led to the development of a CT system for the measurement of bone mineral content at the University of Alberta. Based upon an original design by Elsasser (11) an eight detector, translation-rotation gamma-ray CT system has been constructed. With this scanner, measurements can be made of the distal forearm using Iodine 125 as a photon source.

This thesis is concerned primarily with the control system for the scanner. A microprocessor unit or MPU (Motorola 6800 D2 kit) serves as a system master clock, providing control of scanner movement and for data acquisition through the laboratory computer (HP 2100) and CAMAC systems. An MPU based controller was chosen as it was felt that the timing specifications on the scanner would be too tight to be controllable by the HP running under a multi-user operating system.

II. Bone Physiology and Pathology

In the diagnosis of various diseases of bone it is desirable to measure changes in mineral content of trabecular (spongy) bone. To understand this requirement it is necessary to examine the physiological and pathological characteristics of bone. Once the need for bone mineral density analysis has been described an examination of existing analysis techniques will be presented.

Bone consists of an organic matrix that is greatly strengthened by deposits of calcium salts (12-14). Five percent of this organic matrix is a homogeneous medium called ground substance while the other 95 % is made up of collagen fibers. It is these collagen fibers which give bone its tensile strength while the calcium salts provide its compressional strength. The calcium salts are composed principally of calcium and phosphate and are formed in a crystal called hydroxyapatite. The crystals are between 200 and 400 angstroms in length and between 10 and 30 angstroms in diameter. Under varying nutritional conditions the ratio by weight of calcium to phosphate in these crystals can vary from 1.3 to 2.0.

As well as having the obvious structural function the skeleton acts as a reservoir for many of the body's ions, notably calcium and phosphate. Bone contains 99% of the total body calcium and 90% of the total body phosphorus. From 0.4 to 1.0 % of this bone calcium is readily exchangeable and this exchange occurs at points in the bone undergoing active calcium resorption (remodelling) (15). Thus there is a constant circulation of ions between the skeletal reservoir and body fluids and tissues. Ions not contained in the bone have a wide range of physiological activities. For example; calcium not in the bone affects enzyme activity, cell membrane permeability, cardiac rhythm and neuromuscular excitability. The concentration of calcium in the blood is maintained between 9 and 11 mg per 100 ml and it is the buffering action of bone that provides this control. Exchange of calcium with the bone is controlled mainly by parathyroid hormone from the parathyroids and calcitonin from the thyroid. Another means of control of calcium ion concentration is absorption from the gut. Vitamin D_2, for example, is required for calcium absorption from the gut.

Bone is made up of 3 types of cells..These cells are scattered throughout an interstitial matrix of fibrous protein and occupy 3% of the total volume of the bone. Osteocytes are the ordinary bone cells and comprise the bulk of the cells. They are incapable of mitosis and hence must be removed before new bone can be formed. Osteoblasts are the cells which secrete the ground substance and collagen onto the bone forming surface. This uncalcified organic matrix is known as osteoid. The osteoblasts produce an enzyme, alkaline phosphotase, which is thought to act by splitting organic phosphate compounds thus upsetting the local calcium-phosphate balance causing precipitation of calcium salts in the ground substance. Osteoclasts or bone phagocytes are the cells involved in the removal of bone. This occurs by both a phagocytic action and the release of enzymes and acids which cause the simultaneous dissolution of mineral and matrix. Thus bone is continually being deposited by osteoblasts and being resorbed at sites where osteoclasts are active. Osteoblastic deposition is stimulated by physical stress while sex hormones and calcium supplements decrease both bone resorption and formation.

On a macroscopic level there are two main classes of bone: cortical or compact bone, which is the harder outer bone and trabecular or spongy bone which is the less dense inner bone. Average compact bone consists of 25 % by weight matrix and 75 % calcium salts while trabecular bone is made up of a much higher percentage of matrix. Due to a greater surface area it is in this less dense trabecular bone structure where physiological and pathological changes in matrix structure or calcium salt deposition occur more rapidly and hence it is desirable to measure the distribution of density in trabecular bone.

A. Osteoporosis and Osteomalacia

Two common diseases of bone are osteomalacia or "adult rickets" and osteoporosis. Osteomalacia is a lack of calcium salts and is due to a lack of calcium and phosphates in the extracellular fluid (16). This is often caused by a lack of calcium absorption from the gut possibly due to a vitamin D deficiency. Kidney disorders are one common non-nutritional cause of osteomalacia.

Osteomalacia involves an increase in the proportion of uncalcified bone (osteoid) with a corresponding increase in both the number and thickness of osteoid seams. The mineralized bone mass is usually decreased due to a decrease in calcium and phosphate. The bones are soft and skeletal deformities may ensue. Symptoms of osteomalacia include weakness, bone pain and tenderness as well as radiologic evidence of decreased mineralization and pseudofractures of the long bones.

Osteoporosis is abnormal organic matrix formation (17,18) and may be caused by: (1) lack of use of the bones, (2) malnutrition to the extent that sufficient protein matrix cannot be formed, (3) lack of vitamin C, (4) postmenopausal lack of estrogen secretion, and (5) old age (due to dysfuntion of various protein anabolic functions). Osteoporosis therefore involves a decrease in bone mass due to the absence of a normal quantity of bone caused by deficient production of bone matrix. The mineral content of the bone however, is grossly normal and there is no excess of osteoid. Thus there is an imbalance between osteoblastic bone deposition and osteoclastic bone resorption. Clinically there is a decrease in the number of and thickness of bony trabeculae. The cortical bone later becomes thinned; however, it is still relatively dense compared to the decreased density of the trabecular bone. There is an increase in susceptibility to fractures from compression forces (eg. vertebral bone compression fractures).

The density of bone has been shown to decline in all people from the age of 20 onwards, the process accelerating after middle age (16,18). To this extent bone becomes osteoporotic after 50 years of age to a greater extent and somewhat earlier in women than in men.

Decreased bone mineralization, which may accompany osteopenia (general bone disorders), appears radiologically as areas of decreased density. As such, it is possible to use radiographic techniques to diagnose and to evaluate the effectiveness of treatment in diseases related to changes in bone mineralization.

"Since the disorders causing rickets and osteomalacia are generally of long standing and must have profound effects on matrix compostion or mineral metabolism before they become apparent clinically, therapy may be necessary for months or years before healing is complete." [1] The same may be said for osteoporosis. Some of the present treatments include mineral supplements, gonadal hormones and physical therapy. Some of these treatments may have other effects. For example, pharmacologic doses of vitamin D may cause hypercalcemia. Treatments then are very dose dependant and/or expensive and hence any technique which will provide an early and accurate means of diagnosis and measurement of changes in bone density has potential for both therapeutic

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[1] Current Therapy 1977, Edited by Conn: pg 461.

and financial benefit. Since changes in bone density first appear in trabecular bone, a technique is required that is able to measure density of trabecular bone.

B. Physical Evaluation of Bone

This section will examine some of the techniques presently being used for physical evaluation of bone. These include both invasive and non-invasive techniques.

Bone Biopsy

Bone biopsies involve the physical removal of a sample of bone usually from the iliac crest (hip bone). This sample is then analyzed, usually by microscopic examination of the number of trabeculae and osteoid in a given area within the sample. Although this technique does provide valuable information the sample removal is an invasive technique and is traumatic for the patient. Results obtained vary greatly from site to site and once bone has been removed the same site cannot be re-examined. This then precludes the use of bone biopsies as an accurate means of monitoring both the course of any diseases such as osteoporosis and osteomalacia as well as the effectiveness of any treatments. Thus the use of bone biopsies is severely limited.

Radiology (X-rays)

Prior to 1963, diagnosis of osteoporosis and osteomalacia predominantly involved the radiological evaluation of vertebral skeleton. Radiographic techniques or X-rays show total density of a number of planes superimposed together on one view and hence any one plane of high density will overshadow any of the other planes. Thus it is largely cortical bone that shows on X-rays and hence measurements of density of trabecular bone are not feasible. Accurate diagnosis of either osteoporosis or osteomalacia using X-rays requires the diseases to be significantly advanced involving decreases in bone mass of greater than 25% (16,18) and/or compression fractures. Due to the lack of precision of this technique, longitudinal studies (studies over time) are difficult to perform and it is virtually impossible to evaluate the progress of disease and the effectiveness of treatments. Radiographic distinction between osteoporosis and osteomalacia is also difficult and the two diseases may coexist.

Absorptiometry

In 1963, Cameron and Sorenson (19) reported an improved technique, known as absorptiometry, for the in vivo measurement of bone mineral content. Their method involved the use of a nearly monoenergetic collimated source of photons mechanically coupled and collinear with a collimated scintillation detector. The scintillation detector, with appropriate electronics, is able to measure both the energy of, and the number of photons striking it in an interval of time. Therefore, one can measure both the number and energy of photons striking the detector with and without an object placed between them. These measurements allow the amount of energy absorbed by the object to be calculated. In the case of bone the number of photons removed from the beam by the object is related to the mass of bone present. The measurement technique then involves movement of the source detector system along a linear path across the object with transmission measurements taken at discrete intervals throughout the translation (Figure 3). This results in an absorption profile of the object, the area of which is a function of the number of photons absorbed by the object.

Although this technique is similar to a conventional X-ray in that radiological attenuation is measured, the improved equipment results in marked improvements in both the accuracy and precision of the measurement. Absorptiometry could be considered a digital form of radiology in that quantitative rather than qualitative transmission measurements are made. Measurement of bone mineral content on a cadaver forearm (20) using absorptiometry was found to have an accuracy to within 4-7 % of that obtained from measurements of bone weight. Reproducibility of measurements on a single individual were found to have a coefficient of variation of 2.5 %. These improvements in accuracy and precision of absorptiometry versus conventional X-rays were due to elimination or reduction of such causes of error as film exposure, development and calibration; measurement errors due to scattered radiation and the broad beam polychromatic x-rays as well as errors from soft tissue effects. Variations in quantities of soft tissue are reduced by submersion of the

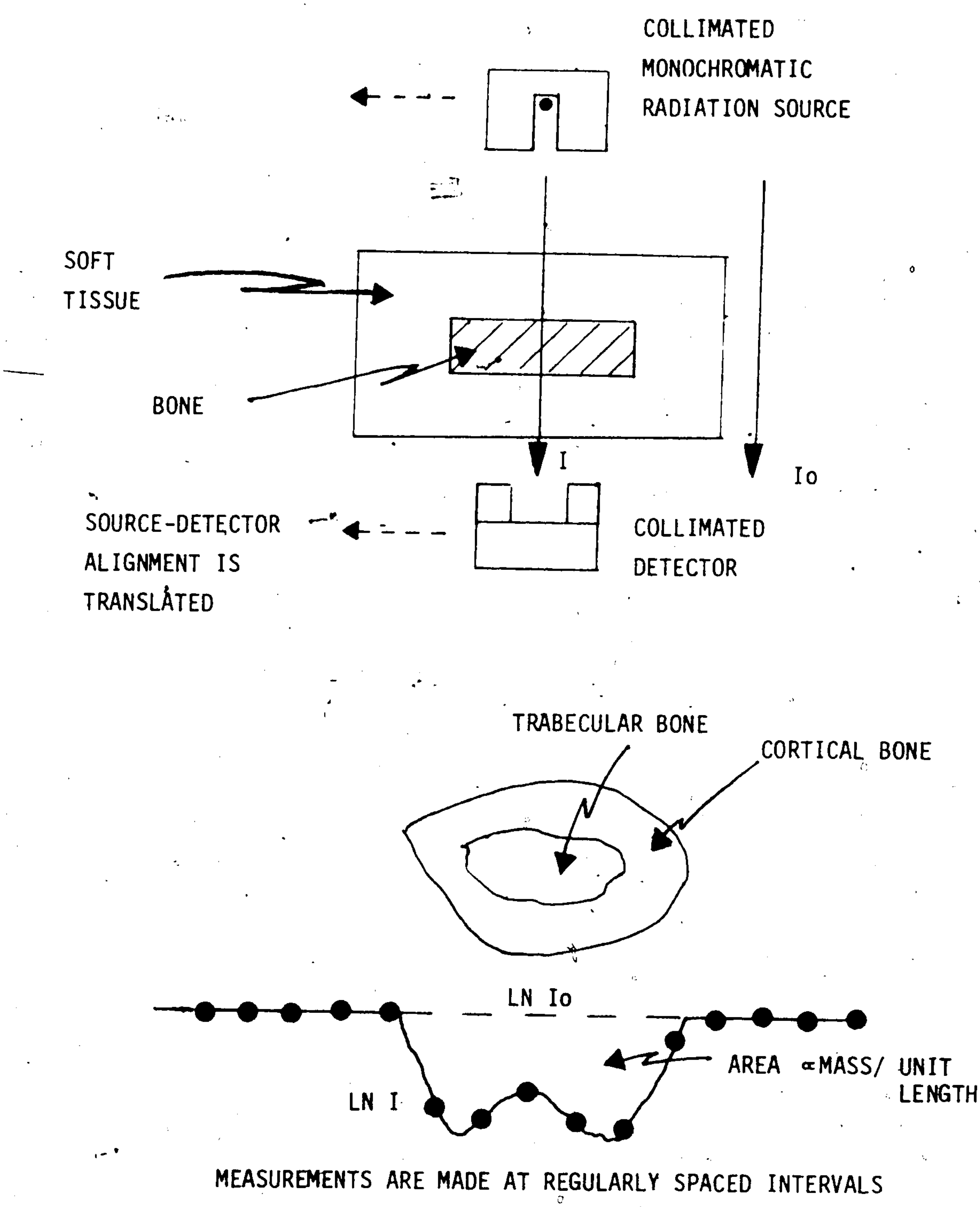

FIGURE 3. ABSORPTIOMETRY

limb in water, thus providing a constant pathlength of attenuation. The radiological attenuation of fat, collagen, water and other organic components of soft tissue are all approximately equal.

Loss of vertebral skeletal bone has been shown to be reflected in measurements of radial bone mass, in particular in the distal region. A study was done (21,22) of 169 white women, 113 of whom had no fractures while 56 had radiological evidence of spinal fractures. Those women with no evidence of fractures had higher values of radial bone mass with a significance of 98.8 % for cortical bone and better than 99.9 % for trabecular bone. Although total bone mineral content was found to be a reliable determinant for diagnosis of decreased mineralization it was also found that in cases of very slight or exceptionally heavy built patients, evaluation of mineral content divided by bone width provided a more useful parameter. Absorptiometry studies involve measurement of a single radiation absorption profile. This allows one to evaluate total bone mineral content and thickness of cortical bone but quantitative evaluation of bone mineral density is not possible. Thus, absorptiometry requires a measurable change in the thickness of cortical bone or total bone mineral content to evaluate disease processes and this typically requires a period of several months to years.

In Vivo Neutron Activation Analysis (IVNAA)

Neutron activation analysis involves irradiation of a volume by neutron bombardment and the subsequent detection of emitted radiation to determine the quantity of a particular material present (23). Irradiation of calcium in bone results in the production of a number of different isotopes of calcium. By measuring the radiation given off by one of these isotopes the quantity of that isotope can be calculated. Using the ratio of the quantity of that isotope to total calcium, total bone calcium can be calculated. This technique although providing valuable information is a costly process (requiring 14 MeV neutron sources) with limited accuracy and precision (of the order of 5 %).

Compton Scattering

Another technique which has been used for the *in vivo* measurement of bone mineral density is the Compton scatter technique (24). This technique involves the irradiation of a collimator defined volume and the measurement of Compton scatter from a particular portion of this volume by a collimated detector. From these measurements it is possible to determine the average electron density of the measured volume. The major limitations of the technique relate to the fact that only a relatively small volume of material can be studied and as such repositioning problems (in order to measure the same volume at a future time) are a major concern.

Radioisotope CT

Although CT, with its ability to image cross sections, would appear to provide a technique for measurement of bone density there are a number of problems with standard commercial X-ray scanners that preclude such measurements. These problems are mainly manifest in the areas of beam hardening (due to use of a multi-energy X-ray source), physical resolution (due to the degree of collimation of the source and detectors) and the sample spacing (which is required to cover an abdominal area, leaving too little matrix in an area of bone for sufficient analyses to be carried out). A technique able to measure density of both trabecular and cortical bone with a precision of better than 1 % using a radioisotope CT scanner was first reported by Rüegsegger and Elsasser (8) in 1975. The principles behind this special purpose CT scanner are the same as for commercial whole body scanners; however, changes have been made in order to facilitate the measurement of bone rather than soft tissue density. The technique involves performing a scan of the radius and ulna at a distal site , usually 1/10 the distance from the ulna styloid to the proximal point of the ulna. At this site although some 50 % of the bone area is trabecular it contributes only 10 % to the total bone mineral content. Given this relatively large area of trabecular bone it is possible to separately determine the density of both trabecular and cortical bone at this site. The technique is non-invasive, not traumatic to the patient and allows repeat measurements of the same sample to be performed.

It is also hoped that CT may be used to examine the bone-cement and metal-cement contact on artificial hips (25). Although this problem is somewhat distinct from the measurement of bone density (and will require an X-ray tube photon source) many of the problems encountered are similar (i.e. measurement of radiological attenuation in high density areas such as bone). Upon implantation it is difficult to determine the degree of bonding of artificial hips and subsequent loosening may result.

Use of a photon source with energies higher than the 27.5 KeV of Iodine 125, such as Gadolinium 153 (40 KeV, 100KeV), may allow the CT scanner to be used for examination of bone gaps and degradation in patients who are having knee joints implanted. A higher energy source is required because of the quantity of bone involved and hence the high radiological attenuation. At present it is often necessary to perform an exploratory operation to determine the sizes of bone gaps and the amount of degradation of the bones. It is possible that the CT Scanner may be used to eliminate this exploratory surgery.

In order to fully appreciate the potential uses of a CT Scanner it is first necessary to have a basic understanding of what computed tomography is and how it works. This area will be explored in the next chapter.

III. Principles of CT Scanners

"CT" or computed tomography (7,26-30) is now the preferred name for "CAT"(computerized (assisted) axial tomography). A tomogram is literally a picture of a slice or section at a given orientation. Radiation in the form of gamma or X-rays are passed through (transmission tomography) or originate (emission tomography) from the desired plane of view without entering other areas. The amount of radiation absorbed along a given pathlength is measured and is an indication of the sum of linear attenuation coefficients or radiological densities along the pathlength. A linear attenuation coefficient is a measure of the absorption of radiation as it traverses a distance in an object. It should be noted that the attenuation coefficient in any object is dependent on the material composition and the incident photon energy. This sum of attenuation coefficients is measured along a sufficient number of pathways so that an image depicting the radiological densities (or attenuation coefficients) throughout the object can be reconstructed. Since this work deals primarily with transmission tomography any further references to CT will be referring to transmission CT.

To this date there have been 5 generations of CT scanners. These generations are identified by the type of mechanical motion used and the number of detectors employed. The first two generations of scanners (Figures 4 & 5) are known as single detector and multiple detector translation-rotation systems. Scans are performed by

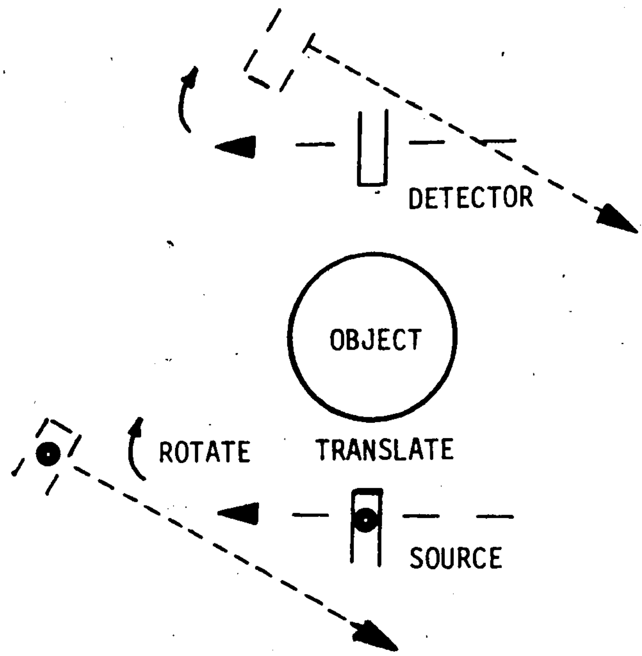

FIGURE 4. CT GENERATION 1

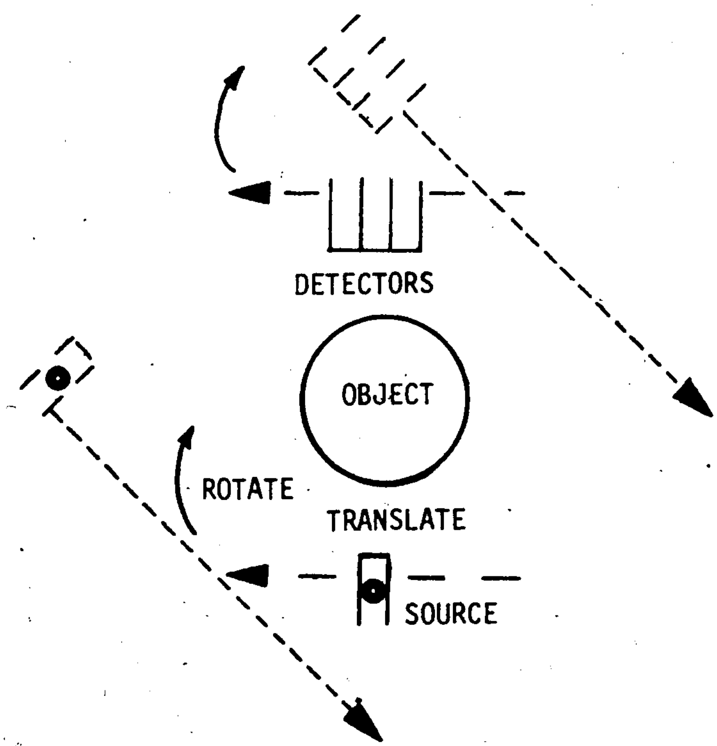

FIGURE 5. CT GENERATION 2

translating the source-detector configuration past the object at which point they are rotated about the object and the translation is repeated in the opposite direction. Measurements of ray-integrals or total radiological attenuation are continually measured during translation. This movement results in a "smoothing" or blurring of the object as the actual measured data is taken over a finite area of the object. Translations and rotations are repeated until the sum of the rotations meets or exceeds the 180 degrees required for complete reconstruction. Multiple detector systems are advantageous in that rotations between translations can be made through a larger angle and hence the number of translations and rotations can be decreased thus reducing scan time.

The third and fourth generation scanners (Figures 6 & 7) are multiple detector rotation-only systems. In the case of the third generation scanner the source and detectors each rotate about the same point through the required circle. In the fourth generation scanner the detectors form a stationary circle of 360 degrees and only the source is rotated. This eliminates the need to rotate the detectors and their associated high voltage and signal cables. The major advantage of the rotation only scanners versus the rotation-translation scanners is the reduction in scanning time. This reduction in scan time is important as it reduces image artifacts that can be caused by patient movement.

The fifth generation scanners (Figure 8) employ both

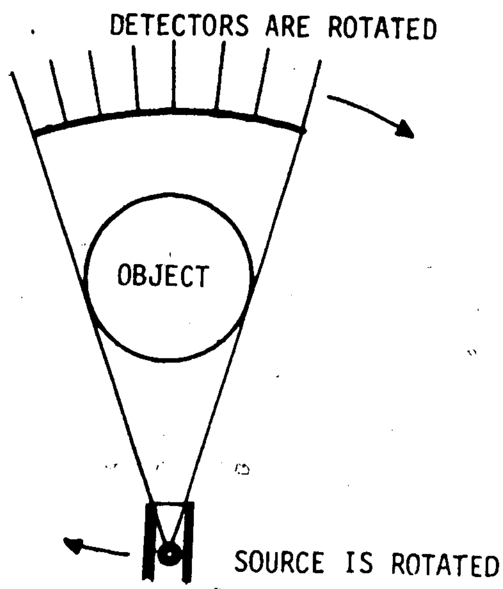

FIGURE 6. CT GENERATION 3

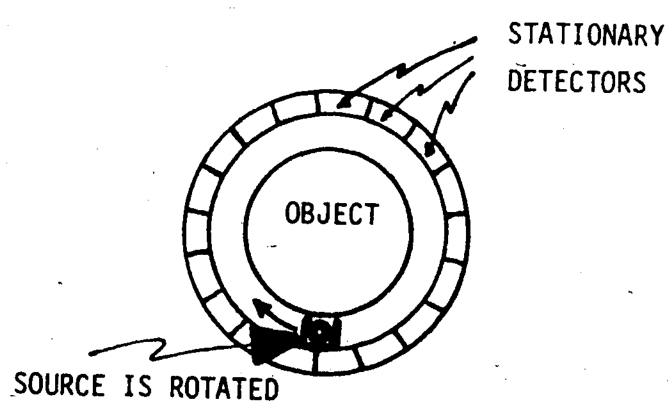

FIGURE 7. CT GENERATION 4

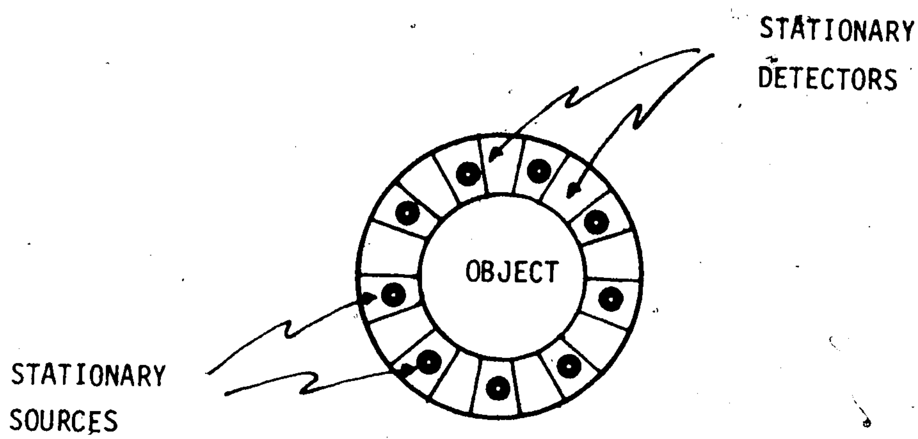

FIGURE 8. CT GENERATION 5

multiple sources and multiple detectors and may allow ultrafast scanning to produce dynamic representations of cardiac motion without special gating or synchronization techniques. Such scanners are a relatively new phenomenon and introduce additional problems in terms of expense and calibration.

Radiological attenuation along lines through the plane is often measured in a parallel beam format associated with the first two generations of scanners. This means that attenuations are measured along a series of regularly spaced parallel lines taken at a number of angular orientations (Figure 9). Each series of measurements taken at a given angular orientation is referred to collectively as a view or profile. In the cases of the third, fourth and fifth generation scanners data is collected in a rotational format(Figure 10). This data can be reordered and interpolated in order to create the parallel beam format thus permitting the same reconstruction algorithms to be used for all generations of scanners. However, special algorithms have been developed for the rotational format (31,32).

A. Back-Projections

The first technique (7) used for reconstruction involved a method now referred to as Back-Projection. [2] For

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[2] Capitalization is used to distinguish between Back-Projection as a method of image reconstruction, and back-projection as a process.

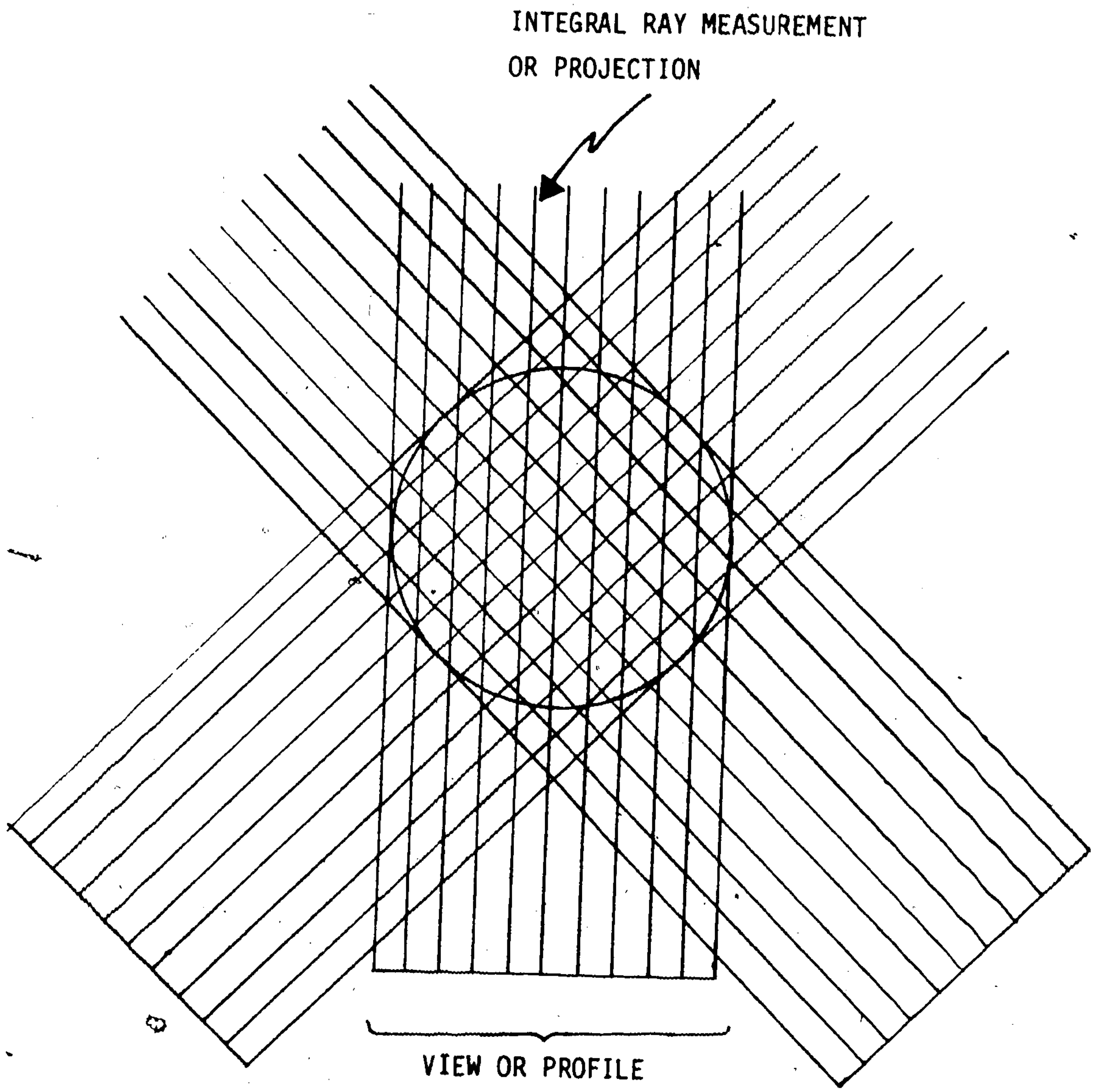


FIGURE 9. PARALLEL BEAM FORMAT

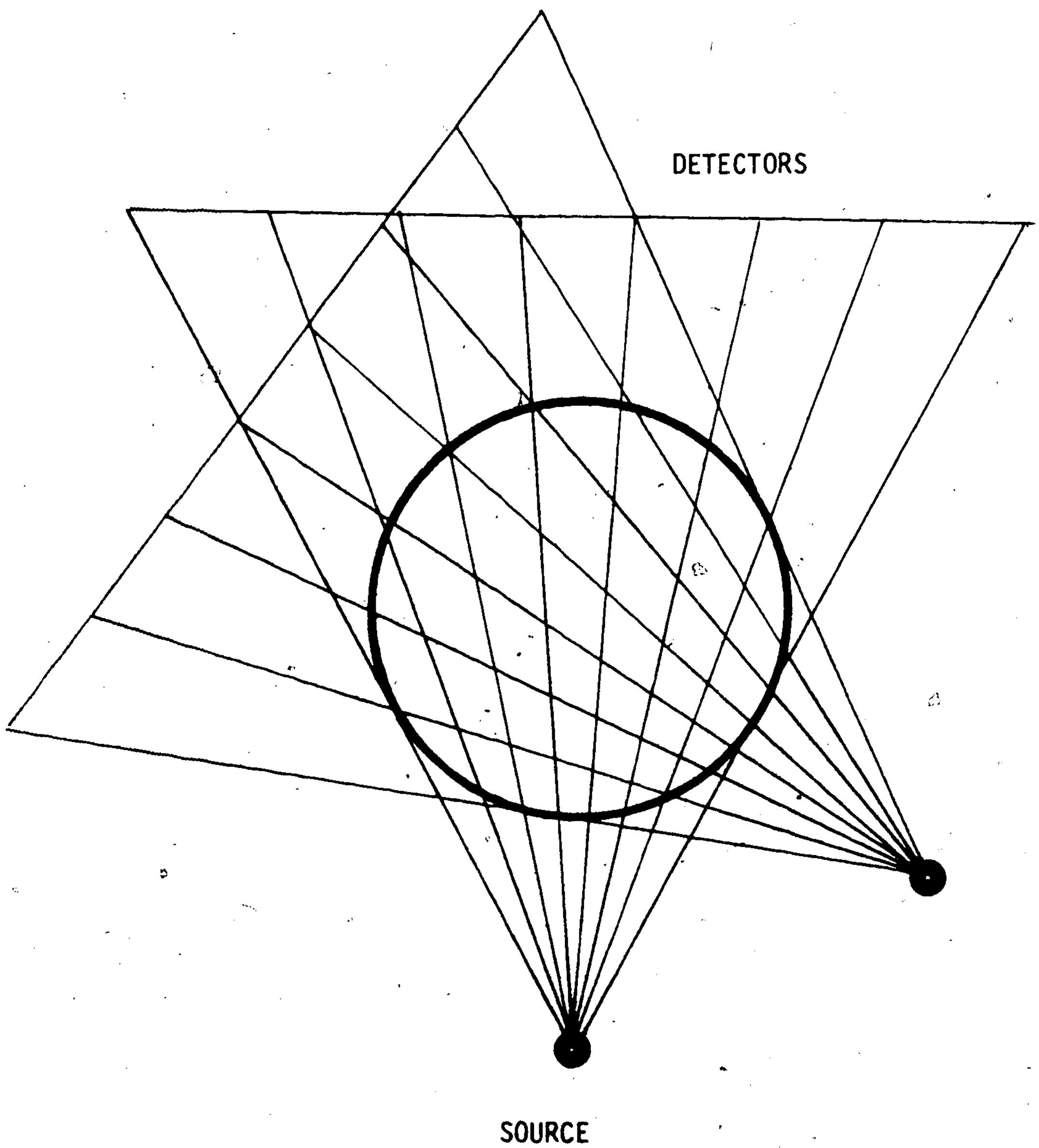
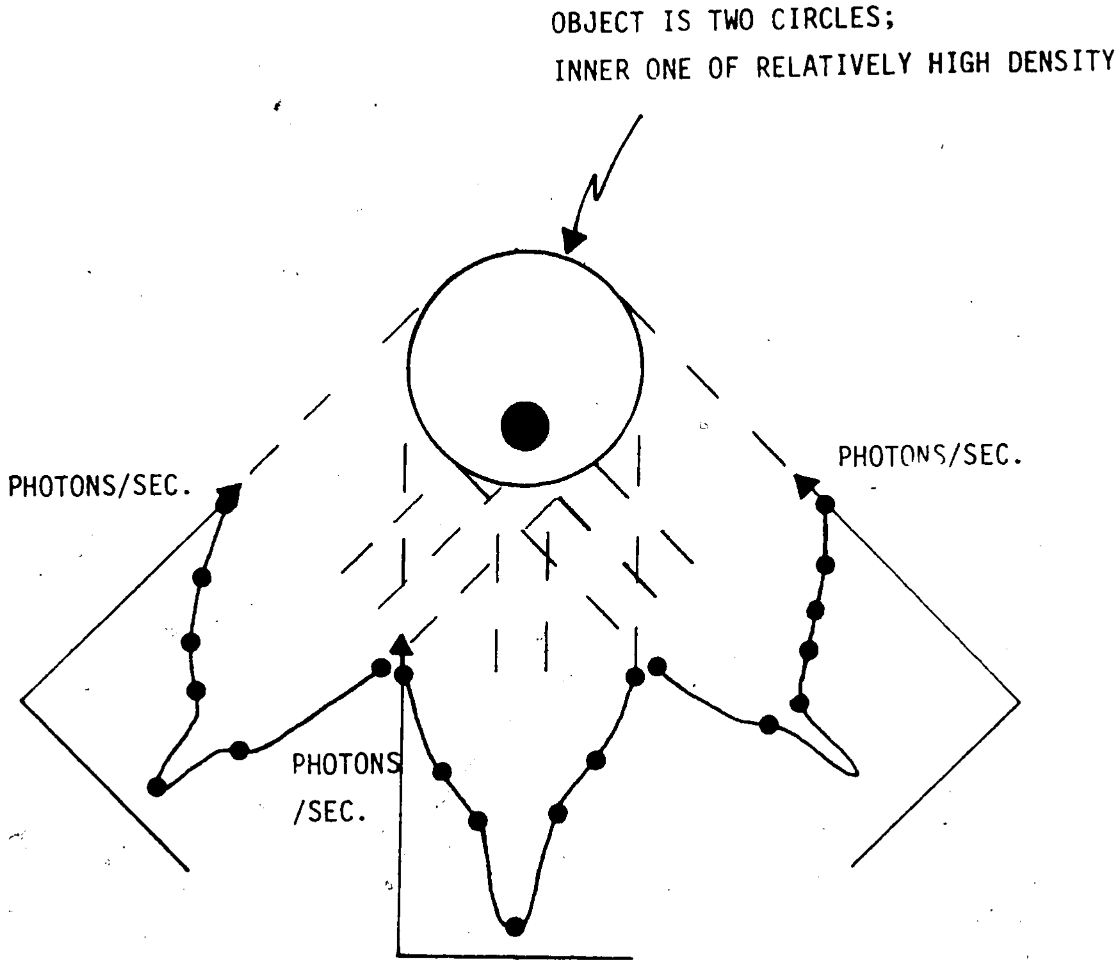


FIGURE 10. ROTATIONAL BEAM FORMAT

illustrative purposes three profiles or projections of an object have been taken (Figure 11) and are then back-projected to reconstruct the object in Figure 12. Each profile is made up of a discrete number of measurements taken at regularily spaced intervals. Reconstruction involves the back-projection of each profile onto a reconstruction matrix. This matrix reconstruction may be performed optically (for example on a screen), using analog electronics or digitally in a computer. The signal intensity according to a given ray sum is applied to all points that make up that ray and this is done for all projections giving an approximation to the original object. As seen in Figure 12 points outside of the original object may receive some of the back-projection intensity resulting in what is known as the "star" artifact. Points within the object also receive components from neighbouring points meaning that subtle differences in density cannot be determined. For these reasons Back-Projection is no longer being used as a technique for reconstruction. However, it does serve as a basis for understanding the reconstruction techniques presently being used.

The technique now used involves measurement of individual profiles and their respective angular orientations and the subsequent storage of this data in a computer. The data is then processed using mathematical algorithms and the reconstructed object is displayed on a plotter using different grey levels to represent the varying



PROFILES ARE MADE UP OF A DISCRETE NUMBER OF MEASUREMENTS

FIGURE 11. BACK-PROJECTION

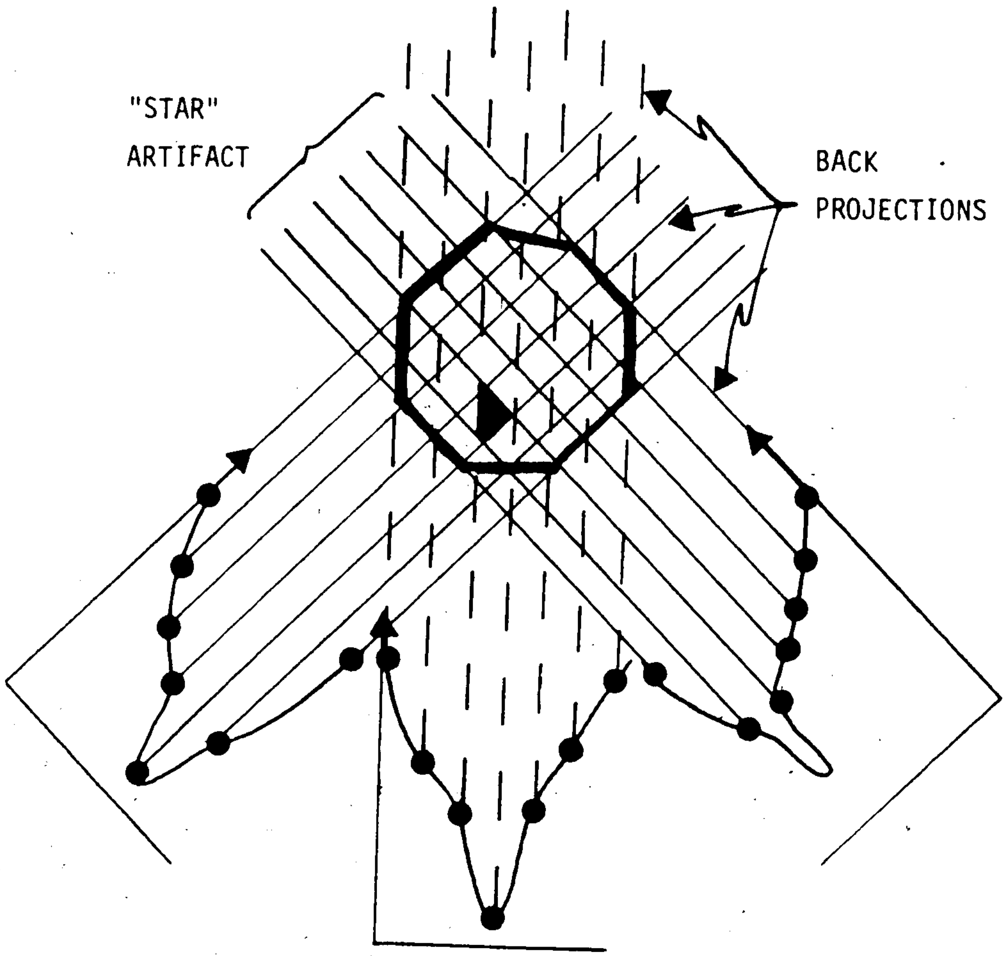


FIGURE 12. IMAGE RECONSTRUCTION USING BACK-PROJECTION

densities. Although the method depicted in Figures 11 and 12 merely sums the individual profiles, mathematical techniques allow for cancellation of back-projections and hence the object can be reconstructed more exactly.

B. Reconstruction Algorithms

The two main classes of algorithms used for reconstruction are the iterative and analytical techniques.

Iterative Technique

The iterative technique involves formulating an arbitrary image, calculating the profiles for that image and comparing these profiles to those obtained experimentally. The starting image is usually a blank screen (all ray profiles equal zero) or a circular grey object of uniform density (all ray profiles equal a constant). According to the differences in the measured profiles and those of the arbitrary image the image is modified and the profiles are compared again. This process is repeated until differences between the theoretical and experimental profiles are within acceptable limits. There are three general classes of the iterative technique (7). In the simultaneous correction technique each cell that contributes to a ray is altered and all projections are corrected simultaneously. This technique tends to overcorrect and hence iterations oscillate about the correct value. In the case of the ray-by-ray correction, corrections are made to all points of one ray at which point these corrections are taken into account prior to making further corrections to a different ray. This technique is found to work best when large angles are taken between consecutive projections to be corrected. The third technique is a point-by-point correction in which each point is corrected for all rays that pass through it and all past changes are embodied into future changes.

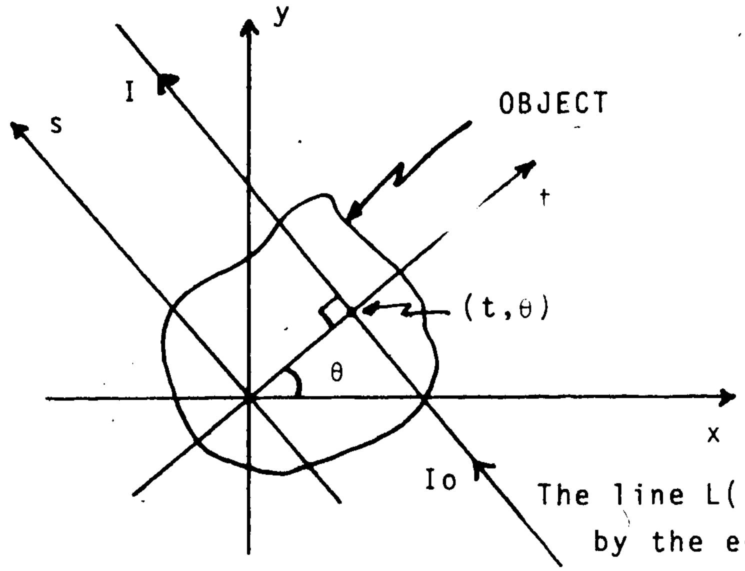
In each of the three cases the corrective mechanism can be either additive or multiplicative. Additive, which is the method primarily used, refers to the fact that correction is divided among cells according to their weighting factor. In the multiplicative method, correction is applied to cells according to their present density. In this case a grey starting level is required. The iterative techniques can require a great number of iterations and computations and hence computing time may be a limiting factor.

Analytical Technique

The Analytical Technique can be subdivided into two major categories: (1) two-dimensional Fourier reconstruction and (2) filtered Back-Projection. Two-dimensional Fourier reconstruction is based on the fact that the Fourier coefficients of the image are equal to the Fourier coefficients of the projections at the same angle. This means that if the Fourier coefficients of the projections are known then the Fourier coefficients of the image are also known and hence by taking their inverse transform the image of the object can be reconstructed. Until the introduction of the Fast Fourier Transform (FFT) and improvements in high speed computers had taken place, this technique was not practical to use due to the number and complexity of the calculations involved. Mathematical development of this two-dimensional Fourier reconstruction technique is shown on the following pages.

Filtered Back-Projection or the convolution technique is based on this two-dimensional Fourier reconstruction. Papers by Shepp & Logan (33) and Ramachandran & Lakshminarayanan (34) have shown that this Fourier reconstruction may be viewed in the spatial domain as the sum of each ray-integral times a weighting function of the distance from the ray to the point of reconstruction. Thus reconstruction involves prefiltering each ray measurement by a weighting function and then back projecting these filtered ray measurements to reconstruct the object. This allows the image to be reconstructed as each ray measurement is taken rather than having to wait for all measurements to be completed. This method of reconstruction is accurate, simple and greatly reduces computation time over other available methods. The mathematical formulation of the convolution technique from Fourier reconstruction is also shown on the following pages.

ANALYTICAL RECONSTRUCTION TECHNIQUES



The line $L(t,\theta)$ is defined by the equation $x \cos\theta + y \sin\theta = t$

Io is the number of incident photons

I is the number of photons after passing through the object along line $L(t,\theta)$

In this diagram there are three different coordinate systems. The s-t coordinates are obtained by rotation of the x-y cartesian coordinate system while the ω-θ polar coordinate system uses ω as the radial component and the angle θ as the angular measure.

A line $L(t,\theta)$ (which is defined by the equation $x \cos\theta + y \sin\theta = t$) is chosen so as to pass through the object. This line depicts a ray integral measurement as would be made during a CT scan. In the parallel beam format a number of parallel ray-integral measurements (profile) would be made along a line $L(t,\theta)$ at which point the angle θ would be altered and another profile measured. It is due to this pattern of measurement that the s-t coordinate system has been introduced. The use of the polar coordinate system permits convolution algorithms to be used for reconstruction. This aspect will be discussed later in the paper.

First, some definitions:

(1) $f(x,y)$ is the linear attenuation coefficient at the point (x,y);

(2) Io is the number of incident photons; and

(3) I is the number of photons after passing through the object along line $L(t,\theta)$.

The total attenuation of the ray along line $L(t,\theta)$ is defined to be $P(t,\theta)$ which is equal to

$$-\ln(I/Io) = \int_L f(x,y)\, ds$$

where ds is an incremental length along $L(t,\theta)$.
The one-dimensional Fourier transform of $P(t,\theta)$ is

$$\hat{P}(\omega,\theta) = \int_{\infty}^{\infty} \exp(-i\omega t)\, P(t,\theta)\, dt = \int_{-\infty}^{\infty}\int f(x,y)\, \exp(-i\omega t)\, dsdt.$$

Now, consider the transform from the s-t coordinates to the x-y coordinates where

$$x = t\cos\theta + s\sin\theta \quad , \quad y = t\sin\theta - s\cos\theta \quad ,$$

$$t = x\cos\theta + y\sin\theta \quad \text{and} \quad s = x\sin\theta - y\cos\theta \ .$$

The Jacobian which arises from the transformation is

$$\frac{\partial(s,t)}{\partial(x,y)} = \begin{vmatrix} \frac{\partial s}{\partial x} & \frac{\partial s}{\partial y} \\ \frac{\partial t}{\partial x} & \frac{\partial t}{\partial y} \end{vmatrix} = \begin{vmatrix} \sin\theta & -\cos\theta \\ \cos\theta & \sin\theta \end{vmatrix} = 1.$$

Hence dsdt which is equal to the magnitude of the Jacobian times dxdy

$$= \left|\frac{\partial(s,t)}{\partial(x,y)}\right| (dxdy) = |1| \; dxdy = dx \; dy.$$

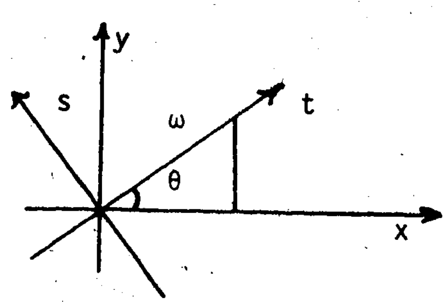
Therefore $\hat{P}(\omega,\theta)$ may be written as

$$\hat{P}(\omega,\theta) = \iint_{-\infty}^{\infty} f(x,y) \exp(-i\omega (x \cos\theta + y \sin\theta)) \, dx \, dy$$

which coincidentally is also the two-dimensional Fourier transform of x and y (define as $\hat{f}(\omega,\theta)$).

i.e. $\hat{f}(\omega,\theta) = \iint_{-\infty}^{\infty} f(x,y) \exp(-i\omega (x \cos\theta + y \sin\theta)) \, dx \, dy$

where ω is the spatial frequency in the direction of the t axis.



Hence, $\cos\theta = x/\omega$ and $\sin\theta = y/\omega$.

Now, if we consider the transformation from the x-y cartesian coordinate system to the ω-θ polar coordinate system the Jacobian arising from this transformation is

$$\frac{\partial(x,y)}{\partial(\omega,\theta)} = \begin{vmatrix} \frac{\partial x}{\partial \omega} & \frac{\partial x}{\partial \theta} \\ \frac{\partial y}{\partial \omega} & \frac{\partial y}{\partial \theta} \end{vmatrix} = \begin{vmatrix} \cos\theta & -\omega \sin\theta \\ \sin\theta & \omega\cos\theta \end{vmatrix} = \omega .$$

Hence, dx dy which is equal to $\left|\frac{\partial(x,y)}{\partial(\omega,\theta)}\right| d\omega\, d\theta$ $= |\omega| d\omega\, d\theta$.

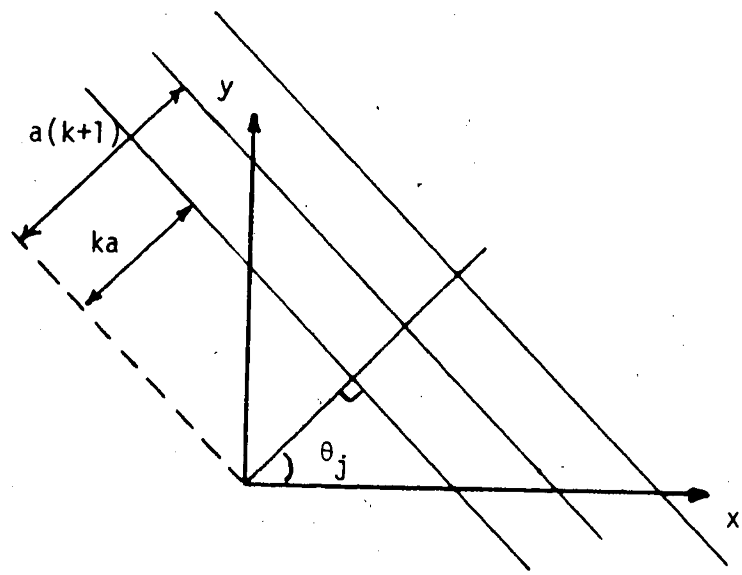
This means that f(x,y) may be written as

$$f(x,y) = \frac{1}{4\pi^2} \int_0^{\pi} d\theta \int_{-\infty}^{\infty} \hat{P}(\omega,\theta) \exp(+i\omega\,(x \cos\theta + y \sin\theta\,))|\omega| d\omega.$$

Thus, f(x,y) may be uniquely determined by a knowledge of $P(t,\theta)$ for all lines L. In reality $P(t,\theta)$ is known only along a finite number of lines which may be denoted by $t = t_k = ka$, $k = 0, \pm 1, \pm 2, \ldots$;

$j = \frac{j\pi}{n}$, $j = 0,1,2,\ldots$

where a is the parallel spacing between projections and n is the number of views.



Rewriting the integrals as finite sums we get

$$\int_0^{\pi} d\theta \simeq \sum_{j=0}^{n-1} \frac{\pi}{n} \quad \text{and} \quad \int_{-\infty}^{\infty} d\omega \simeq \sum_{k=-\infty}^{k=\infty} a.$$

By bounding the object to a circle of diameter a

$$\sum_{k=-\infty}^{k=\infty} \quad \text{can be rewritten as} \quad \sum_{k=-1/a}^{k=1/a}$$

where there are 2/a rays contained in each profile.

Hence, f(x,y) may be written as

$$f(x,y) = \frac{a}{4\pi n} \sum_{j=0}^{n-1} \sum_{k=-1/a}^{k=1/a} \hat{P}(\omega,\theta) \exp(-i\omega(x \cos\theta + y \sin\theta)) |\omega| .$$

Consider the Fourier inversion formula

$$f(x,y) = \frac{1}{4\pi^2} \int_0^{\pi} d\theta \int_{-\infty}^{\infty} \hat{P}(\omega,\theta) \exp(i\omega(x \cos\theta + y \sin\theta)) |\omega| \, d\omega$$

Substituting in $t = x \cos\theta + y \sin\theta$ the inner integral may be written as

$$Q(t,\theta) = \int_{-\infty}^{\infty} \hat{P}(\omega,\theta) \, |\omega| \exp(i\omega t) \, d\omega .$$

Now if $f(x,y)$ is a smooth function, then $P(t,\theta)$ will be a smooth function and hence $\hat{P}(\omega,\theta)$ will be bandlimited.

i.e. for $\omega > \Omega'$ $\hat{P}(\omega,\theta) \to 0$.

If there also exists an even function

$$\hat{\Psi}(\omega) = \int_{-\infty}^{\infty} \Psi(t) \exp(-i\omega t) \, d\omega$$

which equals $|\omega|$ for $\omega < \Omega$

then $Q(t,\theta)$ can be approximated by

$$Q(t,\theta) \simeq \int_{-\infty}^{\infty} \hat{\Psi}(\omega) \hat{P}(\omega,\theta) \exp(i\omega t) \, d\omega$$

which according to the Convolution theorem is a convolution in the time domain.

$$Q(t,\theta) = 2\pi \int_{\infty}^{\infty} \Psi(\tau) \, P(t-\tau) \, d\tau$$

Therefore, $$f(x,y) = \frac{1}{2\pi} \int_0^{\pi} \theta \int_{-\infty}^{\infty} P(\tau,\theta) \, \Psi(t-\tau) \, d\tau$$

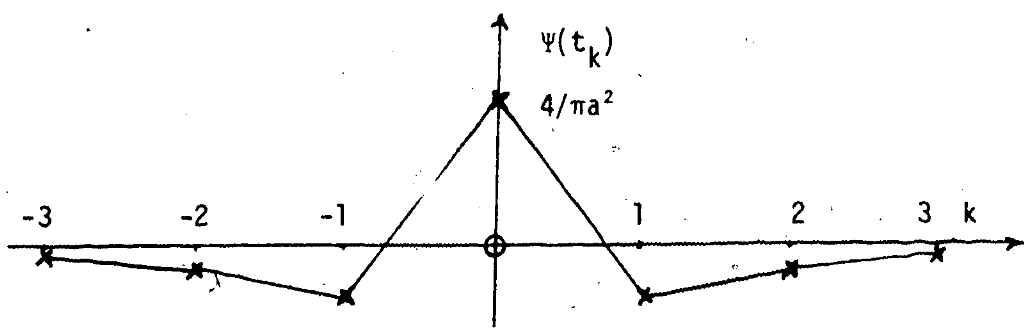
Once again writing the integrals as finite summations we get

$$f(x,y) = \frac{a}{2\pi} \sum_{j=0}^{n-1} \sum_{k=-1/a}^{1/a} P(t_k,\theta) \, \Psi(x \cos \theta_j + y \sin \theta_j - t_k)$$

Thus f(x,y) can be written as a summation of the ray integrals times a weighting function Ψ which is a function of the distance between the line $L(t_k,\theta)$ and the point of reconstruction (x,y). One commonly used weighting function (derived by Shepp and Logan) is

$$\Psi(t_0) = \frac{4}{\pi a^2} \quad ; \quad \Psi(t_k) = -4/(\pi a^2(4k^2-1)) \quad , \; k = \pm 1, \pm 2, \ldots$$

with linear interpolations in the intervals.



Allowing Ψ to be linear in the intervals is not incompatible with the requirement that for small ω, $\hat{\Psi}(\omega) \approx |\omega|$ and greatly reduces computational time since the weighting function can be computed and stored and values at intermediate points can be obtained through interpolation.

The Shepp, Logan weighting function $\Psi(t)$ has the Fourier transform

$$\hat{\Psi}(\omega) = \left|\frac{2}{a}\sin\frac{\omega a}{2}\right| \left(\sin(\omega a/2)/(\omega a/2)\right)^2 .$$

When $\omega \to 0$, $\hat{\Psi}(\omega) \to \left|\frac{2}{a}\,\frac{\omega a}{2}\right| \left((\omega a/2)/(\omega a/2)\right)^2 = |\omega|(1)^2 = |\omega|$

and thus the necessary requirement is met.

Comparison of Analytical and Iterative Algorithms

Both the iterative and analytical techniques have characteristic advantages and disadvantages. It is generally accepted (7) that analytical methods are faster than are iterative. Use of the convolution technique also allows the image to be reconstructed as measurements are made. In both cases there are problems associated with reconstruction. Due to the digitization involved in analytical methods the projections must be bandlimited. This means that an overshoot phenomenon will occur at sharp edges such as bone-flesh interfaces. Although this effect can be partially corrected by prefiltering, any such gains will be offset by a loss in spatial resolution. Iterative techniques are able to avoid the problems associated with sharp interfaces.

With complete data the two types of algorithms are comparable with respect to the amount of noise in the image. However, in cases where some of the data is missing, the iterative technique will prove to be superior since it merely "smooths" the image while the analytical techniques assume the missing data is the same as the available data.

C. Photon Attenuation

The ability of CT to distinguish between narrowly differing densities is dependent on the characteristics of photon attenuation. As photon radiation passes through an absorbing medium it is absorbed exponentially according to the following relationship (35,36);

$I(S) = Io \exp -(\mu(S-So))$

where Io is the number of photons at a distance S=So and μ is the linear attenuation coefficient. This absorption is due to four processes: the photoelectric process, the Compton process, coherent scattering and pair production. The linear attenuation coefficient (μ) has units of inverse distance and is a measure of the fraction of photons removed per unit length of absorber. μ may be converted to either the mass, electronic or atomic attenuation coefficients by dividing by either mass density ρ (gm/cm^3), electron density ρ_e(electrons/cm^3) or atomic density ρ_a(atoms/cm^3). At moderate energies (less than 1.02 Mev) photon attenuation predominantly consists of two contributions: (1) Compton scattering, which is proportional to electron density and (2) photoelectric absorption which is a function of the atomic number of the element (approximately Z^4). At the lower edge of the energy spectrum it is the photoelectric effect which is predominant and as such, measured attenuation is representative of the atomic numbers of the elements contained in the object. If either simultaneous or sequential scans at different energies (37) were used it would be possible to obtain measures of both the Compton and photoelectric effects allowing for evaluation of both electron density and atomic number. Hence an _in vivo_ elemental identification is possible using CT (38).

Statistical Considerations

Radioactive decay is a statistical process that can be modelled by a Poisson distribution (39-43). The characteristic of a Poisson process is that the mean and variance are equal. This means that the coefficient of variation (which is equal to the standard deviation divided by the mean) is equal to $1/\sqrt{(\text{mean})}$. One implication of this is that the higher the associated count rate the lower the coefficient of variation will be. For example; a count rate of 10,000 has an associated coefficient of variation of $1/\sqrt{10,000} = 1\ \%$.

Thus the lower the radioactivity of the source the longer the required counting period in order to maintain the same coefficient of variation. In order that the statistics be maintained and further blurring of the image does not occur (due to measurement over a finite area) the scanner would have to translate more slowly and hence the overall time of the scan would be increased. If the scan takes too long, artifacts due to patient movement become a problem.

<u>Dead Time</u>

Radiation detectors and the associated electronics have finite resolving times. After a system records a pulse it is unable to respond to another pulse for a brief period, known as the dead time. As a result the number of pulses recorded by the counting system is less than the actual number of pulses that occurred.. This means that the measured statistics of the process have been altered due to the measurement equipment.

In order to rectify this situation it is necessary to measure the dead time of the system and to use this measurement to correct the measured pulse count. For most applications system dead time is independent of the count rate. Hence, once system dead time has been established, corrections can be made for any count rate.

System dead time (p) is determined by measuring (in order):

1. the background count rate (B),
2. the count rate resulting from one source (n_A),
3. the count rate resulting from two sources (sources A & B) (n_S), and
4. the count rate resulting from source B (n_B).

If this order of measurement is followed any errors due to accurate repositioning of sources will be eliminated. If the true count rate is N and np < 0.05 then the following expression is valid (44) ;

$N= n(1+np)$

In the cases of the above measurements this results in the following three expressions;

$N_A + B = n_A(1+n_A p)$

$N_B + B = n_B(1+n_B p)$

$N_A + N_B + B = n_S(1+n_S p)$

Adding the first two equations together and subtracting the third equation from the result leads to the following expression (solving for p);

$p = (n_A + n_B - n_S - B)/(n_S^2 - n_A^2 - n_B^2)$.

Thus for a measured count rate (n) the fraction of lost counts is n(p) and hence n(n)(p) counts must be added to n to give the true number of counts N. This correction is easily performed by a computer.

Beam Hardening

Absorption of radiation is dependent on the energy of the radiation. As a polyenergetic beam passes through an object the lower energy beams are more severely attenuated than are the higher energy beams. This results in the beam being "hardened" as it is composed of a greater proportion of high energy beams the farther along it is in the object (45-48). Thus in order to determine absolute densities of an object the beam must be either monoenergetic or else non-linear corrections must be applied. Correction for beam hardening may be either material selective or non-material selective. Material selective correction takes into account the type of absorption material and the resultant hardening. This requires knowledge of the types of absorption materials present and the quantities involved and hence an initial reconstruction must first be made. Then the raw data must be manipulated according to the first reconstruction to produce the corrected image. Non-material selective beam hardening merely recognizes that beam hardening has occurred and uses a beam hardening curve to correct the raw data before reconstruction.

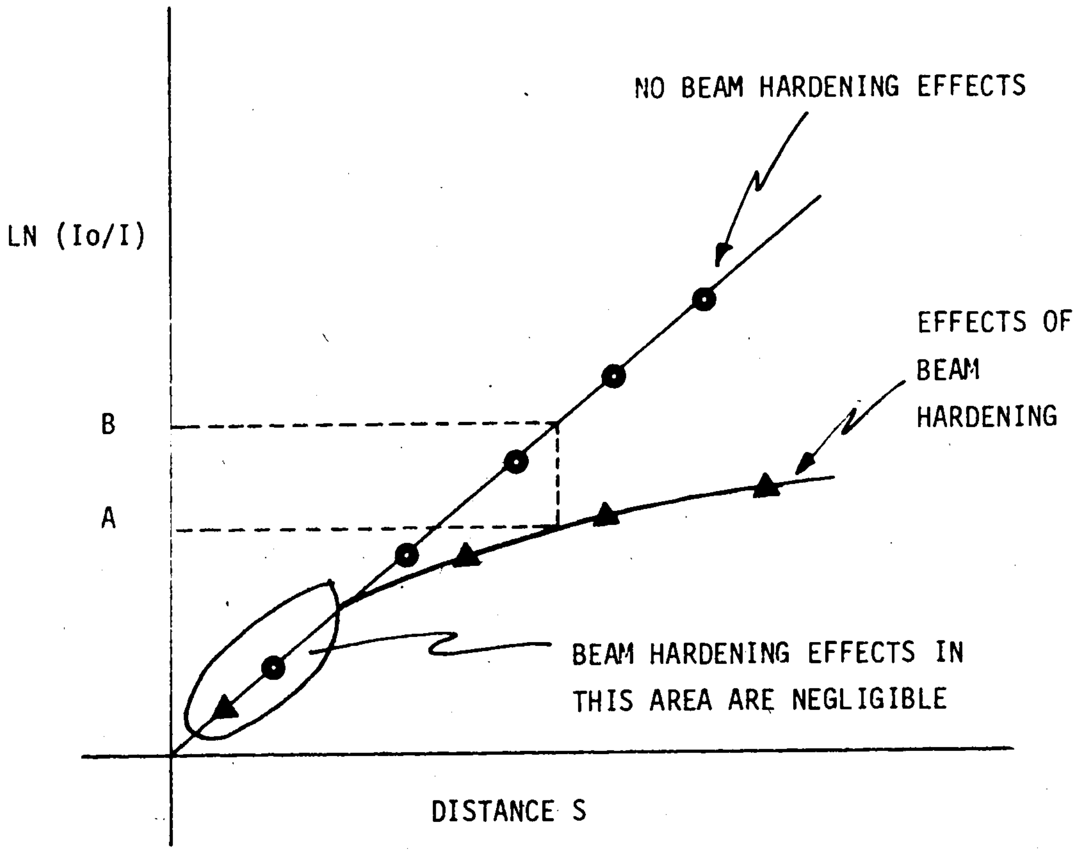
If one measures the amount of beam hardening as a function of distance then these curves may be used to correct raw data. For example, if a measurement of ln (Io/I)

= A (Figure 13) is made, the corrected value B, which is the value that would be otained if there was no beam hardening, is easily obtained from the curve. This procedure can easily be performed for all ray-integral measurements prior to reconstruction. It is interesting to note that the correction required for measurements of relatively low values of ln (Io/I) is small and hence inaccurate corrections for measurements such as through skin (as contrasted to bone) introduces negligible error. Non-material selective correction techniques for beam hardening are obviously not as accurate as are selective techniques however for objects consisting predominantly of one or two absorption materials it does provide a quick and simple correction technique.

D. Detector Types

Primary considerations in regards to selection of a detector include resolution, efficiency, energy of radiation, counting rates, and in some applications, size. Energy resolution of a detector is its ability to distinguish between different energy photons. This is often defined as a percentage or ratio of the full width half maximum divided by the primary energy level. Detector efficiency refers to the percentage of rays at a specified energy that will interact with the detector rather than being simply passed through the detector.

Three common detector types are gas, scintillation and



GIVEN A LINEAR ATTENUATION COEFFICIENT μ

$LN(Io/I) = \mu x S$ OR $I = Io\ EXP(-\mu x S)$

FIGURE 13. BEAM HARDENING CORRECTION

solid state, all of which work on the principle of photons exciting the detector material resulting in the production of ions, optical radiation and electrons respectively. These phenomena are then converted to electronic signals, the magnitude of which are proportional to the photon energy, and the frequency of which are proportional to the number of photons per second striking the detector.

Solid state detectors have better resolution than do the other two types. They are also much more expensive and hence are only used in applications where resolution is critical.

Gas detectors although usually having worse resolution and efficiency than scintillation detectors do have the advantage of generally being more economical and smaller in size. It is for this reason that most fan-beam machines use xenon gas detectors (7) instead of scintillation detectors. Scintillation detectors however are economical, readily available and are easy to use. These factors coupled with reasonable resolution, efficiency, and counting rates have led to the utilization of scintillation detectors for many spectroscopic applications.

Scintillation Detector

A scintillation detector is composed of two basic parts : (1) a scintillation crystal and (2) a photomultiplier tube.

Scintillation Crystal

The scintillation crystal is a transducer that converts photons of energy to light pulses of proportional intensity to the energy of the photon. There are both in-organic and organic materials in use for this purpose. Sodium Iodide crystals NaI(Tl) are the most commonly employed crystals for gamma-ray spectroscopy. It is desirable to have a high ratio of counts under the full energy peak relative to the total number of counts and this ratio is approximately four times greater for Sodium Iodide than for example Cesium Iodide crystals CsI(Tl), with the same gamma-ray primary energy. This ratio is also increased by the use of collimators which restrict radiation to the central part of the crystal. Use of a Thalium impurity (0.1 %) acts as a wavelength shifter causing the crystal to emit 2 or more low energy photons (visible light) instead of a single Ultra Violet photon. UV photons are not suitable since they are absorbed by most materials even those transparent to visible light. At low energies, typical resolution for NaI crystals ranges from 6-8 % for crystals under 5 cm diameter to 8-10 % for crystals from 5-10 cm diameter. Thicker crystals provide more efficient light collection and increase crystal efficiency so that the same crystal may be used with higher energy sources. If the crystal is too thin high energy rays may not interact with crystal.

Photomultiplier Tube

A photoemissive cathode detects the light pulses produced by the crystal and in response produces primary electrons. These electrons are electrostatically focused and accelerated by means of a voltage differential to the first dynode in the tube. Upon arrival at the dynode they possess enough energy so that for every primary electron more than one secondary electron is produced. This procedure is continued throughout an arrangement of dynodes (typically 10) until the final stage which is the anode. The voltage at the anode following the passage of a charged particle through the phosphor of the crystal is equal to the charge of the electrons at the anode divided by the capacitance of the anode and is proportional to the intensity of the light impulse over a wide range of photomultiplier gain. Head on type PMT's have a semitransparent photocathode which is in contact with the inside of the glass window. Head-on versus side-on photomultiplier tubes are generally used for scintillation counting due to improved uniformity and collection efficiency.

Anode Load

The signal on the anode has a rise time which corresponds to the fluorescence decay time of the crystal. This signal decays through the anode load R_L (Figure 14) which is the resistor between the anode and the high voltage supply and the parallel combination of the anode to ground capacitance (typically 10 pf) and the output line capacitance to ground (typically 10 pf).

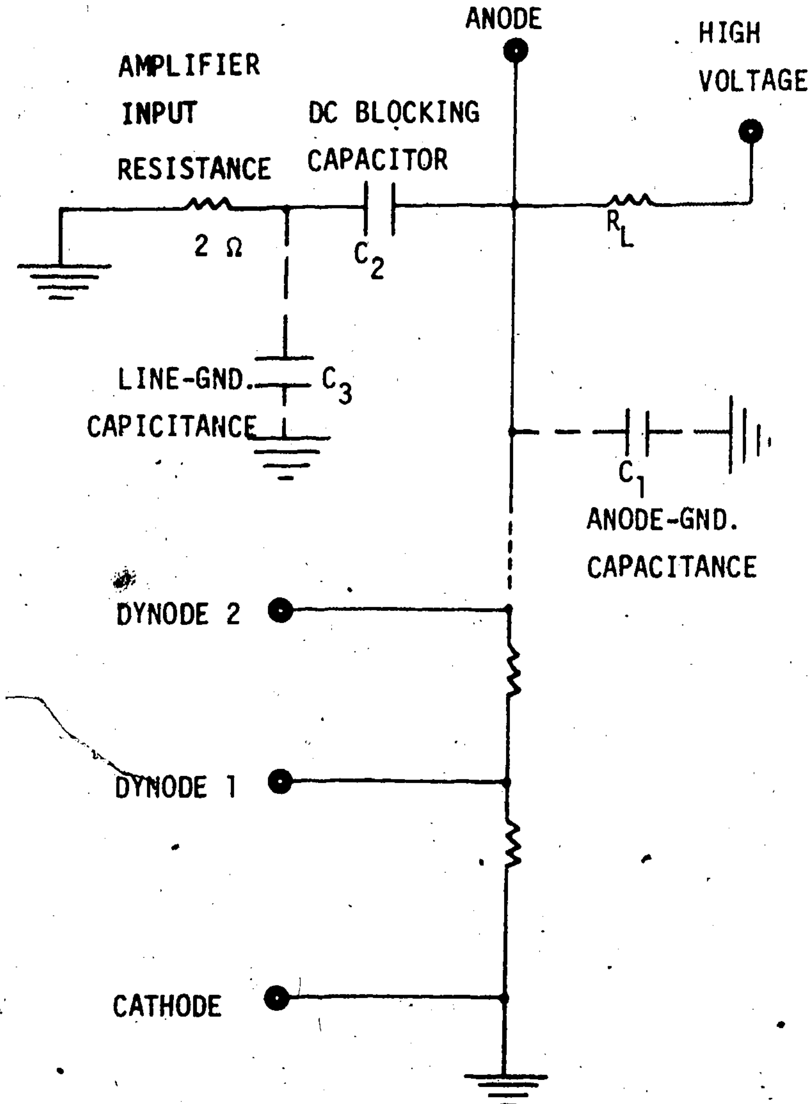


FIGURE 14. PHOTOMULTIPLIER TUBE & RELATED CIRCUITRY

The capacitances are virtually fixed values and are not easily changed and it is only the anode load that is easily altered. The signal on the anode must decay away fast enough so that pulse pileup is not a problem and slow enough so that pulse height is significantly larger than existing noise. Pulse pileup occurs when signals do not have enough time to decay away and the next signal is superimposed on the partially decayed signal.

E. Performance Characteristics Of CT Scanners

Prior to this point it has been the component parts of a CT system that have been reviewed. This section examines how some attributes of the overall system are related to these distinct segments.

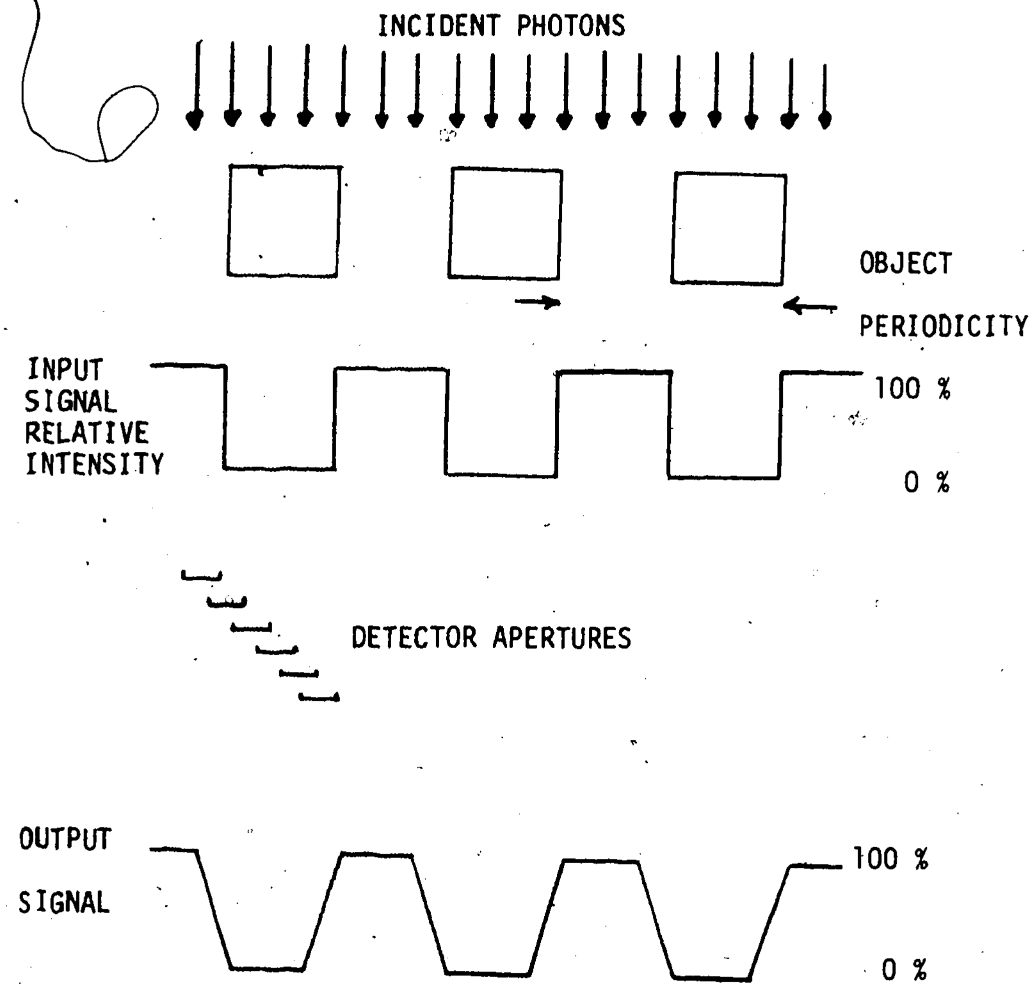
Resolution of a CT system comprises two distinct, although interrelated types of resolution: spatial resolution and contrast resolution. Spatial resolution of a CT system refers to the minimum size object that can be imaged by the system when that object is part of a periodic structure. Contrast resolution is that percent change in contrast at an interface that can still be imaged given a minimum spatial resolution.

Spatial Resolution

Assuming a constant photon flux there are four major factors which affect the spatial resolution of a CT system (49). These are the width of the detector element aperture, the distance between reading or sampling points, the form of the convolution filter used, and the element size of the display picture (pixel size). In order to examine the spatial resolution of a system it is useful to employ a concept known as the Modulation Transfer Function (MTF). Values of MTF vary from 0 to 1 and are an indication of how well the imaging system transfers the frequency components of the [illegible] to the final image. An MTF value of 1 (or 0) means th[illegible] none) of the frequencies present in the orig[illegible] exist in the final image. Good spatial resolution then, would be indicated by large values of the MTF extending into the high frequency range.

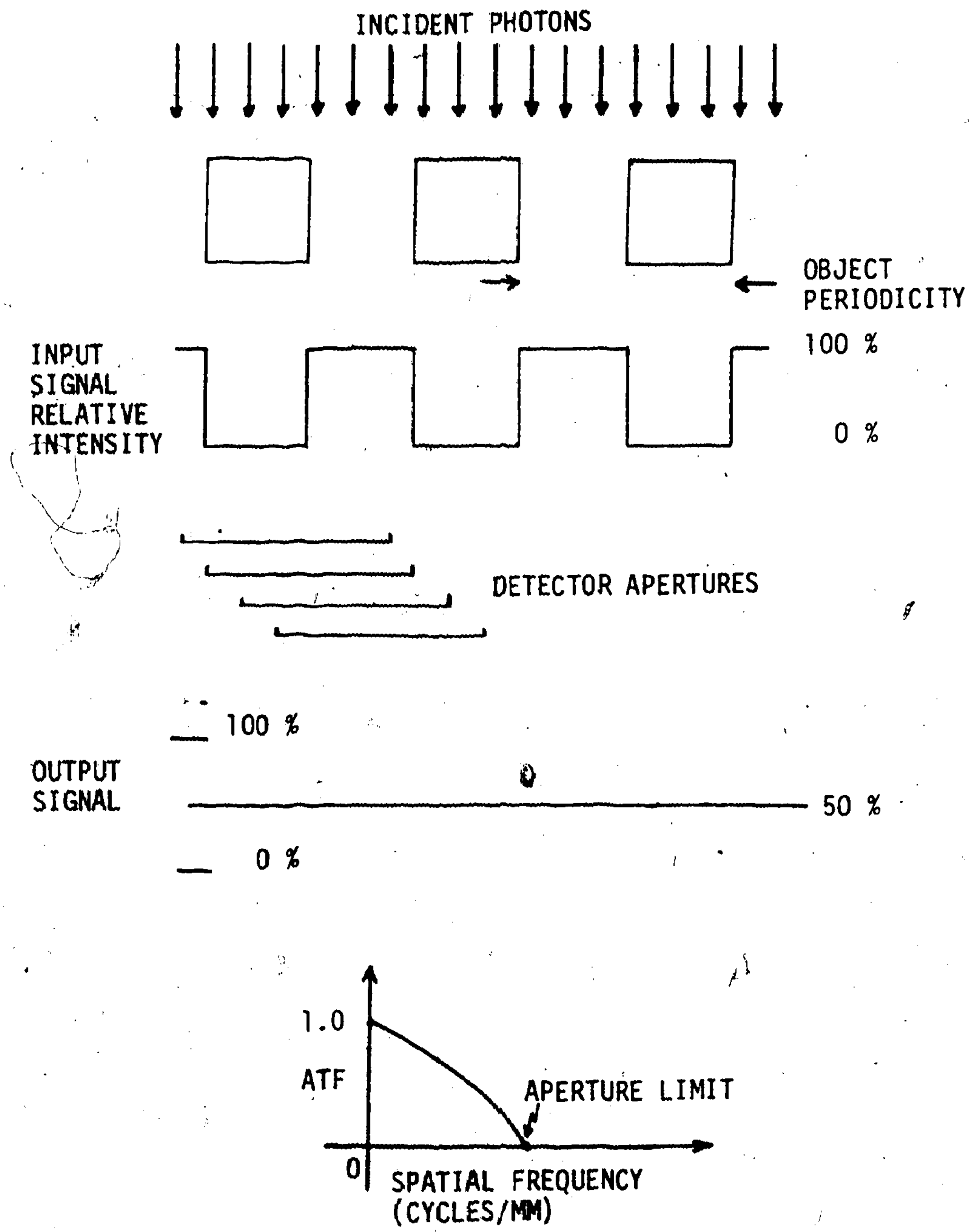
Aperture Size

The effect of aperture size can be seen by examining the results obtained by viewing a series of repeating objects through different size apertures while holding all other variables constant. From Figure 15 we see that when the size of the aperture is much smaller than that of the object spacing, the amplitude of the output signal approaches that of the input signal. Thus the aperture limited MTF or the ATF would approach unity. As can be seen from Figure 16, if the aperture size is increased, the amplitude of the output signal becomes smaller than that of the input signal and the value of the ATF is decreased. When the size of the apertures approach twice that of the object periodicity, all information is lost and the ATF approaches zero. The net effect is: the smaller the aperture, given other



WHEN THE OBJECT SPACING IS MUCH GREATER THAN THE DETECTOR APERTURE THE OUTPUT SIGNAL CLOSELY RESEMBLES THAT OF THE INPUT SIGNAL AND THE ATF APPROACHES 1.

FIGURE 15.



WHEN THE OBJECT SPACING IS EQUAL TO THE DETECTOR APERTURE THE OUTPUT SIGNAL CONTAINS NO INFORMATION AND THE ATF IS 0.

FIGURE 16.

parameters constant, the greater the spatial resolution of the system.

Sampling Rate

According to the Nyquist theorem, in order to accurately reconstruct an object that is part of a periodic structure, the sampling rate must be at least twice that of the highest frequency component in the structure. The tradeoff associated with sampling at high frequencies is a resultant increase in the noise of the reconstructed image (33). If the image contains frequencies greater than the Nyquist frequency, aliasing will occur resulting in streaks in the image emanating from such sources as bone edges which contain a number of high frequency components.

Convolution Filter

It has been found (49) that the higher the frequency response of the convolution filter used in the reconstruction, the greater the resultant spatial resolution. The problem is that the higher the frequency response of the filter, the greater the resultant noise of the reconstructed image and hence a tradeoff is required.

<u>Contrast Resolution</u>

Another area of importance to CT scanners is their contrast resolution. This is measured by examining the system limiting spatial resolution with different contrast media. The limiting factor of CT systems in regards to contrast resolution is the presence of noise. The signal to noise ratio can be improved by increasing the incident photon fluence rate (reflux). However, increasing radioactivity will result in a corresponding increase in patient dosage. One measure of contrast resolution used for commercial scanners (49) is obtained by multiplying the hole diameter (in mm) by the contrast (in percent) and the square root of the dose (in rads). Inclusion of the dose in this calculation results in a measure of contrast resolution independent of dose rate.

F. Commercial CT Scanners

As mentioned previously it is necessary to make alterations to commercial CT scanners before they can be effectively used for measurement of bone mineral density. Having examined some of the theoretical concepts of CT scanning an examination of the differences between a commercial CT scanner and one designed for measurement of bone mineral density will now be given.

Commercial whole body CT scanners are both large and expensive: typically $600,000 - $1,000,000 and the source of radiation is virtually always an X-ray tube. These scanners are oriented towards measurement of soft tissues such as the abdomen. In these areas the application of the scanner is generally to locate tumors which will be depicted as an area of density different from that of the surrounding region. Absolute densities are generally not of concern so much as the ability to distinguish between areas of different densities. Commercial CT scanners are typically fourth generation scanners allowing scans to be performed in under five seconds. Increased speed reduces the problem of movement artifacts and means less inconvenience to the patient.

In transmission CT the source of radiation may be either an X-ray tube or a radioisotope. An X-ray tube is advantageous over a radioisotope due to a greater photon flux. They do however have the disadvantages of a high initial cost ($30,000), large physical size and the requirement of a large amount of calibration and adjustment. There is also the possibility of a higher dose rate of radiation with an X-ray tube source.

There are also many problems that actually preclude the use of commercial CT scanners for measurement of bone density. The major problem is manifest in the area of system spatial resolution. Use of a multi-energy X-ray source means that beam hardening may be a major problem and the fact that such scanners are looking at relative rather than absolute density means that the data required for beam hardening correction is often not readily available. This is further complicated by the fact that the detectors used in commercial CT scanners are not operated in a pulse counting mode and hence they cannot provide energy discrimination of detected photons. Sources and detectors are usually not finely enough collimated to provide the initial physical resolution and the size of the reconstruction matrix usually means that the area associated with an object the size of a bone is too small for an accurate determination of average density of the ar This is due to the fact that the reconstruction matrix of such a system is typically required to cover an area the size of an abdominal cross-section.

Commercial CT scanners could be used for the measurement of bone mineral density. However, they would first require recollimation of the source and detectors and software would have to be altered to correct for beam hardening and to increase the matrix size of the area of analysis. These changes would require a significant disruption of an expensive piece of equipment: a disruption not presently justified given the alternative use of a smaller, radioisotope CT scanner.

IV. System Design and Construction

This section will cover five areas of design and construction of the scanner. These include: (1) mechanical operation, (2) radioactive source and detectors, (3) scanner control, (4) adjustments required for a multi-detector system, and (5) data processing. The initial system design and construction were for a generation one CT system. Subsequent to the successful completion of the generation one system the necessary modifications to convert to a generation two system were undertaken.

A. Mechanical Operations

The major design requirement insofar as mechanical operation of this device is concerned, is the ability to properly position the source-detector configuration in both the longitudinal and rotational axes.

Stand

The stand (Figure 17) was designed to contain the controls such as the microprocessor unit, the power supplies, relays, source and detectors on one portable unit. The ability to position the scanner at different heights was required so as to facilitate measurement of both arms and legs. This was accomplished by mounting the scanner plates on a board with a 12 inch diameter hole cut in the center which slides vertically on ball bushings and is driven by a hand operated ball screw.

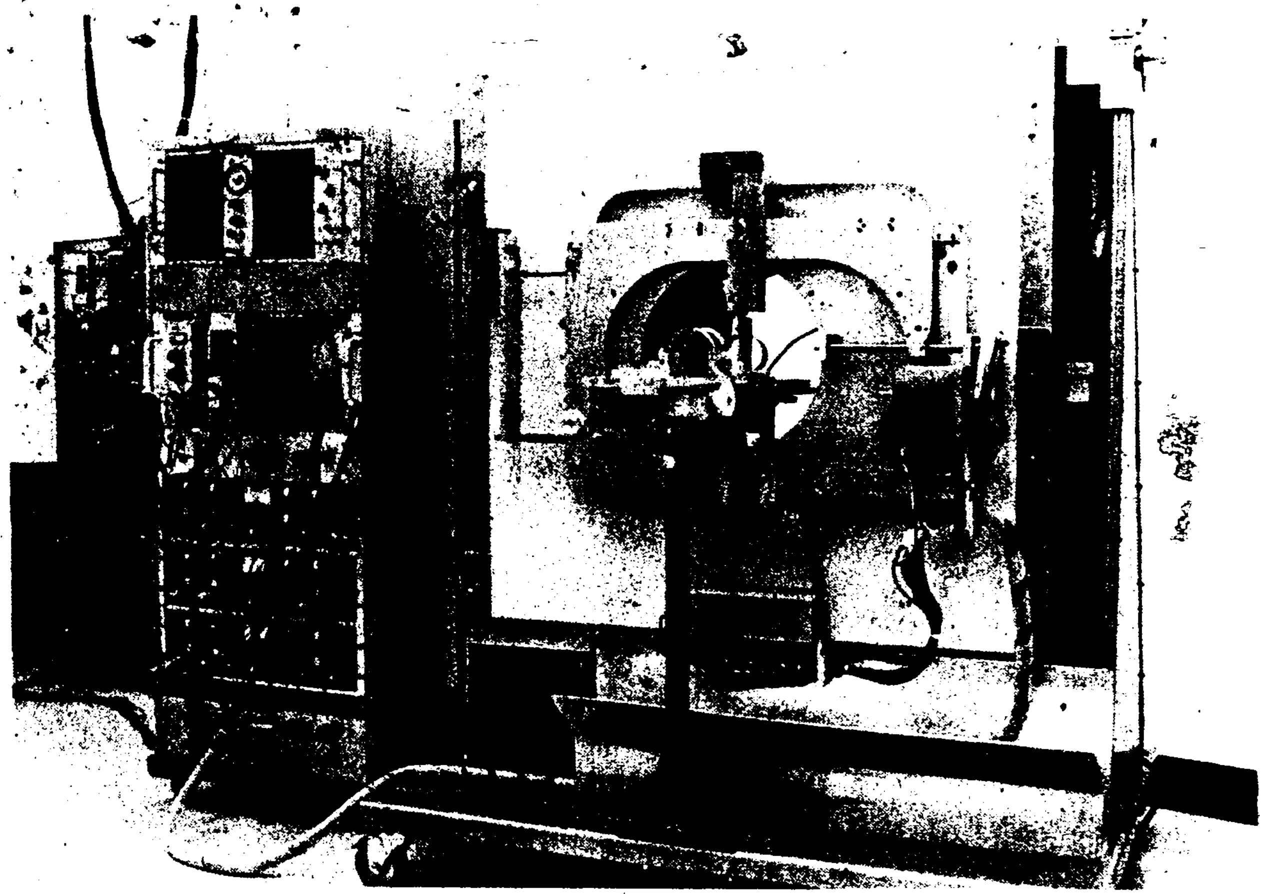


FIGURE 17. CT SYSTEM

Source-Detector Alignment and Motion

The source and detectors are mounted on the top and bottom edge of one side of a 1/2 inch thick aluminum plate with an elongated 12 inch circular hole cut in the center (Figure 18). The geometry of the source with respect to the detectors is fixed. This first plate is mounted upon a second plate (Figure 19) of 1/2 inch thick aluminum with a 12 inch diameter circular hole cut into it. The top plate is attached to the second plate by means of Thomson bearings mounted on the first plate being guided along case hardened steel rails fastened to the second plate. This arrangement allows the first plate (and hence the source-detector arrangement) to translate linearly with respect to the second plate. The second plate is mounted onto a third plate (Figure 20) which contains a circular bearing with an external gear drive which is used to rotate the source-detector arrangment about the object. These plates and the rotational bearing are mounted onto the stand's board.

Mechanical Drives

The drive for the translation is a stepping motor mounted on the second plate. This motor is used to drive a belt connected precision ball screw arrangement which is attached to the first plate. Each revolution of the ball screw results in a 2.5 mm motion of the plate. Since, in the half-step mode, the motor makes one revolution every 400 steps the resolution of one step is 2.5 mm/400 steps or

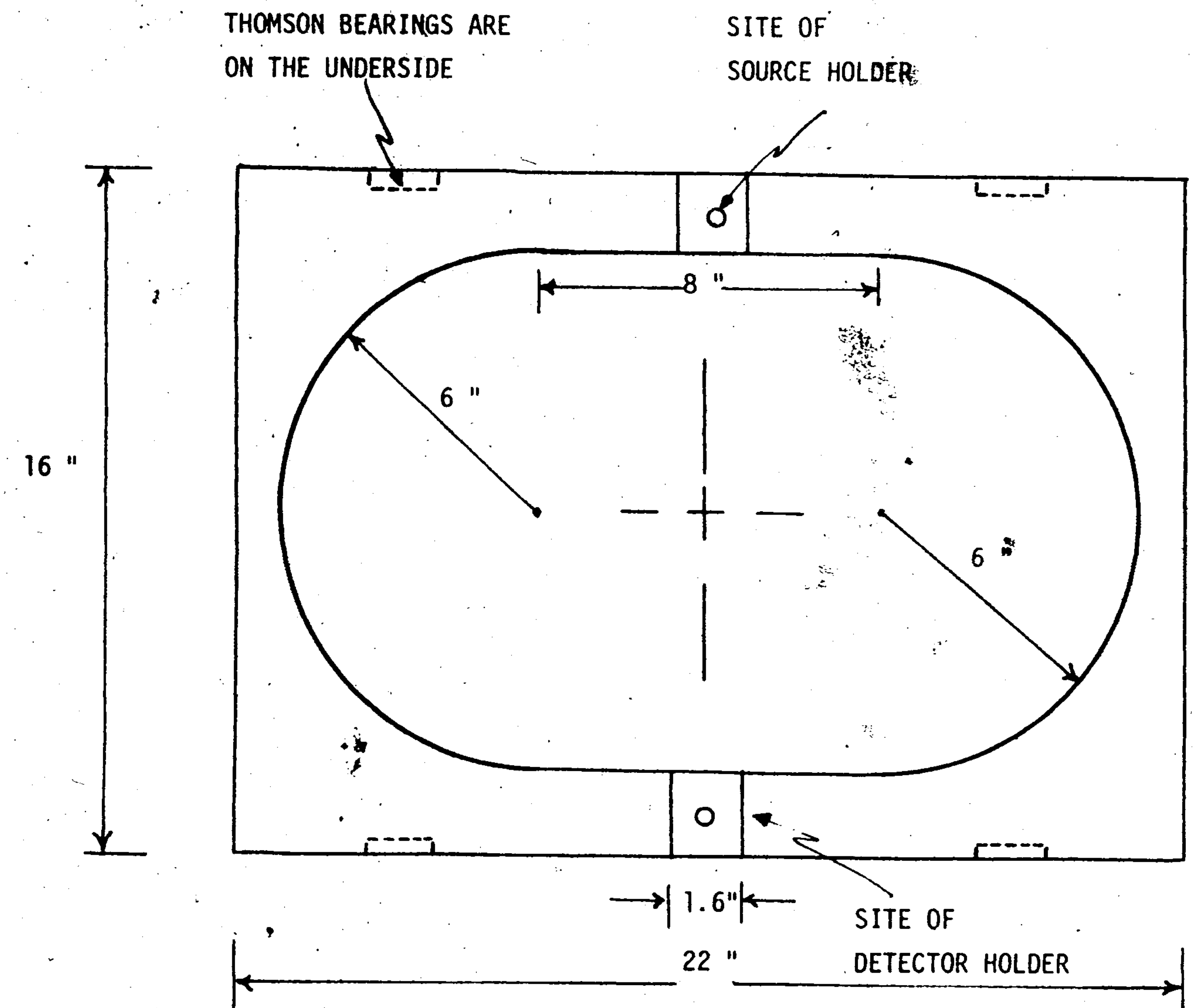


FIGURE 18. PLATE #1

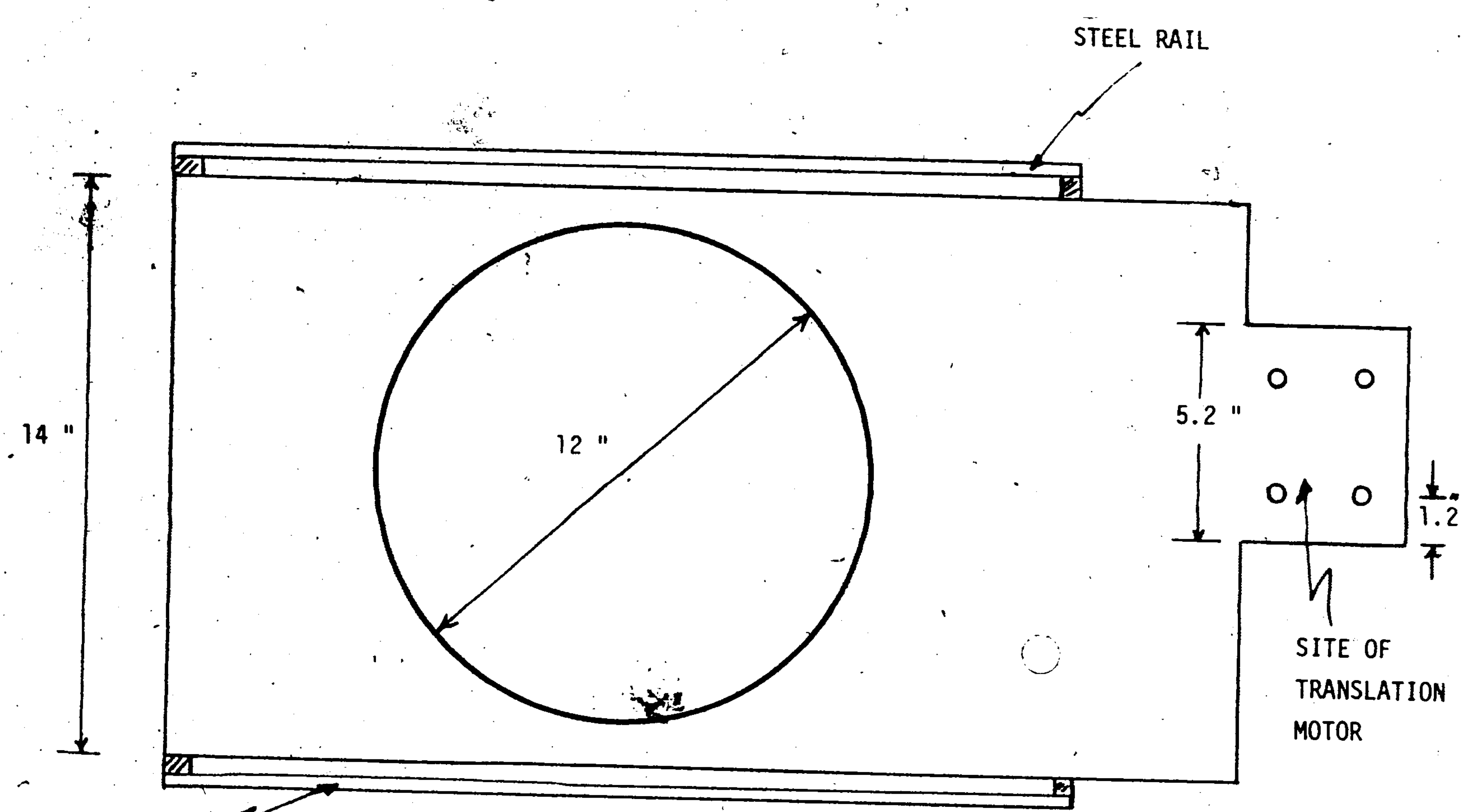


FIGURE 19. PLATE #2

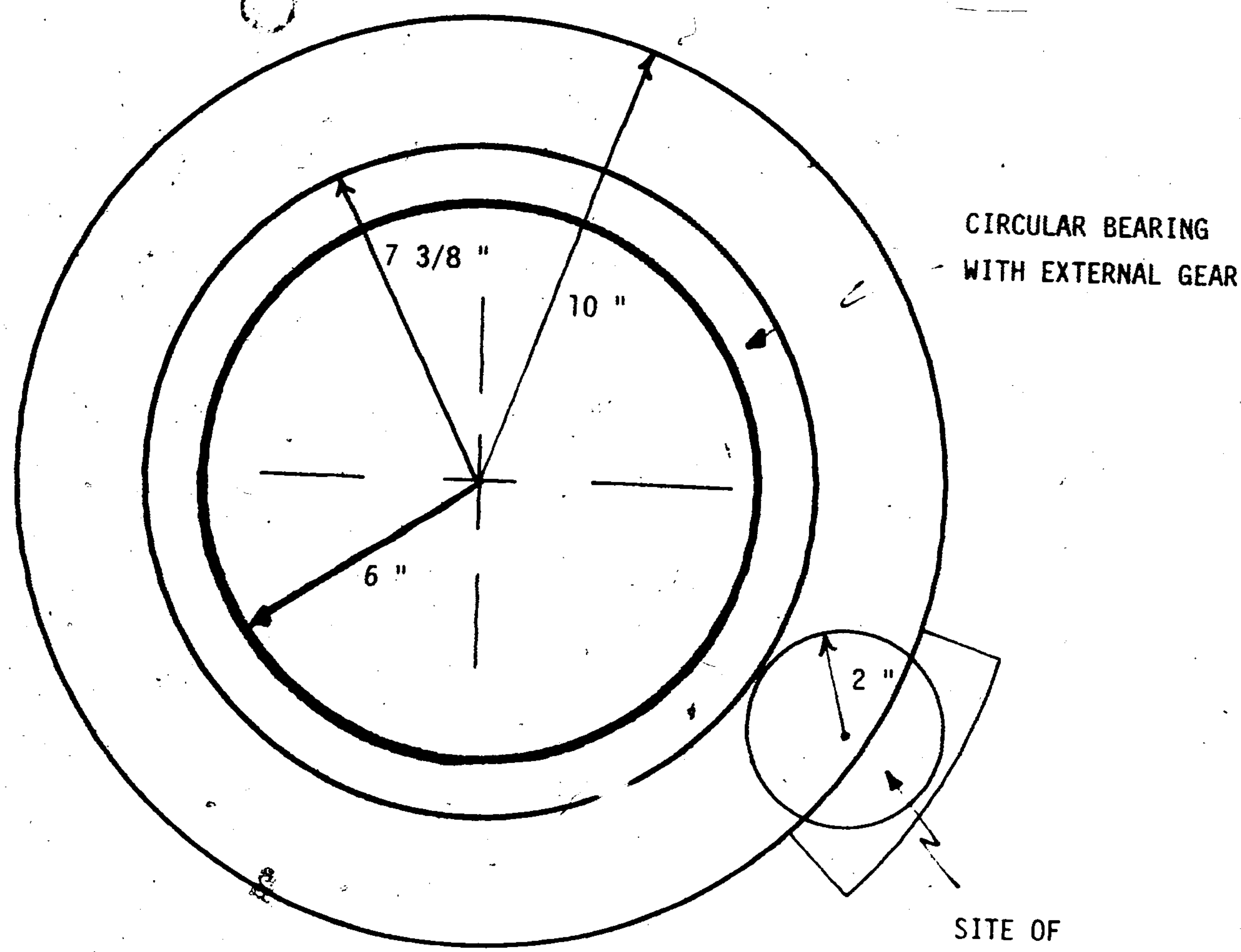


FIGURE 20. PLATE #3

6.25 um/step. The ball screw is of such a high precision and hence expense that a number of microswitch controlled relays have been installed which shut off power to the stepping motors should the scanner be operated outside its normal scanning range. The plates are rotated by a second stepping motor which drives a worm gear arrangement that is mechanically linked to a spur gear which in turn drives the external gear on the circular bearing. The worm gear was necessary in order to reduce gear backlash caused by the stepping nature of the motors. The gearing ratios mean that approximately 200 steps of 0.9 degrees each are required at the motor shaft to induce a 1 degree rotation of the plates. All of the gearing ratios in the system were known exactly except the gearing ratio of the worm gear arrangement. This ratio was determined by counting the number of motor steps required to rotate the plates 360 degrees as measured by a dial gage with a resolution of 0.0005 inches.

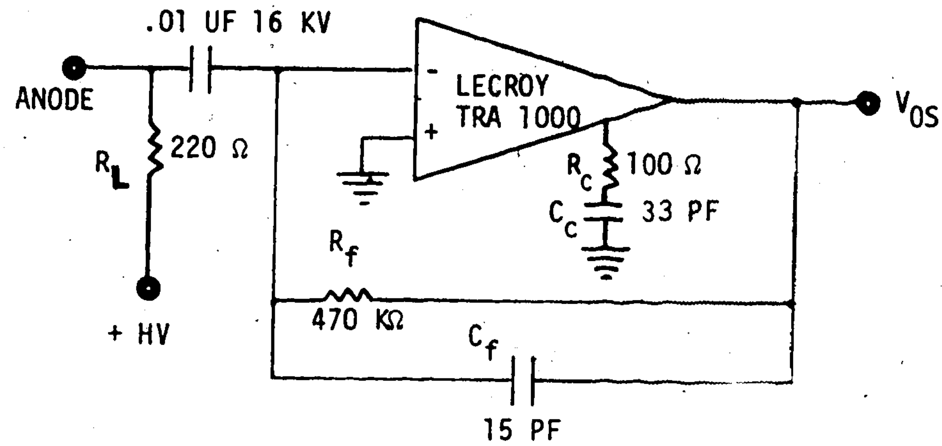
B. Radioactive Sources and Detectors

The underlying principle behind any tomographic scanner is the fact that if attenuation of a photon beam can be measured along a sufficient number of pathways through an object then the radiological density at any point within the object can be determined. The radioisotope photon source is 1.5 Curies of Iodine 125 adsorbed to a 2 mm diameter zeolite bead and encapsulated in aluminum. The detectors are 1/2 inch diameter, 1/2 inch thick NaI(Tl) crystals coupled to 13 mm diameter photomultiplier tubes with borosilicate windows and bialkali dynodes. The detectors were chosen predominantly on the basis of availability and ease of operation. A radioisotope was selected due to financial constraints and ease of operation. The half-life of Iodine 125 is 60.2 days and the source is replaced every 2-3 half-lives at a cost of approximately $1400/source. The mechanism of radioactive decay of Iodine 125 is included in the appendix but basically a gamma-ray of approximately 27.5 Kev is produced during the decay process. This energy, being relatively low, has a comparatively large linear attenuation coefficient meaning that subtle differences in density are discernible. At this energy the total attenuation coefficient for NaI is approximately 23 per cm. Hence in a 1/2 inch thick NaI(Tl) crystal only 2.06×10^{-13} of the photons would be passed through the crystal. For Gadolinium 153 with an upper energy of 100 Kev the attenuation coefficient for NaI is 6 per cm. and the number of photons passing through a 1/2 inch crystal would be 4.91×10^{-4} of those incident upon the crystal.

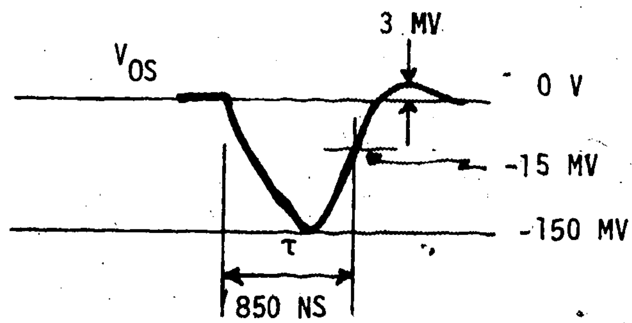
Signal Processing

From the anode of the photomultiplier the signal is fed into a preamp (Figure 21). Since a NaI(Tl) crystal is used the rise time of the signal corresponds to the fluorescent decay time of Sodium Iodide (250 ns). Using a trial and error technique an anode load of 220 ohms was determined to be suitable in terms of pulse height and pulse pileup. The

PREAMP CIRCUIT USED WITH 8 DETECTORS AND NARROW HV CABLES



OUTPUT SIGNAL WITH ^{125}I SOURCE



NOTE: -15 MV IS THE UPDATING POINT OF THE LECROY 623 DISCRIMINATOR.

DECREASING R_1 SHORTENS PULSE WIDTH AND SLIGHTLY DECREASES PULSE HEIGHT.

I.E. IF R_1 IS CHANGED FROM 3.3 KΩ TO 220 Ω: V -300 MV TO -150 MV
τ 2.2 US TO 0.85 US
V_{OS} +18 MV TO + 3 MV

DECREASING C_c AND C_f WILL DECREASE τ BUT ALSO WILL CAUSE OSCILLATION.

THEREFORE, C_c AND C_f WERE CHOSEN AS SMALL AS POSSIBLE WITHOUT OSCILLATION.

FIGURE 21. PREAMP CIRCUIT

preamp then shapes the signal, increases its amplitude and transforms the impedance. The preamp used is a Le Croy TRA 1000 preamp and is able to provide a signal of sufficient amplitude so that a linear amplifier is not required. This particular preamp was chosen because of its low input noise (30 pA/ sqrt (Hz) r.m.s.) and it· frequency response (20 MHz at 1 mV/uA gain). The output of the preamp goes into a Le Croy Model 623 8-channel discriminator which provides an output pulse for every input pulse over a preset energy. Thus lower energy pulses are discriminated against and only pulses above a certain energy are counted. From the discriminators the output pulses are sent to a CAMAC 50 M Hz Scaler which counts the number of pulses received in a given interval. This counting interval can be set by the HP 2100 computer which is interfaced to the CAMAC.

C. Scanner Control

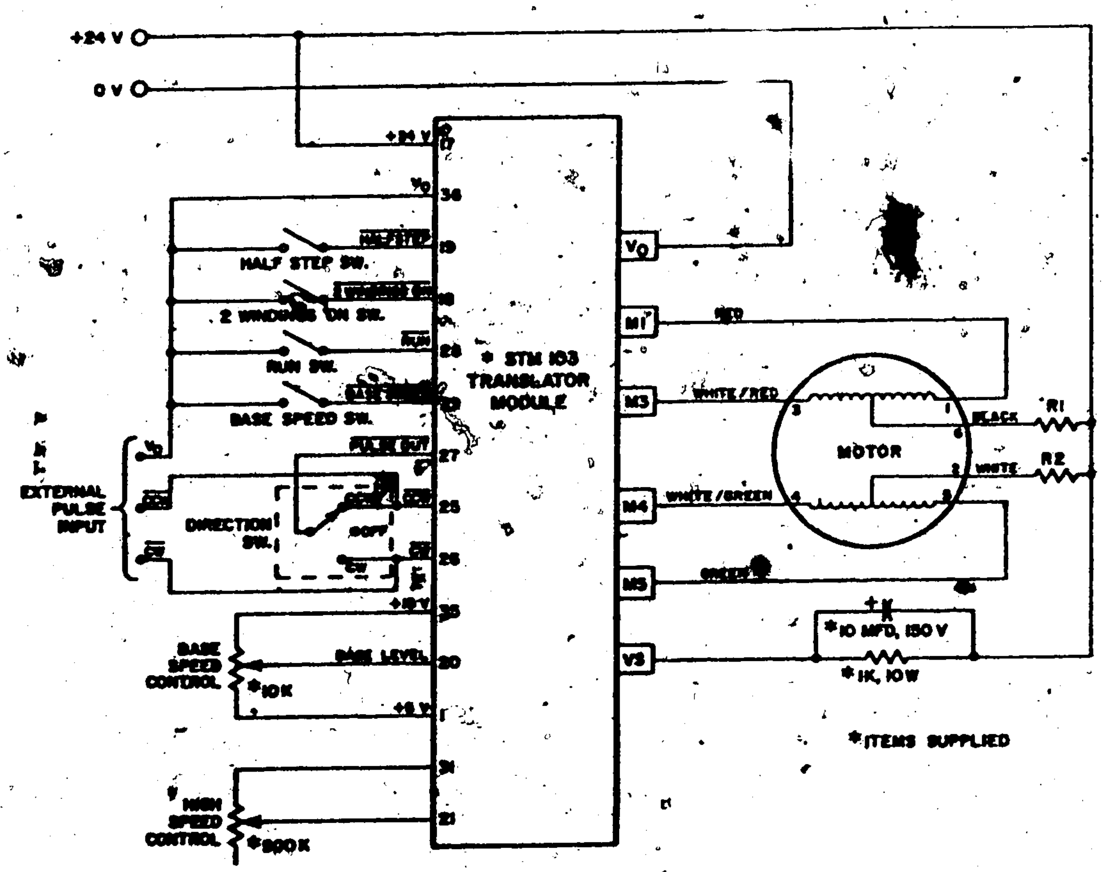
The drive motors are two Slo-Syn model M092 FC09 stepping motors with nominal torque ratings of 200 oz inches. A stepping motor translates an electrical pulse into a precise mechanical motion of the shaft. This motion is in fixed, repeatable increments permitting accurate positioning of the motor shaft. The motors can move in either 0.9 degree or 1.8 degree increments per input pulse. The position of the shaft is to within 3 % accuracy and this error is noncumulative from step to step. These motors require a DC power supply and associated electronic drives which contain the required circuitry to convert pulses into the proper switching sequences for the motor. The power supplies are 2 Slo-Syn MPS3000 24 V DC while the electronic drives are 2 Slo-Syn STM103 3000 step per second motor controllers (Figure 22). These controllers have internal oscillators that can be used to drive the motors in either direction and either in the half-step (0.9 degrees per step) or the full-step (1.8 degrees per step) modes. They are also able to receive pulses from external logic devices such as microprocessors and minicomputers. In the scanner external logic sources include a Motorola microprocessor unit (MPU) and a monostable multivibrator which is used to provide single stepping capability.

The control system for this scanner was designed keeping in mind available equipment and computing facilities (Figure 23). A Motorola MEK 6800 D2 microporocessor unit (MPU) provides the central clock for the system. It is used to control the two stepping motors which provide the rotational and translational motions, as well as to signal the computer when to collect data from the scalers. All MPU software programs have been assembled and are contained in the appendix.

Timing Considerations

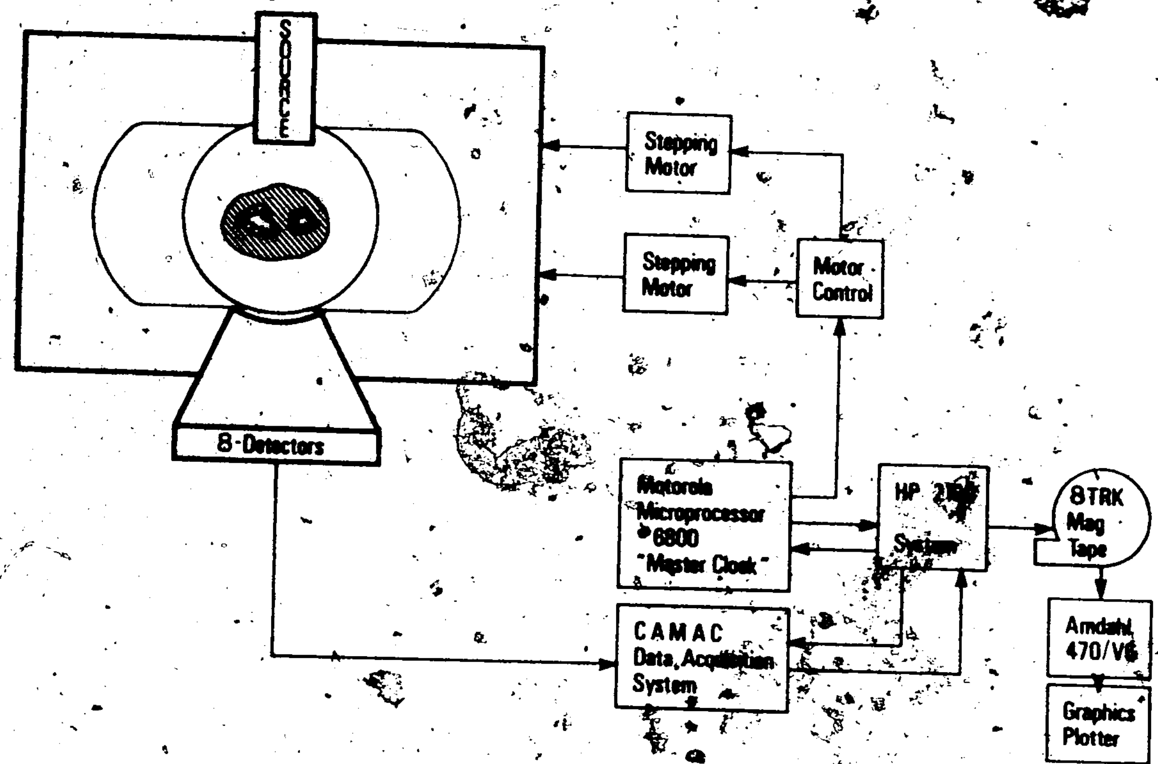
Since the microprocessor unit is used to control the stepping motors and hence scanner position it is also necessary that the MPU clock be used to determine when data from the detectors should be collected. Upon receipt of a

STEPPING MOTOR AND TRANSLATOR CIRCUIT DIAGRAM



CONNECTION DIAGRAM

FIGURE 22.



General System Philosophy: Data Acquisition

FIGURE 23. SYSTEM BLOCK DIAGRAM

signal from the MPU the HP must respond quickly to the signal and collect data since the scanner is continually moving. Although it was hoped that this could be accomplished with the HP running under its normal multi-user operating system it was found necessary to operate the HP in a stand alone mode while operating the CT system. This was due to the fact that under the multi-user system the time from a data-ready signal (from the MPU) until data was actually collected from the scalers was variable over too great a range. The stand-alone mode allows data to be collected within 4 microseconds (2 HP clock pulses) of a signal from the MPU.

<u>MPU Control</u>

The Motorola MPU is the basic D2 kit with the only additions being that of 1/2 K of Ram (M6810) and 2 K of EPROM (Intel P2708). Program development and EPROM loading were done on an AMI microcomputer development center.

The choice of the 8 bit Motorola as the control processor was based upon:

1. availability of parts,
2. price, and
3. facilities for EPROM programming.

Although the trend is toward 16 bit and 32 bit microprocessors it was felt that the 8 bit microprocessor was sufficient for this particular application and this combined with the aforementioned constraints led to the selection of the M 6800.

To facilitate insertion and removal of the object to be scanned the scanner must begin and end operation in a "park" position which may not coincide with the actual scanning path. The sequence of events which occur during a scan are as follows:

1. the MPU determines the park to start distance and direction and sends the appropriate pulses to the motor controllers;
2. a signal is sent to the HP signifying that a scan is to begin;
3. a signal is returned from the HP to the microcomputer unit signifying that the HP is ready and that a scan may begin;
4. a signal is sent from the MPU to the HP signifying that a scan has been started and that acceleration of the translational motor has begun;
5. a second signal sent to the HP from the MPU signifies that acceleration has ceased and that translation velocity has been reached;
6. all translation pulses are sent to the motor controllers from the MPU;
7. pulses are sent from the MPU throughout the scan to the HP signifying when to collect data from the detectors;
8. after translation is complete the MPU decelerates the translational motor, determines the required rotation and sends the appropriate pulses to the motor;
9. the pattern (2-8) is repeated until the appropriate

number of rotations and translations have been performed;

10 at which point the appropriate pulses are sent to the motors so that the scanner is returned to the park position.

Acceleration and deceleration of the stepping motors is required since motor torque decreases with increasing speed. Thus in order to ensure that all pulses sent to the motor result in a step while at the same time operating at the maximum allowable speed, acceleration and deceleration of the motors is needed. This is accomplished by the microprocessor which uses a subroutine ASD (see appendi[illegible] provide a series of pulse trains at varying freque[illegible]

MPU Based PIA

The microprocessor unit (MPU) is interfaced to the motor controllers and the HP by an on-board integrated circuit (M 6820 PIA - peripheral interface adapter). The PIA contains an 8-bit bidirectional data bus for communication with the MPU, two bidirectional 8-bit buses (output registers) for interface to peripherals, two programmable control registers, two programmable data direction registers and four individually-controlled interrupt input lines; two of which are usable as peripheral control outputs. The MEK 6800 D2 kit contains two PIAs, one of which is used to interface the keyboard. The second PIA appears to the MPU as four memory locations located at \$8004 - \$8007 (where "$" denotes a hexadecimal number). The output register (OR) and data direction register (DDR) of port A (one 8-bit bus used for interfacing) are located at $8004 and the control register (CR) is located at $8005 while the corresponding registers for port B are located at $8006 and $8007. The contents of the control registers determine the operation mode of the interrupt lines and whether an output register or data direction register is to be addressed. This latter feature is required since the OR and DDR share the same address location. The data direction register determines whether the corresponding interface lines to the output register are inputs (DDR=0) or outputs (DDR=1).

To control the scanner 6 lines of the PIA port A are used. Two lines are needed for each of the two stepping motors (bidirectional) and two lines are used in a handshake mode between the HP and the MPU. The MPU is initialized by a subroutine (PIA) which first sets control register A (CRA) permitting DDRA to be addressed. DDRA is then set to $1F (bit 7 is the most significant bit) so that lines 0-4 of ORA are outputs while lines 5-7 of ORA are inputs. Control register A is then reset so that ORA can be addressed. The normal state of the output register is set equal to $1F (only bits 0-4 may be set by the MPU). Signals to the translation motor are sent along lines 3 and 4 ($0F and $17) while signals to the rotational motor are sent along lines 0 and 1 ($1D and $1E). A signal to the HP is sent on line 2 ($1B) and a signal from the HP is received on line 5. Bits 0-4 are active on the positive transition meaning that subsequent to the normal state of the output register the desired signal must be stored in ORA and then the normal state must again be stored in ORA at which point the message is transmitted. This pattern was used since it was found necessary to initialize the output register at the start of any scan in order to be certain of the contents of the register. The signal from the HP is level sensitive and responds to a positive signal in bit 5.

Acceleration - Deceleration Ramp

When operating the stepping motor from the internal oscillator, the translator module automatically accelerates and decelerates the motor to prevent the motor from missing steps or overshooting when stopping. This ramping of the pulse rate is also necessary when operating the translator module from pulses supplied by an external source (in this case the MPU). When using the internal oscillator, acceleration and deceleration times to a high speed are independantly adjustable from 50 milliseconds to 1 second. The acceleration and deceleration ramps are used during the transition from a preset although adjustable base speed to a preset although adjustable high speed.

In order that the number of steps and time for acceleration and deceleration be kept to a minimum known quantity a suitable MPU controlled ramp had to be found. The base speed of the control module was set to zero and using the internal oscillator a suitable ramp for a high speed of 1500 Hz was obtained through trial and error. It was found that the same ramp was suitable for use with both stepping motors. The upper speed of 1500 Hz was selected since it had been decided that the maximum speed of the motors during a rotation or a translation would be at approximately that frequency. Once a suitable ramp had been obtained these pulses were then measured with an Ortec ratemeter (which converted the pulse frequency to a proportional voltage): the output from which was displayed on a storage oscilloscope. Thus acceleration via the internal oscillator produced a ramp on the oscilloscope representative of the accelerating ramp used. The next step was to mimic this ramp using the MPU as the pulse source. A subroutine (ASD - acceleration, speed, deceleration) was developed that produces twelve ramps of varying lengths and frequencies. As a test of the MPU generated ramp the MPU output was also measured with the Ortec ratemeter and displayed on the oscilloscope and the resultant waveform was compared with that obtained from the internal oscillator of the control module. Once a close approximation had been found the MPU based ramp was tested on the scanner.

The frequencies, corresponding ramp lengths, and type and direction of motion are loaded by ASD from RAM and must be set in RAM prior to calling ASD. These parameters may be set so that ASD can be used for only acceleration or deceleration or to preset the scanner location. A count of all pulses sent during subroutine ASD is made. A number of smaller subroutines are used in conjunction with ASD. These include various acceleration and deceleration subroutines which set the parameters in RAM so that ASD may be used to accelerate to or from a certain speed. There are three deceleration subroutines (DECEL, DECEL32, DECEL5) which are used to decelerate the scanner from speeds of less than 1500 Hz, less than 400 Hz and less than 100 Hz respectively. The respective lengths of these three ramps are 200 steps, 32 steps and 5 steps. The upper speed limit of acceleration is variable up to 1500 Hz merely by setting the ramp length parameter contained in two RAM locations. For a normal scan (with a top speed of 1318 Hz) the frequencies and ramps are loaded from two subroutines (F and INIT) contained in EPROM into the RAM locations. The frequencies vary from 20.2 Hz to 1115 Hz and the corresponding ramp lengths vary from 1 to 42 pulses. The same ramp is used for acceleration and deceleration with two hundred steps being required in both cases. The number of pulses at a maximum frequency of 1422 Hz can be varied from 0 to over 4 million and hence ASD is suitable for presetting the scanner position.

Another minor subroutine R8D initializes parameters in RAM and then calls ASD resulting in a 25.7 degree rotation of the scanner. This is the rotation required between translations in a normal scan using the 8-detector holder.

Translation

Once the motor has been accelerated up to the desired speed a subroutine (TRANSL) is called. This subroutine makes use of a number of parameters previously loaded into RAM which determine the delay between motor pulses and the number of motor pulses per signal to the HP. A signal to the HP signifies that data is to be collected from the scalers.

Start

At the start of every translation of a scan an MPU based subroutine (START) is called. This routine (which uses the handshake arrangement between the MPU and the HP) sends a signal to the HP (signifying that a translation is desired) and in return waits for a signal from the HP (prior to allowing the MPU program to continue). An additional signal is sent to the HP prior to the start of the ramp.

Program Location

The MPU contains two EPROMs (1024 x 8 bit) located at \$6000-\$63FF and \$C000-\$C3FF. The EPROM located at \$C000 contains all subroutines such as ASD and TRANSL (see flowchart in appendix) while the main body of the control programs are located in the EPROM at $6000. Since the subroutines are rarely altered this means that most changes to the system control programs can be made by altering just one of the EPROMs. The RAM contained in the MPU is static RAM (128 x 8 bit) and is located at addresses $0000 through $03FF. The reason that many of the control parameters are loaded by the subroutines from RAM is that this theoretically makes it possible to alter the control parameters from the keyboard. At present however, most of the RAM parameters are loaded into RAM from the main control programs (which are stored in EPROM) at the same time as the main control program is run. This is due to the fact that the control parameters are at present seldomly changed.

Reproducibility of Measurements

Bone mineral changes are significant along the longitudinal aspect of the extremities. Consequently, for a valid quantification of changes in time in a given individual, it is important to be able to reproduce the site of the scan (50). To achieve this required positioning accuracy a MPU controlled scout scan system was developed which locates the desired measurement site.

The scout scan involves measurement of 26 single profiles at different points axially about the measurement site. Reconstruction of these profiles (see Figure 24) yields an X-ray like image which depicts the relative sizes

FIGURE 24. SCOUT SCAN

and orientations of the bone(s) (femur vs. radius, ulna). This information when compared to the same information obtained from previous scans of the same patient permits accurate location of the desired measurement site.

The anatomical criteria used to define the scan site are the distances from the distal tips of the radius and ulna axially along the bones. At present, sites two millimeters to either side of the measurement site are also being evaluated. These three sets of data are then interpolated in order to provide a reconstructed area identical to those previously measured.

Single Stepping

In order to provide the desired accuracy for initial positioning of the scanner it was necessary to provide some means for single stepping both the rotational and translational drive motors. This was accomplished by use of a 74123 retriggerable monostable multivibrator (51) as shown in Figure 25.

HP/Microprocessor Interfacing

The microprocessor unit both sends and receives positive signals of + 5 volts while positive signals sent from the HP are +12 volts. In order to sink the required current (12 mA) to provide a low signal to the HP the output from the PIA is tied into two inverters before being connected to the HP (Figure 26). Positive signals from the HP are converted to +5 volts by use of a resistor-zener diode divider circuit (Figure 27).

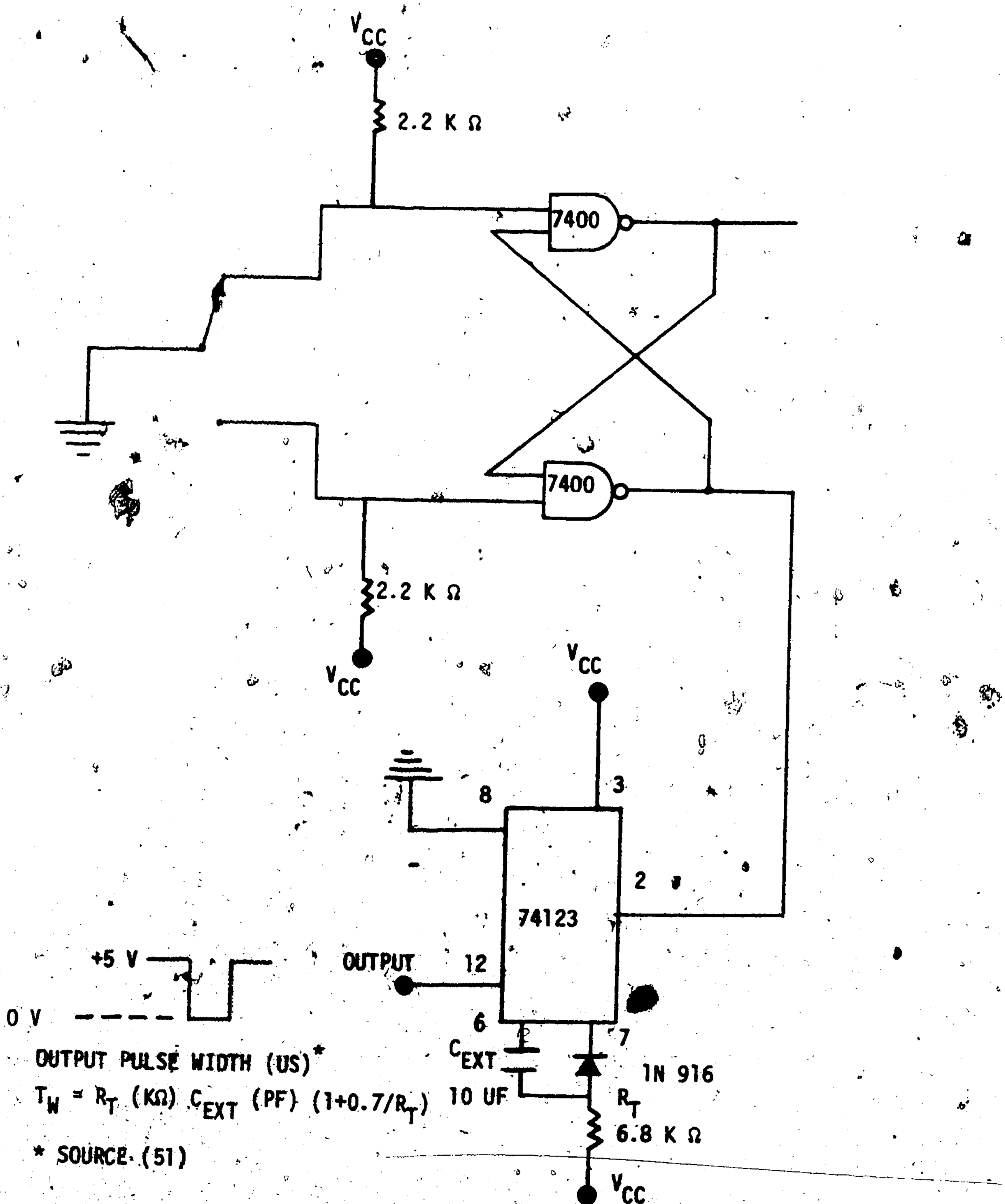


FIGURE 25. SINGLE STEPPING CIRCUITRY

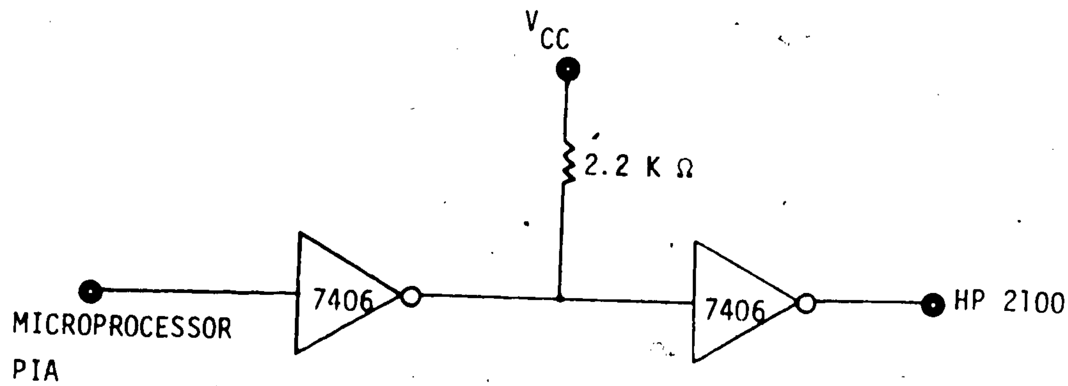


FIGURE 26. CURRENT SINK

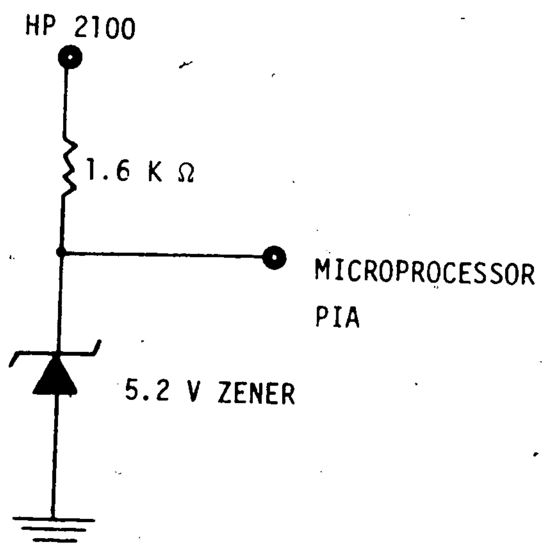


FIGURE 27. VOLTAGE DIVIDER

D. Multidetector System

Upon completion and subsequent testing of the generation one system the modifications necessary to change the system to a generation two configuration were undertaken. This chapter will describe the problems encountered in making these modifications.

The principle of the multidetector system is to collimate a single radioactive source so that an array of photons is produced which subtend a solid angle emanating from the source. The detectors are also collimated so that scattering of rays between detectors is reduced. Linear scans are made in exactly the same fashion as for the single-detector systems however the angle of each rotation is n times greater where n is the number of detectors (this assumes the detectors are spaced at the desired angular increments). Thus the number of linear translations and the number of rotations can be greatly reduced decreasing the time required for a scan. This reduction in scan time is significant in that errors caused by patient movement can be reduced. In this particular scanner moving from a generation one system to an eight-detector generation two system reduced the normal scan time from approximately 5 minutes to approximately 1.5 minutes. Although the order of data collection in a multi-detector system is different from that of a single detector system the reconstruction algorithm used is exactly the same.

Physical Limitations

In order for radiological attenuation of the object to be determined the detectors must be allowed to measure what is known as an "open" or unattenuated count rate during a scan. This means that in the multi-detector system, for each translation, all source-detector alignments must be totally clear of the object at some point in the translation. Given a finite translational pathlength this means that either one or both of the object size and the angle of the detector arc must be restricted. In this arrangement it is always possible to scan larger objects by not using the outer detectors in an array and effectively reducing the angle of the detector array. This would mean that the rotational angle between translations would have to be reduced.

Another limitation is the size of the detectors. The fact that these detectors can only be packed together within limits means that in order to have a desired angle between detectors the source-detector distance cannot be less than a certain minimum. This distance is of utmost importance since the photon flux impinging on the detectors follows the inverse square law meaning that the further the source-detector distance the worse the available statistics given the same radioactive source and counting interval. Other considerations in regards to the detector geometry include scan time, count rates and available channels for signal processing. The final result, an 8 detector unit similar to the one depicted in Figure 28, is a compromise

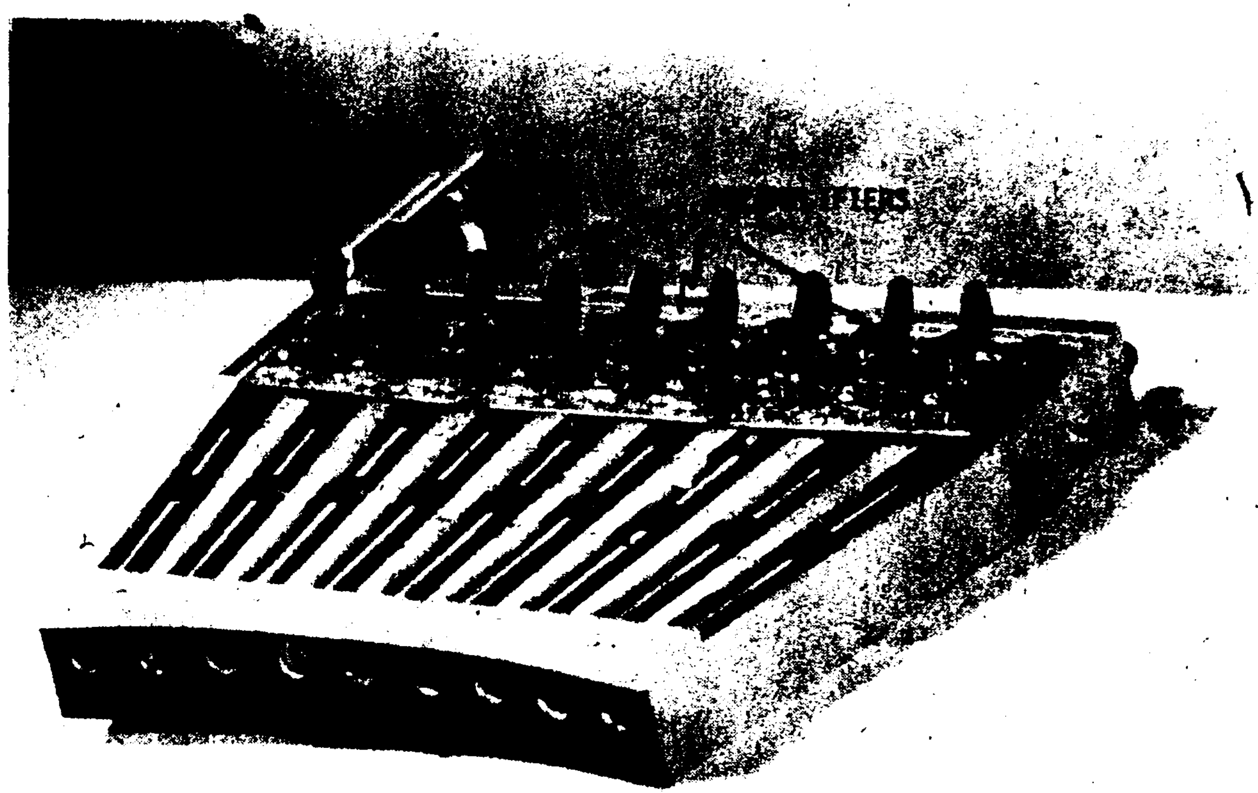


FIGURE 28. DETECTOR HOLDER

between the aforementioned factors.

Intervals and Subintervals

In the case of the multidetector system the translation of the scanner is perpendicular in direction to at most one of the source-detector orientations. Thus, while during a translation one detector may move a distance of "A" perpendicular to its orientation, the other detectors will move distances somewhat different than "A" perpendicular to their own orientations. The reconstruction algorithm requires that all the detector spacings have to be the same distance apart and therefore some means of correction has to be applied. This is done by measuring subintervals; each interval or ray-integral measurement consisting of a number of subintervals. Thus while the detector that is aligned perpendicular to the direction of motion will have intervals consisting of a specified number of subintervals the other detectors' intervals will be made up of more subintervals. The number of subintervals per interval varies from detector to detector according to their respective angular deviation from the line perpendicular to the direction of translation. Interpolation between subintervals is typically required in order to accurately define an interval. In this particular system 8 subintervals are measured for every interval of the detector aligned perpendicularly to the direction of motion. Thus while in a normal scan some 1024 subintervals are measured per detector per translation they are later compressed on the HP to 128 intervals prior to reconstruction.

Centering

The reconstruction matrix requires that each profile consist of equal amounts of data collected about the center of the reconstruction matrix. This is due to the fact that the convolution algorithm weights all ray measurements according to their distance from the point of reconstruction. In the case of the multidetector scanner each detector has a different starting point so it is impossible for all of them to collect equal amounts of data about the center of the scanner. In order to compensate for this, offsets must be calculated and applied to the collected data from each detector.

This problem is most easily seen by examining the case of a single detector scanner. Assume that the scanner moves a distance D1 from the start of a scan to an object O before completing the scan (Figure 29). Then, if the scanner is rotated through 180 degrees and started once again the distance from the start of the scan to the object O is now D2. In order that equal amounts of data be collected about both sides of the object the data will have to be offset by an amount equal to (D1-D2)/2. Thus if equal amounts of data have been collected about both sides of the object D1 is equal to D2 and no offsets are required. These calculations are aided by a special microcomputer program (Cent) which is a high resolution measurement. An object of high density and small physical size (i.e. a wire) is placed approximately in

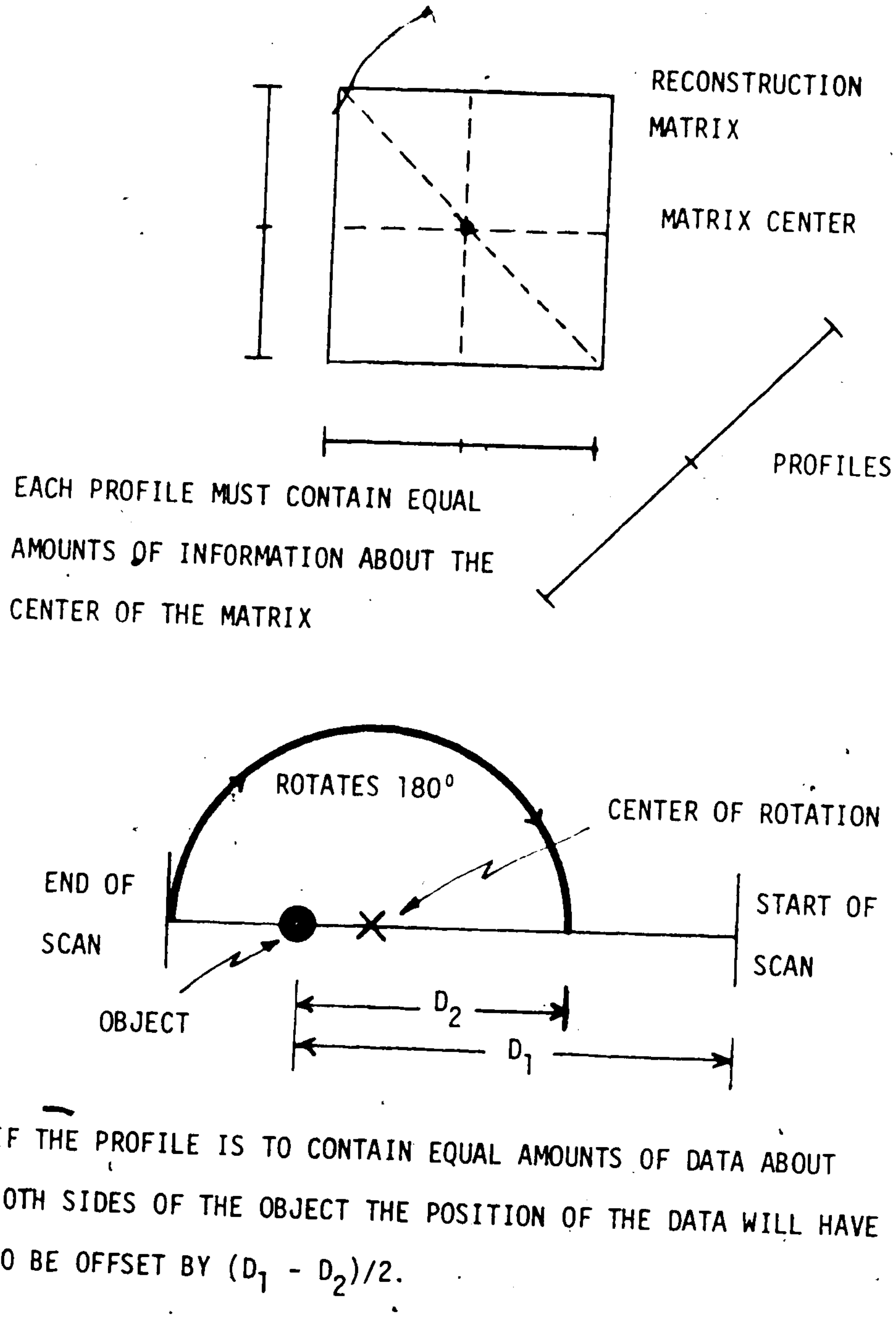


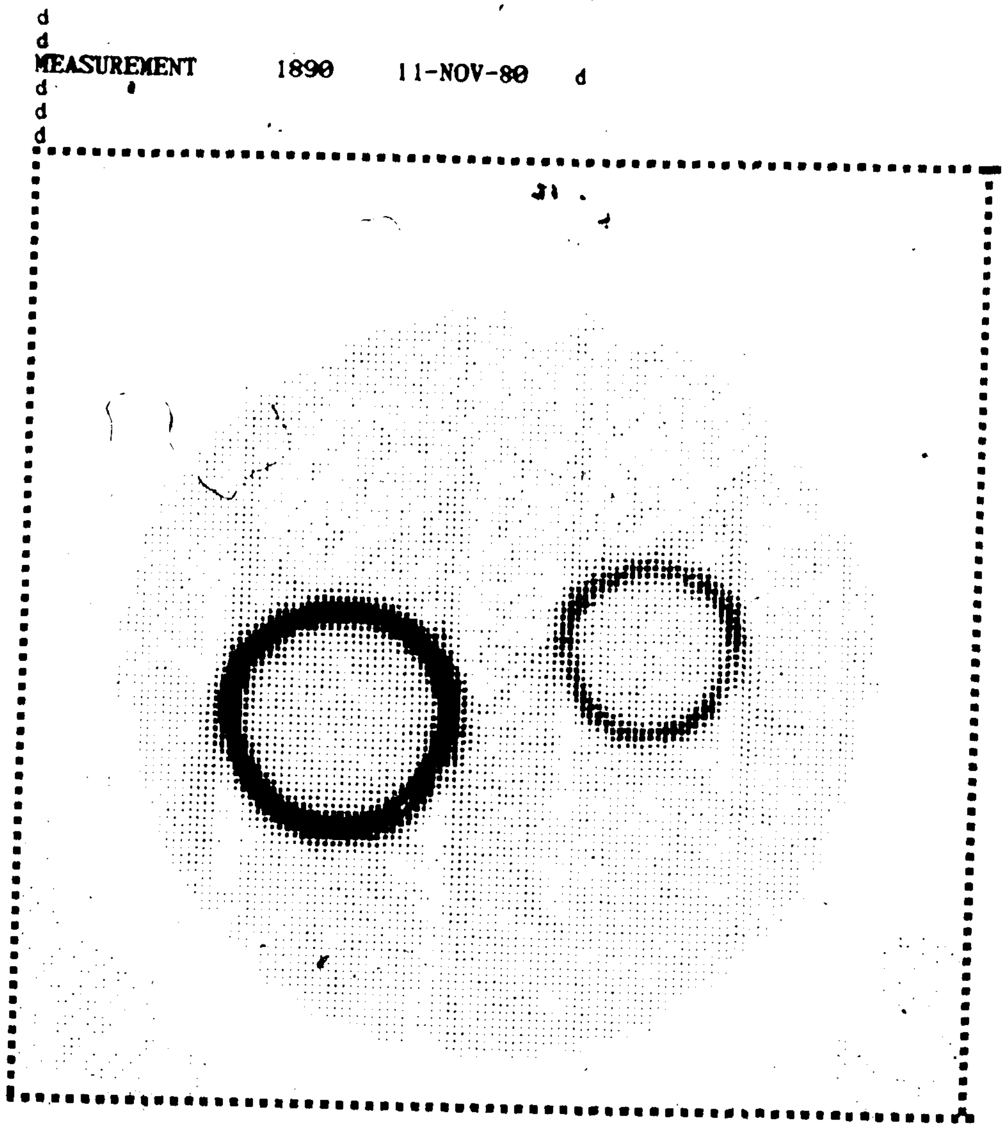
FIGURE 29. CENTERING CONSIDERATIONS

the middle of the scan path. The scanner is then translated at low speed and data is collected every two motor steps. The scanner is then rotated through 180 degrees and the scan is repeated. Looking at Figure 29 we see that information so obtained will allow one to calculate the offsets for each detector. This procedure must be repeated every time the mechanics of the scanner (such as the source) are altered.

With an aluminum-plexiglass model, centering errors of 100 motor steps or 0.625 mm, per detector are noticeable (Figure 30). In the case of a circular object the centering artifact appears on the image as two circles of slightly different radii being joined.

E. Data Processing

Data for one translation is stored in the HP 2100 and during a rotation is transferred to magnetic tape. This is then processed and analyzed on the University Computer (Amdahl 470/ V7) and hard copies of the image are obtained on the associated electrostatic plotter printer. The reconstruction program uses the convolution technique with the filter function proposed by Shepp and Logan to produce a 256 x 256 reconstruction matrix. Prior to reconstruction both dead time and and a non-material selective beam hardening correction are made on the Amdahl as is an interpolation between ray measurements and angles. The interpolation takes the 128 data points per projection and interpolates between them in order to produce 256 data



PLEXIGLASS PHANTOM WITH TWO CIRCULAR ALUMINUM TUBES

FIGURE 30. CENTERING ARTIFACTS

points per projection and also interpolates between projections to produce 112 instead of the normal 56 projections. Although these interpolations do increase computation time and have a smoothing effect on the image the increased matrix size is needed in order that the analysis program has enough pixels with which to work. Interpolation between the angles is necessary to remove angular artifacts from the image. It has been shown (7) that if there are N ray-integral measurements made per projection, then for the data to be complete there should be M projections, where $M=(\pi/4)N$. Thus if there are 128 ray-integral measurements made there should be $128(\pi/4)$ or approximately 100 projections made in order for the data to be complete. Hence if the number of angles were not interpolated to produce 112 projections instead of 56 projections, angular artifacts would be present in the reconstructed image. The interpolation between measurements is to produce 256 ray-integral measurements so that reconstruction on a 256 x 256 matrix can be performed easily. It should be noted that it is not necessary to reconstruct on a matrix the same size as there are number of ray-integrals per profile; however, reconstruction is facilitated if this is the case.

In order to use a Grinnell CRT display, which is tied into the HP system it is necessary to either reconstruct the matrix on the HP 2100, which takes about 30 minutes (for a 128 x 128 matrix), or else to load the Amdahl reconstructed matrix onto magnetic tape and to subsequently load the matrix into the HP.

Open Count

Although the open count rate is measured for all detectors during each translation the statistics associated with such counts are of limited accuracy due to a short counting period. For this reason an open count rate is made at the start of the day. This count is made over a period of several minutes (using the Scout Scan program) so that the coefficient of variation associated with the measurement is low. This open count (Io) is then used throughout the day to calculate the ray-integral measurements (ln(Io/I)).

Pixel Size and Grey Levels

The pixel size of the system is determined by dividing the scan length by the number of pixels in one side of the reconstruction matrix. The maximum size object that can be scanned is 76.8 mm and the size of the reconstruction matrix is 256 x 256. Hence the pixel size of the system is 76.8 mm/256 or 0.3 mm. Each pixel level in the reconstructed matrix is defined by a two's complement 16 bit number and hence there are 32,768 possible positive pixel levels. Although the system pixel size is 0.3 mm corresponding to a 256 x 256 matrix the hard copy imaging system is not able to resolve points of that size. As a result a 128 x 128 matrix with 32 grey levels is used with a corresponding pixel size of 0.6 mm. Since there are four times as many pixels in the actual reconstruction matrix as in the image matrix, four of the actual pixels are averaged together to produce one of the image pixels. The Grinnell display system is able to reproduce matrices of size 512 x 512 with 256 grey levels and hence the system limiting pixel size for this display mode (CRT) is 0.3 mm.

Analysis Program

While the image depicting the bones may be aesthetically pleasing it is really the parameters contained in subsequent analysis programs that are of clinical value. The image's main function is to ensure that the scan was done properly and that no artifacts due to such factors as patient movement have been introduced. Once an image of the bone has been reconstructed an analysis program is used to identify such parameters as area, total mineral content and average density. These paramenters are defined for total bone, trabecular bone and the cortical or compact bone (Table 1).

The analysis program first defines where the cortical bone begins and strips away the surrounding fleshy material (Figure 31). This is done by defining a cortical bone density level (8000 is normally used) and defining all levels of material outside of those pixels with levels greater than or equal to 8000 as being zero. The radius and ulna are then separated and each bone is analyzed separately. Each bone is then separated into a number of shells by essentially moving the cortical bone contour in, one pixel at a time. Trabecular bone was initially defined

MEASUREMENT: 596 13-MAR-80

RADIUS EVALUATION LEVEL 8000

	SHELL					REST BONE					REST AREA	
	DENSITY [G/CCM]	SM [G/CCM]	SM [%]	S [G/CCM]	S [%]	DENSITY [G/CCM]	SM [G/CCM]	SM [%]	S [G/CCM]	S [%]	A [PIX]	[%]
0	0.0	0.0	0.0	0.0	0.0	0.8266	0.0093	1.12	0.5374	65.01	3374	100.00
1	0.9161	0.0078	0.85	0.1115	12.18	0.8208	0.0098	1.20	0.5534	67.42	3168	93.89
2	1.2343	0.0148	1.20	0.2095	16.97	0.7931	0.0102	1.29	0.5582	70.38	2969	88.00
3	1.4802	0.0206	1.39	0.2867	19.37	0.7453	0.0103	1.38	0.5408	72.56	2776	82.28
4	1.5982	0.0256	1.60	0.3503	21.92	0.6837	0.0098	1.43	0.4984	72.90	2589	76.73
5	1.5921	0.0281	1.76	0.3777	23.72	0.6155	0.0089	1.44	0.4356	70.78	2408	71.37
6	1.4603	0.0267	1.83	0.3521	24.11	0.5497	0.0078	1.41	0.3674	66.84	2234	66.21
7	1.2573	0.0253	2.01	0.3300	26.24	0.4914	0.0067	1.36	0.3042	61.90	2064	61.17
8	1.0512	0.0247	2.35	0.3176	30.21	0.4427	0.0057	1.29	0.2495	56.35	1899	56.28
9	0.8709	0.0207	2.37	0.2623	30.12	0.4031	0.0050	1.24	0.2076	51.51	1738	51.51
10	0.7063	0.0183	2.59	0.2282	32.32	0.3732	0.0045	1.21	0.1797	48.14	1582	46.89
11	0.5664	0.0147	2.60	0.1810	31.96	0.3528	0.0044	1.25	0.1670	47.34	1431	42.41
12	0.4715	0.0135	2.87	0.1624	34.44	0.3395	0.0045	1.33	0.1622	47.79	1287	38.14
13	0.4059	0.0152	3.75	0.1774	43.71	0.3316	0.0047	1.41	0.1586	47.82	1151	34.11
14	0.3710	0.0163	4.38	0.1860	50.15	0.3266	0.0048	1.48	0.1541	47.19	1020	30.23
15	0.3742	0.0145	3.87	0.1625	43.42	0.3199	0.0051	1.59	0.1518	47.45	894	26.50
16	0.3983	0.0133	3.34	0.1462	36.71	0.3076	0.0054	1.74	0.1490	48.45	773	22.91
17	0.3875	0.0139	3.58	0.1490	38.44	0.2936	0.0056	1.92	0.1447	49.28	658	19.50
18	0.3479	0.0136	3.90	0.1409	40.49	0.2830	0.0061	2.16	0.1432	50.59	550	16.30
19	0.3233	0.0152	4.69	0.1530	47.34	0.2738	0.0066	2.40	0.1394	50.90	448	13.28
20	0.3108	0.0157	5.06	0.1509	48.54	0.2642	0.0071	2.70	0.1348	51.02	356	10.55
21	0.2817	0.0156	5.53	0.1436	50.97	0.2587	0.0080	3.09	0.1317	50.91	271	8.03
22	0.2499	0.0169	6.74	0.1460	58.39	0.2621	0.0090	3.44	0.1261	48.11	196	5.81
23	0.2554	0.0179	7.02	0.1468	57.48	0.2656	0.0101	3.79	0.1144	43.06	129	3.82
24	0.2970	0.0159	5.34	0.1156	38.90	0.2437	0.0125	5.13	0.1090	44.73	76	2.25
25	0.2514	0.0200	7.95	0.1199	47.70	0.2367	0.0157	6.62	0.0992	41.90	40	1.19
26	0.2250	0.0215	9.56	0.1032	45.86	0.2525	0.0228	9.05	0.0942	37.29	17	0.50

TOTAL BONE:

AREA: 3.0366 [CM**2]
TOT.MIN.CONTENT: 2.5102 [G/CM]
AV.DENSITY (BD): 0.8266 [G/CCM]

TRABECULAR BONE:

REL. AREA: 61.1737 [%]
TOT.MIN.CONTENT: 0.9128 [G/CM]
AV.DENSITY (TBD): 0.4914 [G/CCM]

COMPACT BONE:

ABS.THICKNESS: 0.2142 [CM]
REL.THICKNESS: 21.7863 [%]
AV.DENSITY (CBD): 1.3549 [G/CCM]

TABLE 1.

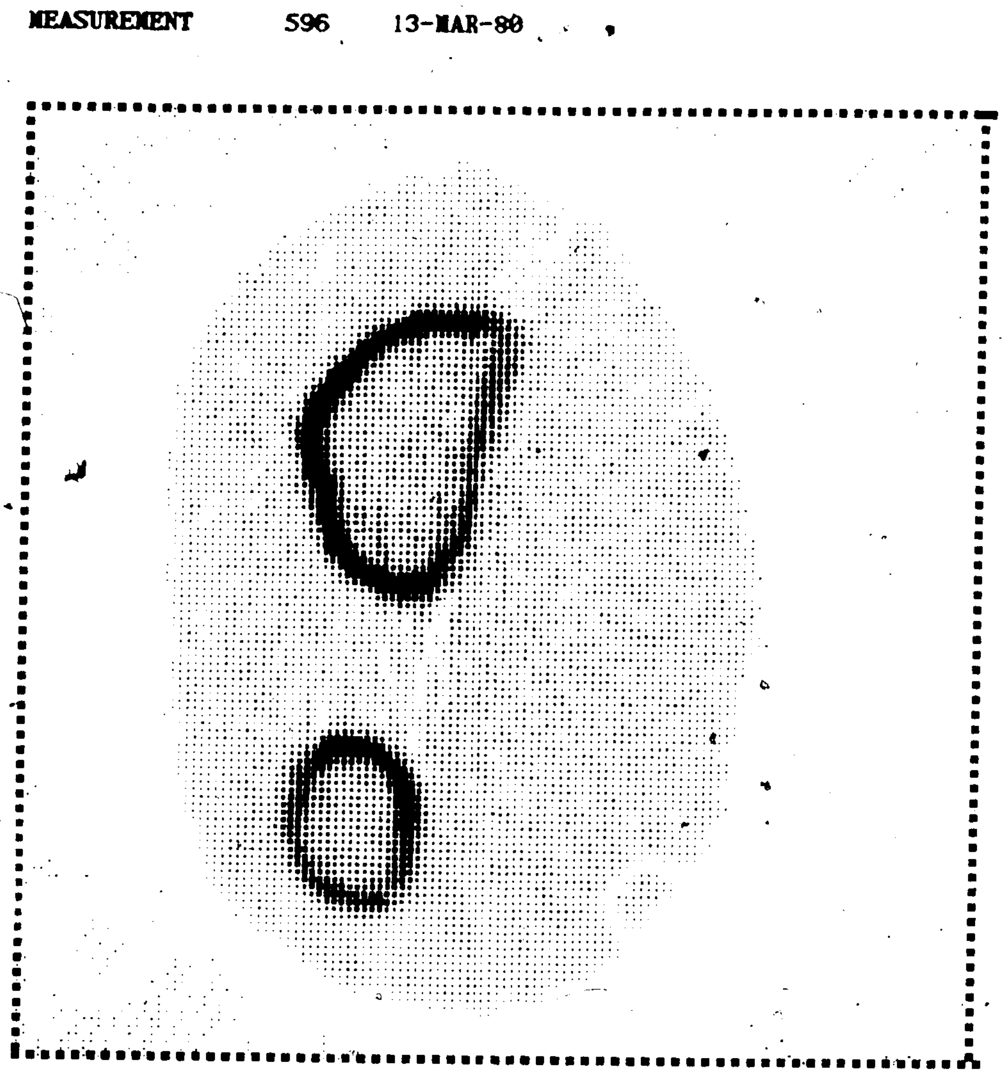


FIGURE 31. NORMAL RADIUS-ULNA

as that area of bone which comprised the inner 50 percent of the total bone area. It was felt that this arbitrary definition of trabecular bone may not be applicable to all patients. The criteria now used involves first calculating the average densities for all shells. Trabecular bone is then defined as the inner area of bone bordered by the maximum change in adjacent shell densities. The CT attenuation coefficients are converted to bone density (g/ccm) by dividing μ by the average mass attenuation coefficient for bone. Due to the fact that a non-material selective beam hardening correction is applied (as contrasted to a material selective correction (48)) these measures of density are only valid for longitudinal studies. Cross-sectional (i.e. population) studies require absolute values for bone density and this necessitates material selective corrections.

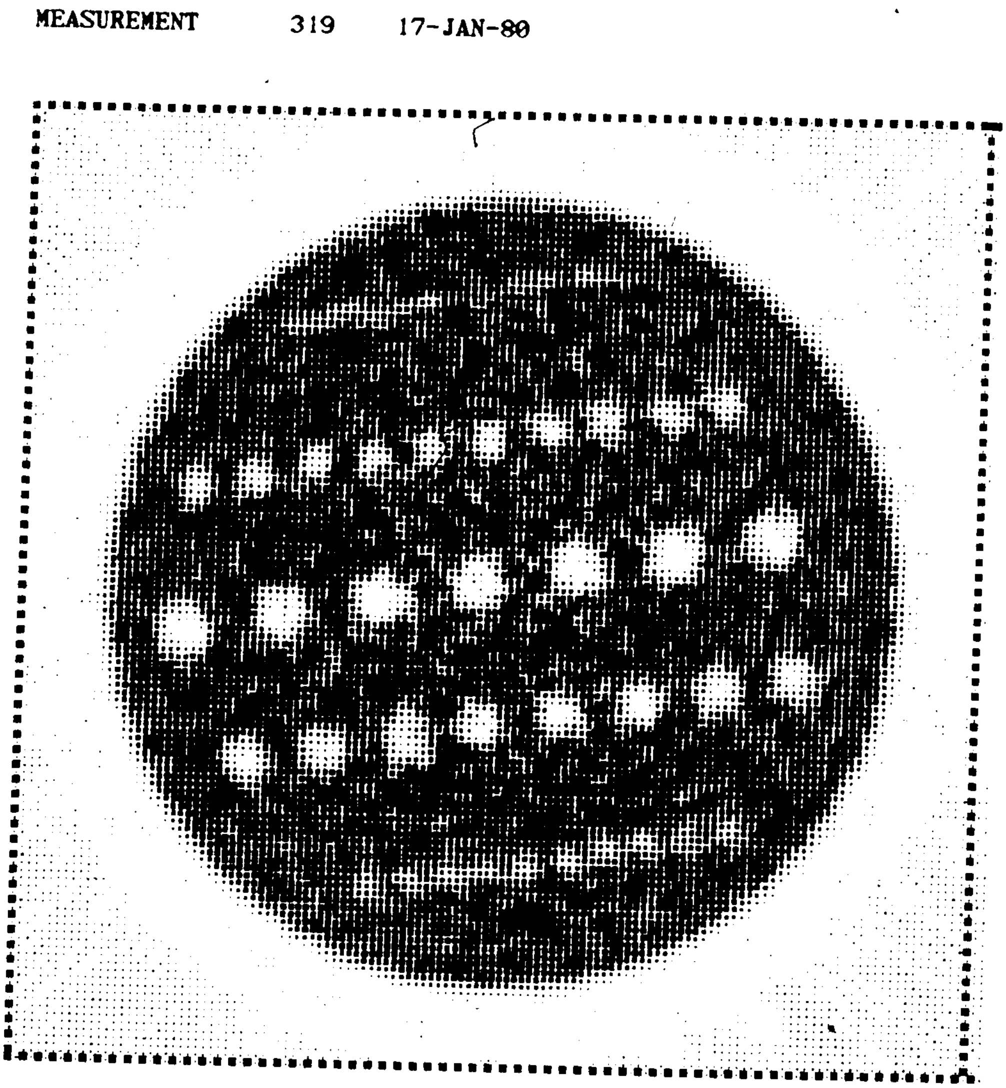
V. System Evaluation

The scanner was evaluated in terms of: (1) spatial resolution, (2) precision, (3) accuracy and (4) radiological dosage. The results of these evaluations are discussed in this section and a technique for measuring contrast resolution is suggested.

A. Spatial Resolution

Physical resolution of the system was evaluated using a 2 1/2 inch diameter circular plexiglass model with periodic structures of various size holes drilled into it. These holes are drilled so that the space between holes is equal to the hole diameter. The model was subsequently scanned at three different speeds and the holes were left open to the air. Motor pulses during the three speeds of translations were sent at 1318 Hz, 330 Hz and 82.4 Hz and data from the CAMAC scalers was collected every 12 motor pulses in all three instances. Since these speeds are ratios of 16:4:1 respectively the resultant subinterval counts obtained during the scan were in the ratios of 1:4:16 respectively. Since coefficient of variation (C.V.) is inversely proportional to the square root of the number of counts this means that the C.V.s associated with each of the ray-integral measurements were improved by a factor of two with each decreasing speed of scan. The open count rates of the source for detectors 1 through 8 were; on a subinterval basis: (1) 374.42, (2) 349.92, (3) 334.08, (4) 345.49, (5) 389.56, (6)369.65, (7) 405.93, and (8) 379.45. Thus the C.V. associated with an average 8 subinterval per interval open count measurement is $1/\sqrt{(2948.49)}$ = 1.84 %. At the medium speed (interval C.V. = 0.92 %) the spatial resolution has been improved whereas a further decrease in the interval C.V. to 0.46 % (low speed) had no apparent impact on the spatial resolution of the system.

It should be noted that one limit of the system in regards to spatial resolution is the spatial resolution of the imaging system. In this case although the reconstructed matrix is of size 256 x 256 with a 0.3 mm pixel size, the hard copy image matrix is only of size 128 x 128 with a corresponding pixel size of 0.6 mm. To further complicate matters, the image display has a magnification effect of 1.6 times that of the object. Using the Grinnell display system one can display a matrix size of 512 x 512 and hence it is possible to view the system limited resolution on the video screen and yet essentially the same results were obtained as with the hard copy display. At high speed (Figure 32) the system spatial resolution was limited to 1/8 of an inch while at medium speed (Figure 33) the spatial resolution is improved to one sixteenth of an inch (approximately 1.5 mm). However, a further decrease in speed to low speed (Figure 34) had little or no effect on the spatial resolution. This is evidence that beyond a certain level statistics are no longer the limiting factor in regards to spatial resolution.



HOLE DIAMETERS (INCHES) ARE (TOP TO BOTTOM) 1/32, 3/32, 5/32, 1/8 AND 1/16.

FIGURE 32. HIGH SPEED RESOLUTION

MEASUREMENT 318 17-JAN-80

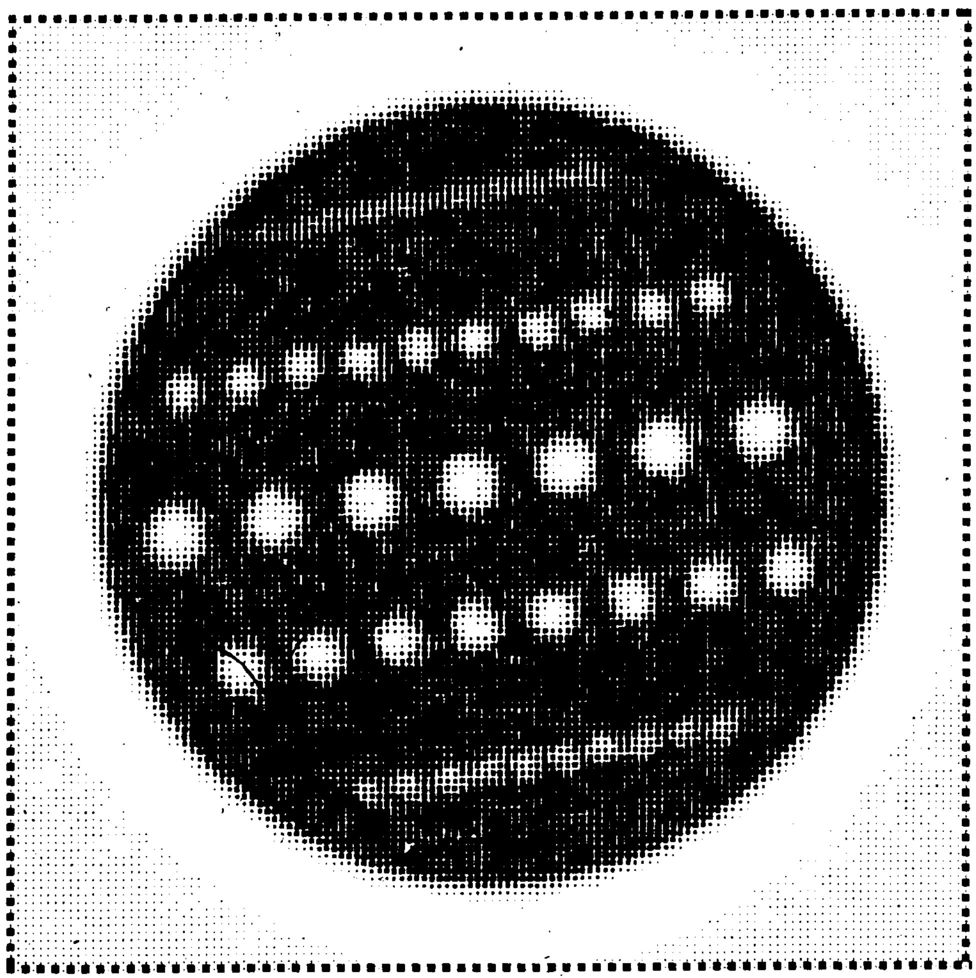


FIGURE 33. MEDIUM SPEED RESOLUTION

MEASUREMENT 317 17-JAN-80 d

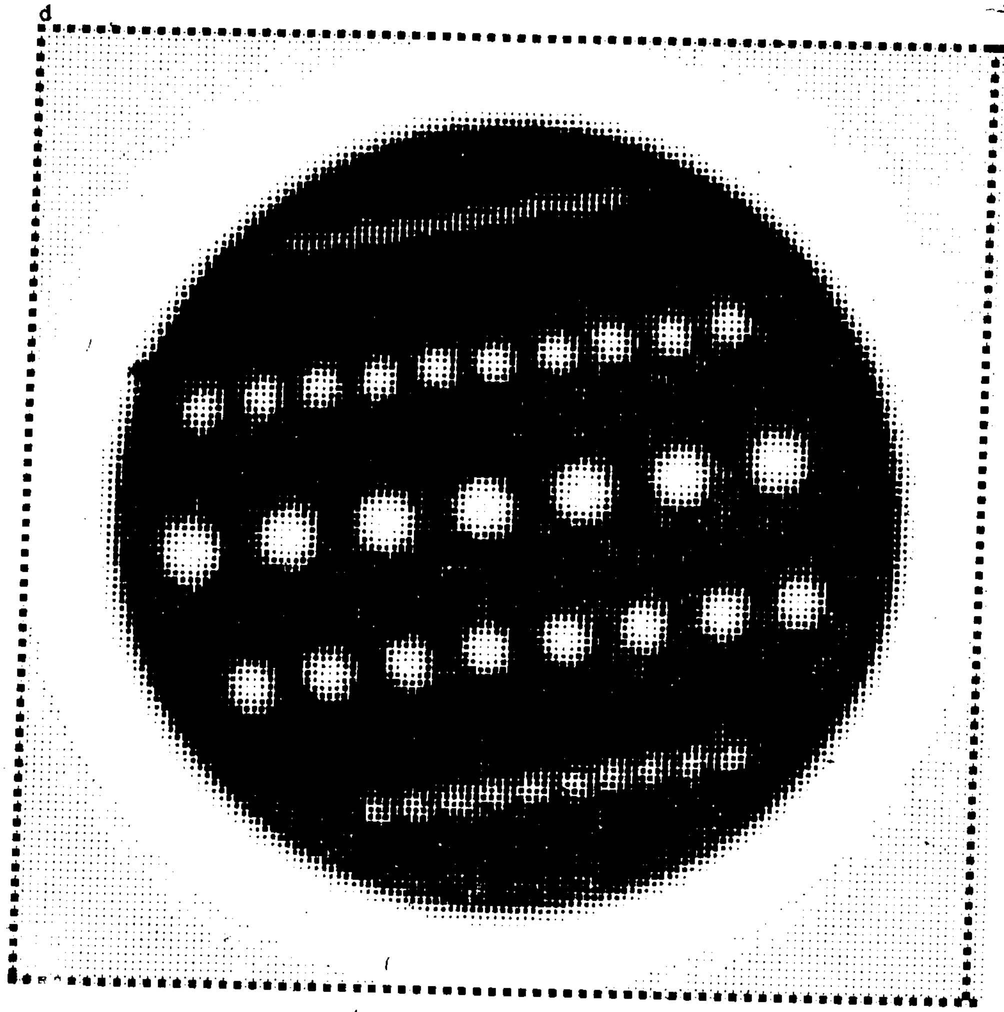


FIGURE 34. LOW SPEED RESOLUTION

B. Precision

The critical factor for the overall precision of the system is not so much the precision associated with the machine as with the ability to remeasure the patient at exactly the same site since both the amount of trabecular bone and its density vary significantly at sites axially along the arms or legs. This means that precision of the system is predominantly a function of the ability to determine the site of measurement. To reduce these errors in precision due to repositioning use of the scout scan was instituted.

A number of repeat scans were made on a 40 year old male normal (52). Each scan involved repositioning the arm using the scout scan and making three measurements on the arm. The first measurement was made at the site selected by the scout scan while the other two measurements were made at sites two millimeters of either side. These three measurements were then interpolated to produce the same area of bone for analysis purposes as was used in prior analyses. The data obtained from these measurements (Table 2) shows that a precision of better than ± 1 % is achievable for both cortical and trabecular bone density.

C. Accuracy

Accuracy of the scanner is a difficult quantity to define since it relates to a "correct" value being obtained. For measurements of bone this means that the bone would have

PRECISION EVALUATION

Measurement Number	Scan * Number	Bone Area (Pixels)	Trabecular Bone Density	Cortical Bone Density
1	1	3362	0.503	1.301
	-	-	-	-
2	1	3577	0.491	1.250
	2	3293	0.519	1.333
3	1	3552	0.501	1.250
	2	3248	0.522	1.328
4	1	3389	0.509	1.275
	2	3155	0.535	1.370
5	1	3586	0.497	1.237
	2	3308	0.510	1.330

With the above scans interpolated so as to produce a bone area of 3362 the results are as follows

1	3362	0.503	1.301
2	3362	0.512	1.313
3	3362	0.514	1.299
4	3362	0.512	1.286
5	3362	0.507	1.292

These interpolated results, when considering a normal distribution, result in standard deviations of 0.884 % for trabecular bone density and 0.847 % for cortical bone density.

*Note: Although three scans were made during each measurement only those two scans that were used in the interpolation are included.

TABLE 2.

to be removed and "ashed" to determine the true mineral content. Obviously, this is only possible to perform with cadavers and consequently the sources of data are limited.

To circumvent these problems a solution of di-Potassium Hydrogen Phosphate (K_2HPO_4) was used (53). The solution was mixed to a known concentration and placed inside a plexiglass cylinder (Figure 35). CT measurements were made on a number of different concentrations and the resultant linear attenuation coefficients of the image were compared to the known values of the dilute solution. Results of these tests are shown in Table 3 and Figure 36. These results show that although the absolute values may not be entirely accurate, the percentage change in value is quite accurate. Since the mineral content present in bone does vary tremendously from person to person it is measures of percentage changes that are of clinical importance rather than absolute values of mineral density. One reason that the absolute values are of limited accuracy is that the linear attenuation coefficients for the solution are calculated on the basis of a single energy which does not coincide exactly with the energy spectrum of Iodine 125.

D. Radiation Dose

Calculations of a theoretical radiological dose (as performed on the following pages) show that the total radiation dose received during the course of a scout scan and three adjacent scans is 0.9 rads.

MEASUREMENT 861 3-JUN-80

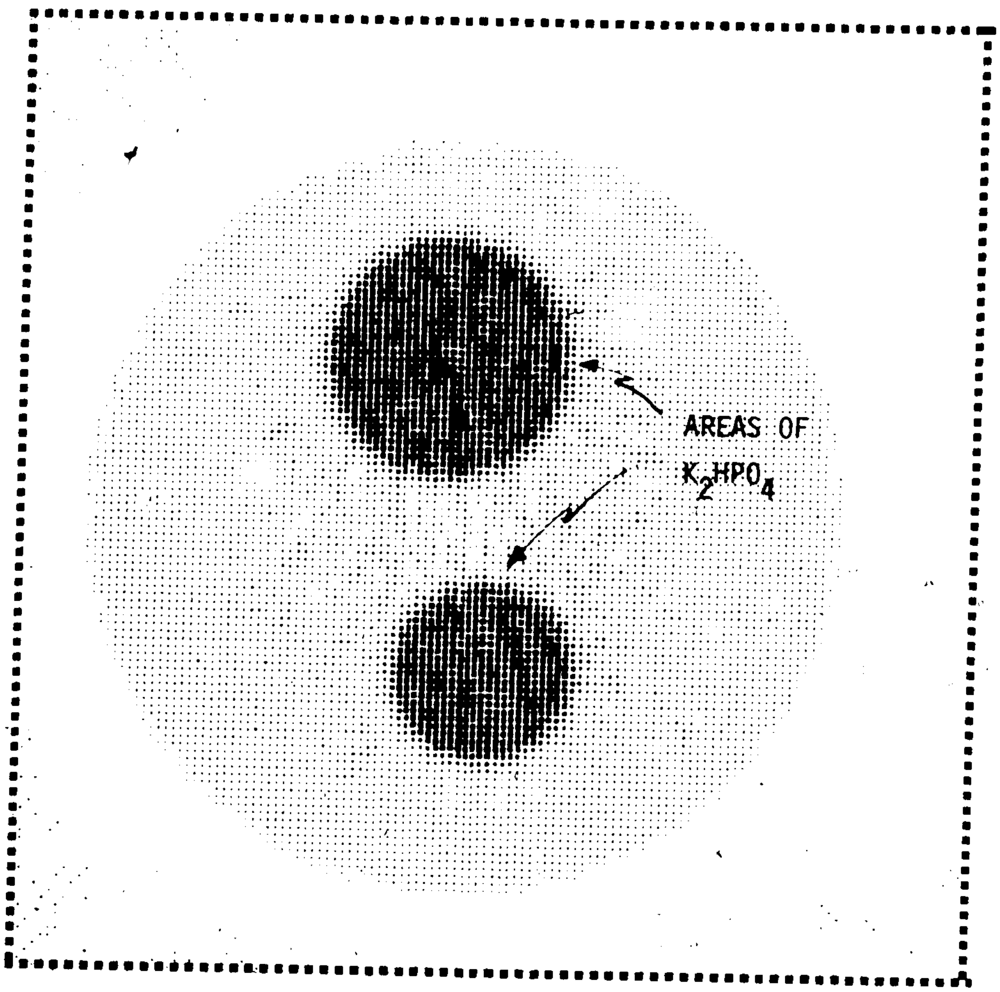


FIGURE 35. TEST CYLINDER

ACCURACY EVALUATION

Measured Linear Attenuation Coefficient (per cm.)	Percent of Maximum Measured Linear Attenuation Coefficient	Calculated Linear Atten. Coefficient (per cm.)	Percent of Maximum Calculated Linear Attenaution Coefficient
1.882	100	1.709	100
1.852	98.4	1.675	98.0
1.814	96.4	1.641	96.0
1.776	94.4	1.606	94.0
1.765	93.8	1.598	93.5
1.747	92.8	1.589	93.0
1.739	92.4	1.581	92.5
1.735	92.2	1.572	92.0
1.690	89.8	1.538	90.0
1.654	87.9	1.504	88.0
1.615	85.8	1.470	86.0

A linear regression of the percent calculated linear attenuation coefficient was run against the percent measured linear attenuation cefficient (using the SPSS package) and found to have a correlation of $r=0.9987$ and a significance of $p < 0.0001$

TABLE 3.

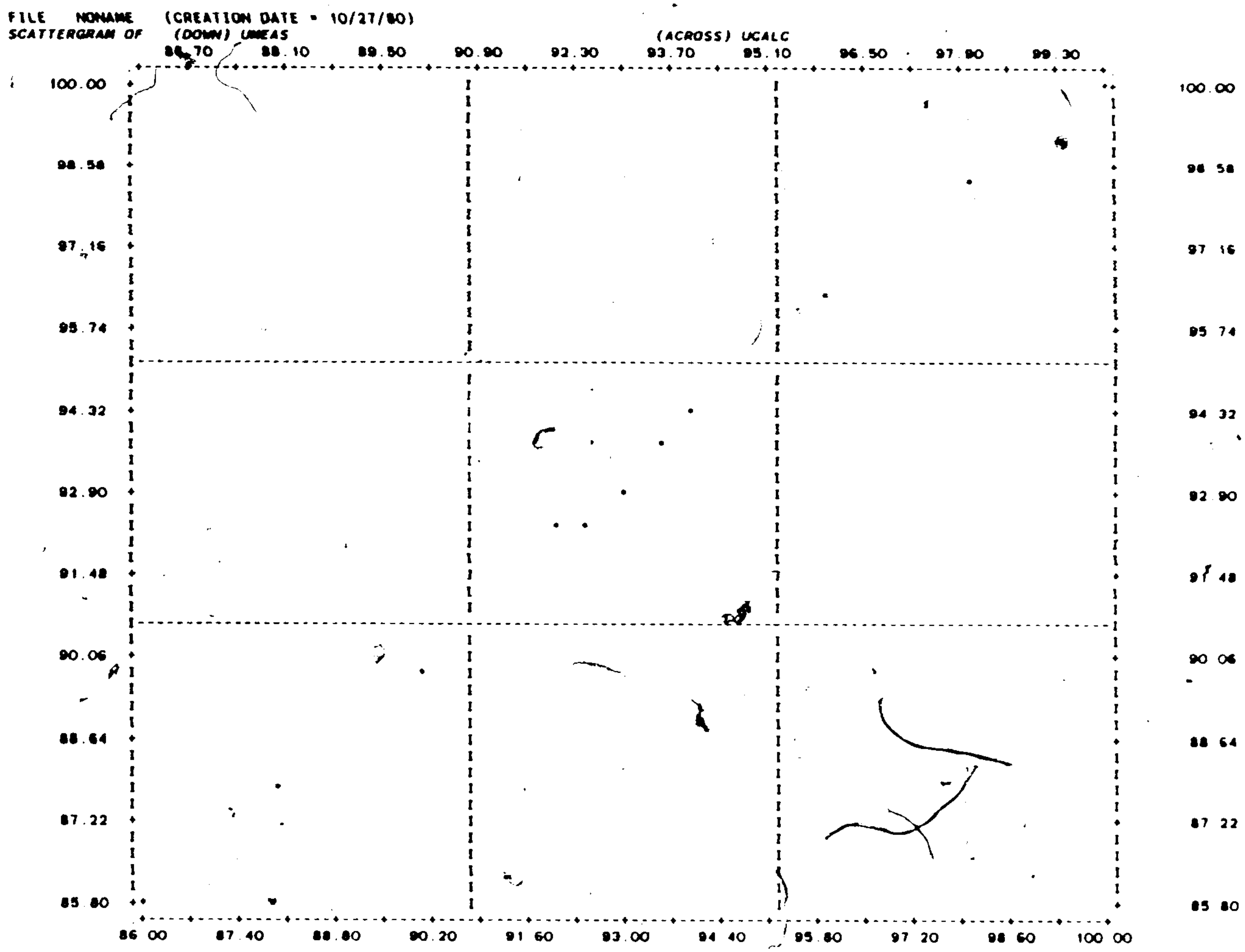
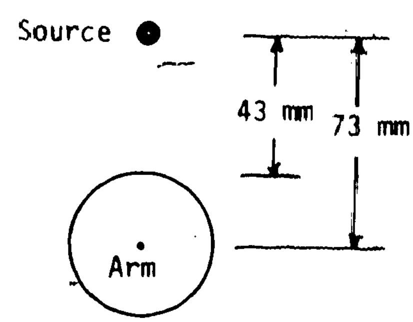


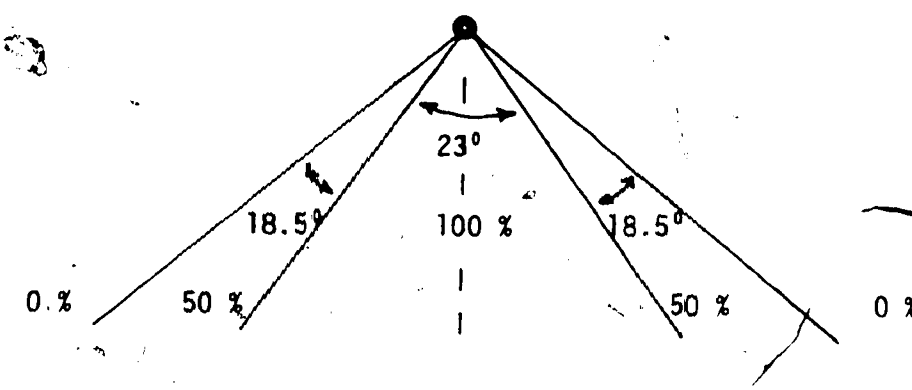
FIGURE 36. SYSTEM ACCURACY

Radiological Dose

Assume that an arm of diameter 60 mm is placed in the center of the scan area. This results in a distance of 43 mm from the source to the nearest point on the arm.



At a distance of 43 mm the source-collimator configuration was found to produce the following approximate intensity distribution.



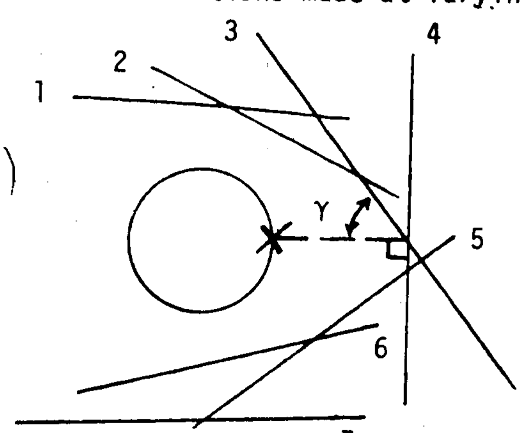
The scanner moves a distance of 106.2 mm in 12.89 sec resulting in an average speed of 8.24 mm / sec. The area irradiated at 100 % intensity consists of a length L, where $L = 2 \tan (23^\circ/2)$ (43 mm) = 17.5 mm. The area irradiated at an average intensity of 1/2(100% + 50%) = 75% consists of 2 lengths, each one of length L1, where

L1 = 1/2 (2 tan ($60^0/2$) 43 mm - 17.5 mm) = 16.08 mm.

Thus the total length irradiated at effectively 100% is

17.5 (100%) + 2(16.08)(75%) = 41.61 mm which at a speed of 8.24 mm / sec means an effective (100%) irradiation time of 41.61 mm / 8.24 mm/sec = 5.05 seconds.

This is the irradiation time for one translation. However, each scan consists of 7 translations made at varying angles.



The rotation between angles is 25.71° and the approximate "hotspot" (area subjected to maximum radiation) is that point on the arm perpendicular to translation number 4 (denoted by point x).

Thus at $\gamma = 90^0$ the irradiation is 100% while at $\gamma = 0^0$ the irradiation is 0 %. Hence irradiation is proportional to sin γ.

Translation 1	$\gamma = 90^0 - 3(25.71^0) = 12.87^0$	$\sin\gamma = 0.22$
Translation 2	$\gamma = 90^0 - 2(25.71^0) = 38.58^0$	$\sin\gamma = 0.62$
Translation 3	$\gamma = 90^0 - 1(25.71^0) = 64.29^0$	$\sin\gamma = 0.90$
Translation 4	$\gamma = 90^0$	$\sin\gamma = 1.00$
Translation 5	$\gamma = 64.29^0$	$\sin\gamma = 0.90$
Translation 6	$\gamma = 38.58^0$	$\sin\gamma = 0.62$
Translation 7	$\gamma = 12.87^0$	$\sin\gamma = 0.22$
		4.48

Therefore, for one scan the total irradiation time (100% intensity) at the hotspot is 4.48 (5.05 sec) = 22.62 seconds.

A new source of 1.5 Ci I^{125} was measured using an ionization chamber at a distance of 0.5 meters (so that the entire chamber was irradiated) and found to have a dose rate of 11 mr/min. Using the inverse square law the dose rate at 43 mm would be 11 mr/min $(500/43)^2$ = 1487 mr/min = 24.79 mr/sec. This means that the total dose per scan at the hotspot is 24.79 mr/sec (22.62 sec) = 560 mr. Since the speed of the scout scan is the same as for a normal scan the effective (100%) irradiation time is 5.05 sec leading to a dose of 5.05 sec (24.79 mr/sec) = 125 mr. Assuming adjacent measurements produce a 50 % increase in dose the total exposure dose for 1 scout scan and 3 adjacent scans is

1.5(560 + 125) = 1027.5 mr ≈ 1 r (roentgen).

At 30 KeV this is approximately equal to an absorbed dose of 0.9 rad.

Although these theoretical calculations do provide a rough estimate of dose there are a number of associated problems. One major problem is the assumption that absorption of radiation is only significant at the surface. Although this may be true for denser objects with high linear attenuation coefficients it is not true for skin. The linear attenuation coefficent of skin is approximately 0.45/cm (for 30 KeV) which means that some 1.5 cm of skin must be traversed before the intensity of the radiation is down by a factor of 2. Since there is a limited amount of skin at the forearm, radiation passing through skin only will have a significant effect on the dose rate whether or not it directly impinges upon the area of consideration. Although most rays do pass through some bone and thereafter do not contribute appreciably to skin dose it should be realized that there is an element of variation from person to person.

Another area of concern is the effects of scattered radiation from the bones and their effect on the dose at a particular point. It is relatively complicated to account for scatter on a theoretical basis and as such the use of Thermo-Luminescent Detectors (TLD's) has been suggested.

Thermo-Luminescent Detectors

TLD's are capsules of powder or crystals that react chemically when exposed to radiation. After exposure to radiation the powder is removed from the capsules and heated up. When heated the powder produces photon flashes the number and intensity of which are proportional to the radiation exposure.

It is presently being proposed that a number of holes be drilled in a plexiglass phantom and filled with TLD powder. Subsequent to making a number of measurements on the phantom the powder is to be removed and tested to determine the radiation exposure. The plexiglass phantom will contain aluminum tubes to simulate the radius and ulna and will be made in the approximate size and shape of a human forearm. Measurements of dose performed in this manner should produce a profile of the radiation dose throughout a human forearm.

E. Contrast Resolution

Contrast resolution could be measured by examining the limiting spatial resolution with differing contrast media in the holes. One possible contrast media is di-Potassium Hydrogen Phosphate (K_2HPO_4) mixed in a water base. If solutions of varying concentration were placed in the 1/16 th of an inch diameter holes then the system limited contrast resolution would be defined by comparing the smallest linear attenuation coefficient of the imageable solutions with that of plexiglass. The major difficulties in these measurements will be placing the proper concentration solutions in the holes and ensuring that no air bubbles are present. Measurements of contrast resolution have not been made due to these difficulties and the limited usefulness of such measurements.

VI. System Limitations and Future Plans

Four of the major limitations at present are: (1) statistical limitations due to the use of a radioisotope photon source, (2) a lengthy scanning time due to use of a second (versus a third or fourth) generation CT scanner, (3) a lack of flexibility in regards to scanner control, and (4) an inability to view the image immediately after scanning due to limited processing facilities. This chapter will explore these limitations and will suggest ways to reduce or eliminate them.

A. X-Ray Tube

Subject to appropriate funding present plans include the purchase of a mini-X-ray tube. As a result of physical considerations photon flux is more limited in the case of a radioisotope source than for an X-ray tube. Although use of an X-ray tube may result in higher radiation dose rates it will also facilitate a reduced scan time and hence the net radiation exposure may not be radically altered. With the aid of filters the X-ray tube will also allow a greater selection of photon energies to be used. This may allow the scanner to perform elemental identifications through the use of differential energy scans and to be used for scans of areas of high radiological attenuation such as the knee. The use of filters may also reduce the problems of beam hardening caused by the multi-energy X-ray radiation. Although beam hardening will be a problem, so long as the "raw" data is available, it is possible to correct for.

B. System Three

Present plans include the construction of a generation three system. This improvement will result in a substantial decrease in scan time hence reducing movement artifacts, improving patient comfort and reducing the dose of radiation.

C. Control System

The major constraint at present is that the control programs are "locked" into EPROMs. In order to alter any of the scan parameters, such as speed, scan length, distance between projections and/or the angle between profiles, a new program must be written into the EPROMs. To circumvent this problem it is proposed that the EPROMs be programmed with a number of basic subroutines and that all control parameters be loaded into the MPU from external sources. One of the subroutines contained in the EPROMs would be a program that reads data into the MPU via the PIA (M6820). Data to be read into the MPU would be loaded into the HP 2100 from either the console or a series of programs contained on magnetic tape. This data will be subsequently transferred from the HP 2100 into a memory module contained in the CAMAC system. The transfer of data from the HP to the memory module (4000 16 bit words) will free the HP to perform other functions subsequent to loading the CAMAC. Control parameters will be loaded from the memory module to the 512 bytes of RAM (M6810) contained in the MPU. Direct interrupts will also be provided between the MPU and the HP 2100 computer. The read program which is the subroutine that reads data from the memory module to the microprocessor unit has been assembled and is included in the appendix.

D. Processing Facilities

A new 32 bit minicomputer has been ordered (SEL 32/57) and delivery is expected for the fall of 1980. Purchase of an array processor dedicated to the reconstruction is also being considered and appears likely at this point in time. The resultant upgrading of computing facilites will allow the image to be displayed on the Grinnell imaging system at the same time or shortly after a scan is performed. This means that the validity of the scan can be checked immediately and should another scan be required the patient will still be present.

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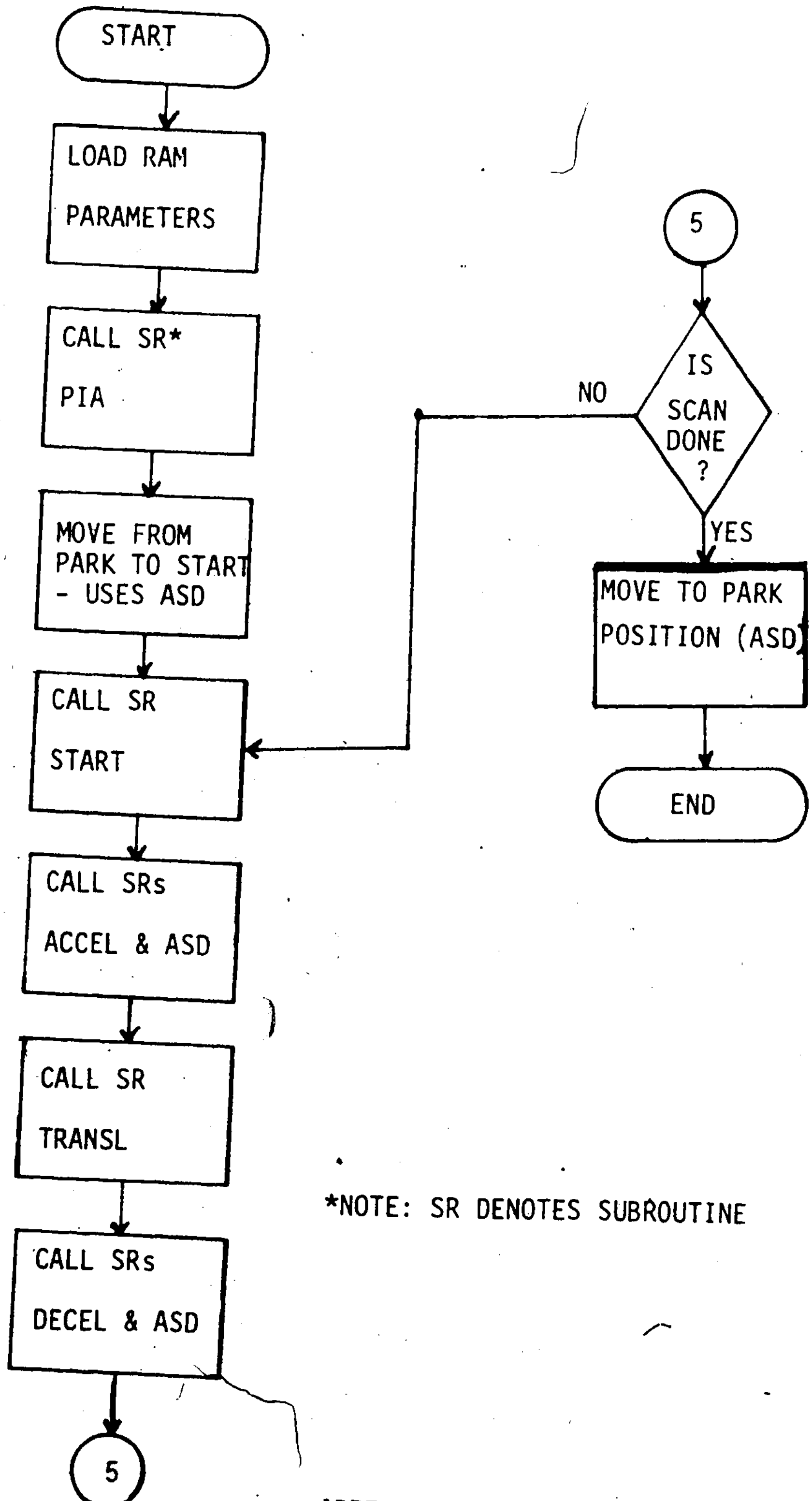
$^{125}_{53}$I DECAY MECHANISM

$^{125}_{53}$I decays via electron capture to $^{125}_{52}$Te and in doing so gives off a 35.4 KeV gamma ray. This gamma ray will:

1) appear on the outside .0666 times per disintegration;
2) cause a k alpha-1 x-ray .7615 times per disintegration (27.4 KeV);
3) cause a k alpha-2 x-ray .3906 times per disintegration (27.2 KeV);
4) cause a k beta-1 x-ray .2056 times per disintegration (30.9 KeV);
5) cause a k beta-2 x-ray .0426 times per disintegration (31.8 KeV);
6) cause an L x-ray .2226 times per disintegration (3.7 KeV).

Thus in general the original gamma ray energizes an electron in the k shell causing it to move out of its orbit. This gap is filled by a loose electron which, in falling into the k shell, gives off an x-ray of approximately 27.5 KeV. The associated half-life is 60.2 days.

NORMAL SCAN



*NOTE: SR DENOTES SUBROUTINE

APPENDIX B.

*** <u>APPENDIX</u> <u>C</u> - CONTAINS ALL MAIN PROGRAMS AND ALL SUBROUTINES (EXCEPT READ) ***

Motorola M68SAM Cross-Assembler

```
      8004   ORA   EQU    $8004
      8004   DDRA  EQU    $8004
      8005   CRA   EQU    $8005
      0000   M     EQU    $0000
                   NAM    DET8
6000               ORG    $6000
             *
             * REVISED APR 24, 1980
             *
             * THIS PROGRAM WILL BE USED TO PERFORM
             * A MULTI-DETECTOR SCAN USING 8 DETECTORS.
             * USING A 1:1 PULLEY RATIO THE TRANSLATION
             * IS 16992 STEPS (106.2 MM).
             * 1 MM = 1600 STEPS.
             * BE SCANNED WITH THIS ARRANGEMENT IS 76.8 MM DIAMETER.
             * THE SCANNER WILL COLLECT 1416 DATA.  FROM THE
             * PARK POSITION THE SCANNER MUST MOVE BACK (-)
             * 1447 STEPS (+200 FOR ACCEL.) BEFORE
             * STARTING THE SCAN.
             * MOTOR PULSES ARE SENT EVERY 466 CYCLES
             * (APPROX. 760 US) AND DATA PULSES ARE SENT TO
             * THE HP EVERY 12 MOTOR PULSES.  THE SCANNER IS TO
             * ROTATE 25.71 DEG. AFTER EACH TRANSLATION & IS
             * TO PERFORM 7 TRANSLATIONS & 6 ROTATIONS.
             * EACH TRANSLATION TAKES APPROX. 13 SECONDS
6000 86 0F         LDA A  #$0F
6002 97 2C         STA A  M+44    ;TRANSLATIONAL DIRECTION
6004 86 07         LDA A  #07
6006 97 2D         STA A  M+45    ;# OF TRANSLATIONS
6008 86 0C         LDA A  #12
600A 97 32         STA A  M+50    ;# OF MOTOR STEPS / HP SIGNAL
600C 86 08         LDA A  #08
600E 97 33         STA A  M+51    ;DETERMINES DELAY IN TRANSLATION
6010 86 04         LDA A  #04
6012 97 34         STA A  M+52    ;DETERMINES DELAY IN TRANSLATION
6014 86 05         LDA A  #05
6016 97 35         STA A  M+53    ;DETERMINES DELAY IN TRANSLATION
6018 BD C278       JSR    PIA     ;CALLS INIT AND F AS WELL
             *
             * MOVE FROM PARK TO START POSITION
             * ONE GROUP OF 1647 PULSES
             * MAKES USE OF THE R21 RAMP
             * 1647=400 + 1247*1
601B 86 17         LDA A  #$17    ; TRANSLATIONAL DIRECTION
```

```
6010 97 00          STA A   M
601F CE 04DF        LDX     #1247
6022 DF 2A          STX     M+42
6024 CE 066F        LDX     #1647
6027 DF 25          STX     M+37
6029 BD C000        JSR     ASD
602C BD C29B ST     JSR     START
602F C6 1B          LDA B   #$1B       ;TELL HP RAMP IS BEGINNING
6031 F7 8004        STA B   ORA
6034 BD C28C        JSR     DELAY
6037 C6 1F          LDA B   #$1F
6039 F7 8004        STA B   ORA
                 * BEGIN ACCELERATION
603C BD C1F5        JSR     ACCEL
603F BD C000        JSR     ASD
                 * SCAN - UP PROVIDES A MOTOR PULSE EVERY 466 CYCLES & A HP PULSE
                 * EVERY 12 MOTOR PULSES
                 * HP PULSE IS SENT EVERY M+46 MOTOR PULSES
6042 CE 4260        LDX     #16992     ;PULSES REQUIRED FOR SCAN
6045 BD C2B0        JSR     TRANSL
                 * DELAY BETWEEN PULSES =4*68 + 42 + 8*12 + 56 = 466 CYCLES
6048 BD C208        JSR     DECEL
604B 7A 002D        DEC     M+45       ;IS SCAN FINISHED?
604E 27 0C          BEQ     JMPE
6050 BD C28D        JSR     R8D
6053 96 2C          LDA A   M+44
6055 88 18          EOR A   #$18       ;CHANGE TRANSLATIONAL DIRECTION
6057 97 2C          STA A   M+44
6059 7E 602C        JMP     ST
                 *MOVE BACK TO PARK POSITION
                 *MUST MOVE BACKWARD 15744 PULSES
                 *SHOULD MOVE BACK 15745
                 *HOWEVER SCAN APPEARS TO BE OUT
                 *ONE PULSE AT SOME POINT
605C 86 17       JMPE   LDA A   #$17
605E 97 00          STA A   M
6060 CE 3BF0        LDX     #15344
6063 DF 2A          STX     M+42
6065 CE 3D80        LDX     #15744
6068 DF 25          STX     M+37
606A BD C000        JSR     ASD
                 * ROTATE BACK TO PARK POSITION
                 * MUST ROTATE BACK 31080 PULSES
606D CE 77D8        LDX     #30680     ;30680*M+12 PULSES
6070 DF 2A          STX     M+42
6072 CE 7968        LDX     #31080
6075 DF 25          STX     M+37
```

```
00096 6077 86 1D          LDA A  #$1D
00097 6079 97 00          STA A  M
00098 607B BD C000        JSR    ASD
00099 607E 3F             SWI
00103 6090                ORG    $6090
00104                * NAM CENT
00105                * THIS IS A PROGRAM WHICH WILL BE USED TO CENTER
00106                * THE SCANNER.  IT'S USE WILL BE REQUIRED WHENEVER ANY OF
00107                * THE MECHANICS OF THE SYSTEM (SUCH AS THE SOURCE)
00108                * ARE CHANGED.  USING A 1:1 RATIO ON THE PULLEYS
00109                * THE TRANSLATIONAL MOTION IS 6500 STEPS
00110                * PULSES ARE SENT TO THE HP EVERY 2 MOTOR PULSES
00111                * (EVER 6.79 MS OR EVERY 4172 CYCLES).
00112                * THE TRANSLATION TAKES 22 SEC. AT WHICH POINT THE
00113                * SCANNER IS ROTATED THROUGH 180 DEG. AND THEN THE TRANSLATION
00114                * IS PERFORMED ONE ADDITIONAL TIME AND THE SCANNER IS ROTATED
00115                * BACK. PRESET FROM THE PARK POSITION IS +3600 -32 FOR ACCEL = 3568 STEPS
00116 6090 86 0F          LDA A  #$0F       ;TRANSLATION DIRECTION
00117 6092 97 2C          STA A  M+44
00118 6094 86 02          LDA A  #02
00119 6096 97 2D          STA A  M+45       ;# OF ROTATIONS
00120 6098 86 02          LDA A  #02
00121 609A 97 32          STA A  M+50       ;PULSE TO HP EVERY 2 STEPS
00122 609C 86 01          LDA A  #01
00123 609E 97 33          STA A  M+51       ;DETERMINES DELAY BETWEEN STEPS
00124 60A0 86 13          LDA A  #19
00125 60A2 97 34          STA A  M+52       ;DETERMINES DELAY BETWEEN STEPS
00126 60A4 86 08          LDA A  #08
00127 60A6 97 35          STA A  M+53       ;DETERMINES DELAY BETWEEN STEPS
00128 60A8 BD C278        JSR    PIA        ;CALLS INIT AND F AS WELL
00130                * MOVE FROM PARK TO START POSITION
00131                * ONE GROUP OF 3568 PULSES
00132                * MAKES USE OF THE R21 RAMP
00133                * 3568 = 400 + 3168
00134 60AB 86 0F          LDA A  #$0F
00135 60AD 97 00          STA A  M
00136 60AF CE 0C60        LDX    #3168
00137 60B2 DF 2A          STX    M+42
00138 60B4 CE 0DF0        LDX    #3568
00139 60B7 DF 25          STX    M+37
00140 60B9 BD C000        JSR    ASD
00141 60BC BD C29B ST1    JSR    START
00142 60BF C6 1B          LDA B  #$1B       ;TELL HP RAMP IS BEGINNING
00143 60C1 F7 8004        STA B  ORA
00144 60C4 BD C28C        JSR    DELAY
00145 60C7 C6 1F          LDA B  #$1F
```

```
6OC9 F7 8004        STA B  ORA
                  *
                  * SCAN - UP PROVIDES A MOTOR PULSE EVERY 2086 CYCLES & A HP PULSE
                  * EVERY 412 CYCLES (EVERY 2 MOTOR PULSES)
                  * HP PULSE IS SENT EVERY M+46 MOTOR PULSES
                  * ACCELERATE UP TO SPEED
6OCC BD C1F5        JSR    ACCEL
6OCF CE 0020        LDX    #32
6OD2 DF 25          STX    M+37
6OD4 BD COOO        JSR    ASD
6OD7 CE 1964        LDX    #6500    ;PULSES REQUIRED FOR SCAN
6ODA BD C2BO        JSR    TRANSL
                  *DELAY BETWEEN PULSES =98 + 12 + 19*104 = 2086 CYCLES
                  *DECEL.
6ODD BD C242        JSR    DEC32
                  * THIS IS A PROGRAM DESIGNED TO OBTAIN A SUITABLE RAMP
                  * FOR ROTATION OF 18O DEGREES.
                  * IT MAKES USE OF A GENERAL SUBROUTINE ASD (ACCELERATION,SPEED,DECELERATION
                  * 35882 PULSES ARE
                  * REQUIRED AT MAX. FREQ.  WE WILL USE 35882 GROUPS
                  * 1 PULSE
6OEO 96 2D          LDA A  M+45
6OE2 81 O2          CMP A  #O2
6OE4 27 O4          BEQ    K5
6OE6 86 1D          LDA A  #$1D
6OE8 97 OO          STA A  M
6OEA 7A OO2D K5     DEC    M+45
6OED CE 8C2A        LDX    #35882   ;35882*M+12
6OFO DF 2A          STX    M+42
6OF2 CE 8DBA        LDX    #36282
6OF5 DF 25          STX    M+37
6OF7 BD COOO        JSR    ASD
6OFA 96 2D          LDA A  M+45
6OFC 27 O9          BEQ    JMPE1
6OFE 96 2C          LDA A  M+44
61OO 88 18          EOR A  #$18     ;CHANGE TRANSLATION DIRECTION
61O2 97 2C          STA A  M+44
61O4 7E 6OBC        JMP    ST1
                  * MOVE BACK TO PARK POSITION
                  * MUST MOVE FORWARD 3568 STEPS
61O7 86 17   JMPE1  LDA A  #$17
61O9 97 OO          STA A  M
61OB CE ODFO        LDX    #3568
61OE DF 25          STX    M+37
611O CE OC6O        LDX    #3168
6113 DF 2A          STX    M+42
```

```
6115 BD C000          JSR    ASD
                    * NAM SSCAN
6140                  ORG    $6140
                    * (SCOUT SCAN)
                    * THIS SCAN IS USED TO AID IN ACCURATE REPOSITIONING
                    * OF THE ARM.  THE SCANNER WILL ROTATE 90 DEGREES, PRESET
                    * A DISTANCE OF +3125 - 200 (FOR ACCEL.) STEPS FROM THE PARK POSITION
                    * AND THEN START MEASUREMENTS.  THERE WILL BE 26 MEASUREMENTS
                    * EACH CONSISTING OF A SINGLE SCAN OF 7848 STEPS AND 654 INTERVALS
                    * (760 US / STEP).  THERE IS A 2 SEC. DELAY BETWEEN EACH MEASUREMENT
                    * TO ALLOW THE ARM POSITION TO BE ADJUSTED.  THE SCANNER IS THEN SET
                    * BACK TO THE PARK POSITION AND IS ROTATED BACK TO ITS ORIGINAL POSITION
6140 86 OF            LDA A  #$OF
6142 97 2C            STA A  M+44      ;TRANSLATIONAL DIRECTION
6144 86 1A            LDA A  #26
6146 97 2D            STA A  M+45      ;# OF TRANSLATIONS
6148 86 OC            LDA A  #12
614A 97 32            STA A  M+50      ;# OF MOTOR STEPS / HP SIGNAL
614C 86 08            LDA A  #08
614E 97 33            STA A  M+51      ;DETERMINES DELAY IN TRANSLATION
6150 86 04            LDA A  #04
6152 97 34            STA A  M+52      ;DETERMINES DELAY IN TRANSLATION
6154 86 05            LDA A  #05
6156 97 35            STA A  M+53      ;DETERMINES DELAY IN TRANSLATION
6158 BD C278          JSR    PIA       ;CALLS INIT AND F AS WELL
                    * 90 DEGREES=18141 STEPS
615B CE 454D          LDX    #17741    ;17741*M+12 PULSES
615E DF 2A            STX    M+42
6160 CE 46DD          LDX    #18141
6163 DF 25            STX    M+37
6165 BD C000          JSR    ASD
                    * MOVE FROM PARK TO START POSITION
                    * ONE GROUP OF 2925 PULSES
                    * MAKES USE OF THE R21 RAMP
                    * 2925 = 400 + 2525*1
6168 86 OF            LDA A  #$OF      ; TRANSLATIONAL DIRECTION
616A 97 00            STA A  M
616C CE 09DD          LDX    #2525
616F DF 2A            STX    M+42
6171 CE 0B6D          LDX    #2925
6174 DF 25            STX    M+37
6176 BD C000          JSR    ASD
6179 BD C29B ST2      JSR    START
617C C6 1B            LDA B  #$1B      ;TELL HP RAMP IS BEGINNING
617E F7 8004          STA B  ORA
```

```
00246 6181 BD C28C          JSR    DELAY
00247 6184 C6 1F            LDA B  #$1F
00248 6186 F7 8004          STA B  ORA
00249                * BEGIN ACCELERATION
00250 6189 BD C1F5          JSR    ACCEL
00251 618C BD C000          JSR    ASD
00253                * SCAN - UP PROVIDES A MOTOR PULSE EVERY 1010 CYCLES & A HP PULSE
00254                * EVERY 12 MOTOR PULSES
00255                * HP PULSE IS SENT EVERY M+46 MOTOR PULSES
00256 618F CE 1EA8          LDX    #7848     ;PULSES REQUIRED FOR SCAN
00257 6192 BD C2B0          JSR    TRANSL
00258                * DELAY BETWEEN PULSES =152 + 42 + 68*4=466 CYCLES
00259                * DECELERATION
00260 6195 BD C208          JSR    DECEL
00261 6198 7A 002D          DEC    M+45      ;IS SCAN FINISHED?
00262 619B 27 18            BEQ    JMPE2
00263 619D 96 2C            LDA A  M+44
00264 619F 88 18            EOR A  #$18      ;CHANGE TRANSLATIONAL DIRECTION
00265 61A1 97 2C            STA A  M+44
00267                * 2 SECOND DELAY BETWEEN MEASUREMENTS
00268 61A3 CE B89C          LDX    #47260
00269 61A6 78 0029   MC1    ASL    M+41      ;6
00270 61A9 78 0029          ASL    M+41      ;6
00271 61AC 78 0029          ASL    M+41      ;6
00272 61AF 09               DEX              ;4
00273 61B0 26 F4            BNE    MC1       ;4
00274 61B2 7E 6179          JMP    ST2
00276                *MOVE BACK TO PARK POSITION
00277                *MUST MOVE FORWARD 2925
00278 61B5 86              JMPE2  LDA A  #$17
00279 61B7 97 00            STA A  M
00280 61B9 CE 09D           LDX    #2525
00281 61BC DF 2A            STX    M+42
00282 61BE CE 0B6D          LDX    #2925
00283 61C1 DF 25            STX    M+37
00284 61C3 BD C00C          JSR    ASD
00285                * ROTATE BACK 90 DEGREES
00286 61C6 CE 454D          LDX    #17741
00287 61C9 DF 2A            STX    M+42
00288 61CB CE 46DD          LDX    #18141
00289 61CE DF 25            STX    M+37
00290 61D0 86 1D            LDA A  #$1D
00291 61D2 97 00            STA A  M
00292 61D4 BD C000          JSR    ASD
00293 61D7 3F               SWI
```

```
61E0              ORG    $61E0
                * NAM 8DETM
                * THIS PROGRAM WILL BE USED TO PERFORM
                * A MULTI-DETECTOR SCAN USING 8 DETECTORS.
                * USING A 1:1 PULLEY RATIO THE TRANSLATION
                * IS 16992 STEPS (106.2 MM)  IT WILL COLLECT 1416 DATA. FROM THE
                * PARK POSITION THE SCANNER M''ST MO'E BACK
                * 1447 STEPS +32 FOR ACCEL.) ʙEF(   STARTING THE SCAN.
                * MOTOR PULSES ARE SENT EVERY 1862 CYCLES
                * (APPROX. 3030.6 US) AND DATA PULSES ARE SENT TO
                * THE HP EVERY 12 MOTOR PULSES.  THᴇ SCANNER IS TO
                * ROTATE 25.71 DEG. AFTER EACH TRANSLATION & IS
                * TO PERFORM 7 TRANSLATIONS & 6 ROTATIONS.
61E0 86 0F              LDA A  #$0F
61E2 97 2C              STA A  M+44     ;TRANSLATIONAL DIRECTION
61E4 86 07              LDA A  #07
61E6 97 2D              STA A  M+45     ;# OF TRANSLATIONS
61E8 86 0C              LDA A  #12
61EA 97 32              STA A  M+50     ;# OF MOTOR STEPS / HP SIGNAL
61EC 86 07              LDA A  #07
61EE 97 33              STA A  M+51     ;DETERMINES DELAY IN TRANSLATION
61F0 86 54              LDA A  #84
61F2 97 34              STA A  M+52     ;DETERMINES DELAY IN TRANSLATION
61F4 86 01              LDA A  #01
61F6 97 35              STA A  M+53     ;DETERMINES DELAY IN TRANSLATION
                * MOVE FROM PARK TO START POSITION
                * ONE GROUP OF 1479 PULSES
                * MAKES USE OF THE R21 RAMP
                * 1479=400 + 1079*1
61F8 BD C278            JSR    PIA      ; CALLS INIT AND F AS WELL
61FB 86 17              LDA A  #$17     ; TRANSLATIONAL DIRECTION
61FD 97 00              STA A  M
61FF CE 0437            LDX    #1079
6202 DF 2A              STX    M+42
6204 CE 05C7            LDX    #1479
6207 DF 25              STX    M+37
6209 BD C000            JSR    ASD
620C BD C29B ST3        JSR    START
620F C6 1B              LDA B  #$1B     ;TELL HP RAMP IS BEGINNING
6211 F7 8004            STA B  ORA
6214 BD C28C            JSR    DELAY
6217 C6 1F              LDA B  #$1F
6219 F7 8004            STA B  ORA
621C BD C1F5            JSR    ACCEL
621F CE 0020            LDX    #32
6222 DF 25              STX    M+37
6224 BD C000            JSR    ASD
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00347                 * SCAN - UP PROVIDES A MOTOR PULSE EVERY 1862 CYCLES & A HP PULSE
00348                 * EVERY 12 MOTOR PULSES
00349                 * HP PULSE IS SENT EVERY M+46 MOTOR PULSES
00350 6227 CE 4260          LDX    #16992   ;PULSES REQUIRED FOR SCAN
00351 622A BD C2B0          JSR    TRANSL
00352                 * DELAY BETWEEN PULSES = 84*(1*12+8)+42+7*12+6 = 1862 CYCLES
00353                 * DECELERATION
00354 622D BD C242          JSR    DEC32
00355 6230 7A 002D          DEC    M+45     ;IS SCAN FINISHED?
00356 6233 27 0C            BEQ    JMPE3
00357 6235 BD C28D          JSR    R8D
00358 6238 96 2C            LDA A  M+44
00359 623A 88 18            EOR A  #$18     ;CHANGE TRANSLATIONAL DIRECTION
00360 623C 97 2C            STA A  M+44
00361 623E 7E 620C          JMP    ST3
00363                 *MOVE BACK TO PARK POSITION
00364                 *SHOULD MOVE BACK 15577
00365                 *HOWEVER APPEARS TO BE OUT ONE PULSE
00366 6241 86 17     JMPE3  LDA A  #$17
00367 6243 97 00            STA A  M
00368 6245 CE 3B48          LDX    #15176
00369 6248 DF 2A            STX    M+42
00370 624A CE 3CD8          LDX    #15576
00371 624D DF 25            STX    M+37
00372 624F BD C000          JSR    ASD
00373                 * ROTATE BACK TO PARK POSITION
00374                 * MUST ROTATE BACK 31080 PULSES
00375 6252 CE 77D8          LDX    #30680   ;30680*M+12 PULSES
00376 6255 DF 2A            STX    M+42
00377 6257 CE 7968          LDX    #31080
00378 625A DF 25            STX    M+37
00379 625C 86 1D            LDA A  #$1D
00380 625E 97 00            STA A  M
00381 6260 BD C000          JSR    ASD
00382 6263 3F               SWI
00386 6280                  ORG    $6280
00387                 * NAM 8DETS
00388                 * THIS PROGRAM WILL BE USED TO PERFORM
00389                 * A MULTI-DETECTOR SCAN USING 8 DETECTORS.
00390                 * USING A 1:1 PULLEY RATIO THE TRANSLATION
00391                 * IS 16992 STEPS (106.2 MM). IT WILL COLLECT 1416 DATA. FROM THE
00392                 * PARK POSITION THE SCANNER MUST MOVE BACK
00393                 * 1447 STEPS +5 FOR ACCEL.) BEFORE
00394                 * STARTING THE SCAN.
00395                 * MOTOR PULSES ARE SENT EVERY 7458 CYCLES
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                       * (APPROX. 12139 US) AND DATA PULSES ARE SENT TO
                       * THE HP EVERY 12 MOTOR PULSES.  THE SCANNER IS TO
                       * ROTATE 25.71 DEG. AFTER EACH TRANSLATION & IS
                       * TO PERFORM 7 TRANSLATIONS & 6 ROTATIONS.
6280 86 OF                   LDA A  #$OF
6282 97 2C                   STA A  M+44     ;TRANSLATIONAL DIRECTION
6284 86 07                   LDA A  #07
6286 97 2D                   STA A  M+45     ;# OF TRANSLATIONS
6288 86 OC                   LDA A  #12
628A 97 32                   STA A  M+50     ;# OF MOTOR STEPS / HP SIGNAL
628C 86 08                   LDA A  #08
628E 97 33                   STA A  M+51     ;DETERMINES DELAY IN TRANSLATION
6290 86 E3                   LDA A  #227
6292 97 34                   STA A  M+52     ;DETERMINES DELAY IN TRANSLATION
6294 86 02                   LDA A  #02
6296 97 35                   STA A  M+53     ;DETERMINES DELAY IN TRANSLATION
                       * MOVE FROM PARK TO START POSITION
                       * ONE GROUP OF 1452 PULSES
                       * 1452=400 + 1052*1
6298 BD C278                 JSR    PIA      ; CALLS INIT AND F AS WELL
629B 86 17                   LDA A  #$17     ; TRANSLATIONAL DIRECTION
629D 97 00                   STA A  M
629F CE 041C                 LDX    #1052
62A2 DF 2A                   STX    M+42
62A4 CE 05AC                 LDX    #1452
62A7 DF 25                   STX    M+37
62A9 BD C000                 JSR    ASD
62AC BD C29B      ST4        JSR    START
62AF C6 1B                   LDA B  #$1B     ;TELL HP RAMP IS BEGINNING
62B1 F7 8004                 STA B  ORA
62B4 BD C28C                 JSR    DELAY
62B7 C6 1F                   LDA B  #$1F
62B9 F7 8004                 STA B  ORA
                       * BEGIN ACCELERATION
62BC BD C1F5                 JSR    ACCEL
62BF CE 0005                 LDX    #05
62C2 DF 25                   STX    M+37
62C4 BD C000                 JSR    ASD
                       * SCAN - UP PROVIDES A MOTOR PULSE EVERY 7458 CYCLES & A HP PULSE
                       * EVERY 12 MOTOR PULSES
                       * HP PULSE IS SENT EVERY M+46 MOTOR PULSES
62C7 CE 4260                 LDX    #16992   ;PULSES REQUIRED FOR SCAN
62CA BD C280                 JSR    TRANSL
                       * DELAY BETWEEN PULSES = 227(2*12+8)+42+56+8*12 = 7458 CYCLES
                       * DECELERATION
62CD BD C25D                 JSR    DEC5
62D0 7A 002D                 DEC    M+45     ;IS SCAN FINISHED?
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62D3 27 0C          BEQ    JMPE4
62D5 BD C28D        JSR    R8D
62D8 96 2C          LDA A  M+44
62DA 88 18          EOR A  #$18     ;CHANGE TRANSLATIONAL DIRECTION
62DC 97 2C          STA A  M+44
62DE 7E 62AC        JMP    ST4
                *MOVE BACK TO PARK POSITION
                *SHOULD MOVE BACK 15550
                *HOWEVER APPEARS TO BE OUT ONE PULSE
62E1 86 17      JMPE4 LDA A  #$17
62E3 97 00          STA A  M
62E5 CE 3B2D        LDX    #15149
62E8 DF 2A          STX    M+42
62EA CE 3CBD        LDX    #15549
62ED DF 25          STX    M+37
62EF BD C000        JSR    ASD
                * ROTATE BACK TO PARK POSITION
                * MUST ROTATE BACK 31080 PULSES
62F2 CE 77D8        LDX    #30680   ;30680*M+12 PULSES
62F5 DF 2A          STX    M+42
62F7 CE 7968        LDX    #31080
62FA DF 25          STX    M+37
62FC 86 1D          LDA A  #$1D
62FE 97 00          STA A  M
6300 BD C000        JSR    ASD
6303 3F             SWI
6320                ORG    $6320
                * NAM RAT
                * THIS PROGRAM WILL BE USED TO PERFORM
                * A SINGLE DETECTOR SCAN.
                * USING A 1:1 PULLEY RATIO THE TRANSLATION
                * IS 2560 STEPS (16.0 MM). IT WILL COLLECT 128 DATA. FROM THE
                * PARK POSITION THE SCANNER MUST MOVE FORWARD(+)
                * 6500 STEPS -0 FOR ACCEL. BEFORE
                * STARTING THE SCAN.
                * MOTOR PULSES ARE SENT EVERY 15,358 CYCLES
                * (APPROX. 24,997 US) AND DATA PULSES ARE SENT TO
                * THE HP EVERY 20 MOTOR PULSES.  THE SCANNER IS TO
                * ROTATE 3.214 DEG. AFTER EACH TRANSLATION & IS
                * TO PERFORM 56 TRANSLATIONS & 55 ROTATIONS.
6320 86 0F          LDA A  #$0F
6322 97 2C          STA A  M+44     ;TRANSLATIONAL DIRECTION
6324 86 38          LDA A  #56
6326 97 2D          STA A  M+45     ;# OF TRANSLATIONS
6328 86 14          LDA A  #20
632A 97 32          STA A  M+50     ;# OF MOTOR STEPS / HP SIGNAL
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6332C 86 08            LDA A  #08
632E 97 33             STA A  M+51      ;DETERMINES DELAY IN TRANSLATION
6330 86 DF             LDA A  #223
6332 97 34             STA A  M+52      ;DETERMINES DELAY IN TRANSLATION
6334 86 05             LDA A  #05
6336 97 35             STA A  M+53      ;DETERMINES DELAY IN TRANSLATION
                 * MOVE FROM PARK TO START POSITION
                 * ONE GROUP OF 6500 PULSES
                 * MAKES USE OF THE R21 RAMP
                 * 6500=400 + 6100*1
6338 BD C278           JSR    PIA       ; CALLS INIT AND F AS WELL
633B 86 0F             LDA A  #$0F      ; TRANSLATIONAL DIRECTION (+)
633D 97 00             STA A  M
633F CE 17D4           LDX    #6100
6342 DF 2A             STX    M+42
6344 CE 1964           LDX    #6500
6347 DF 25             STX    M+37
6349 BD C000           JSR    ASD
634C BD C29B     ST5   JSR    START
634F C6 1B             LDA B  #$1B      ;TELL HP RAMP IS BEGINNING
6351 F7 8004           STA B  ORA
6354 BD C28C           JSR    DELAY
6357 C6 1F             LDA B  #$1F
6359 F7 8004           STA B  ORA
                 * SCAN - UP PROVIDES A MOTOR PULSE EVERY 15,358 CYCLES & A HP PULSE
                 * EVERY 20 MOTOR PULSES
                 * HP PULSE IS SENT EVERY M+46 MOTOR PULSES
635C CE 0A00           LDX    #2560     ;PULSES REQUIRED FOR SCAN
635F BD C2B0           JSR    TRANSL
                 * DELAY BETWEEN PULSES = 223(5*12+8)+42+56+8*12 = 15358 CYCLES
6362 7A 002D           DEC    M+45      ;IS SCAN FINISHED?
6365 27 1A             BEQ    JMPE5
                 * ROTATE 3.214 DEG. (648 PULSES)
6367 86 1E             LDA A  #$1E
6369 97 00             STA A  M
636B CE 00F8           LDX    #248      ;248 * M+12 PULSES
636E DF 2A             STX    M+42
6370 CE 0288           LDX    #648
6373 DF 25             STX    M+37
6375 BD C000           JSR    ASD
6378 96 2C             LDA A  M+44
637A 88 18             EOR A  #$18      ;CHANGE TRANSLATIONAL DIRECTION
637C 97 2C             STA A  M+44
637E 7E 634C           JMP    ST5
                 *MOVE BACK TO PARK POSITION
                 *SHOULD MOVE BACK 6500
                 *HOWEVER APPEARS TO BE OUT ONE PULSE
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6381 86 17   JMPE5  LDA A  #$17
6383 97 00        . STA A  M
6385 CE 17D3        LDX    #6099
6388 DF 2A          STX    M+42
638A CE 1963        LDX    #6499
638D DF 25          STX    M+37
638F BD C000        JSR    ASD
                  * ROTATE BACK TO PARK POSITION
                  * MUST ROTATE BACK 35640 PULSES
6392 CE 89A8        LDX    #35240    ;35240*M+12 PULSES
6395 DF 2A          STX    M+42
6397 CE 8B38        LDX    #35640
639A DF 25          STX    M+37
639C 86 1D          LDA A  #$1D
639E 97 00          STA A  M
63A0 BD C000        JSR    ASD
63A3 3F             SWI
C000                ORG    $C000
                  * NAM SUBROUTINES
                  * SUBROUTINE ASD (ACCELERATION,SPEED,
                  * DECELERATION)
                  * MUST HAVE PARAMETERS SET INTO M THROUGH M+43 BEFORE INITIATION.
                  * M IS THE TYPE OF MOTION AND DIRECTION ($1D & $1E ARE ROTATION WHILE
                  * $0F & $17 ARE TRANSLATION).  M+1 THROUGH M+23 ARE
                  * THE LENGTHS OF THE RAMPS INVOLVED
                  * M+25 THROUGH M+36 ARE THE FREQUENCY OF THE PULSES ASSOCIATED WITH
                  * THE SETS OF RAMPS
                  * WHILE M+24 IS A SPARE LOCATION.
                  * M+37  M+38 IS THE TOTAL # OF PULSES NEEDED FOR THE DESIRED MOTION
                  * M+39 IS A PROGRAM LOCATION DEVICE
                  * M+40 IS USED TO STORE ACC A
                  * M+41 IS MERELY A DELAY MECHANISM
                  * M+42 M+43 = THE # OF TIMES THE RAMP AT
                  * MAX FREQ. IS TO BE PERFORMED
C000 CE 0000 ASD    LDX    #$0000
C003 7F 0027        CLR    M+39
C006 7F 0028        CLR    M+40
C009 08      ASD0   INX              ;4
C00A 96 28          LDA A  M+40      ;3
C00C 4C             INC A            ;2
C00D 97 28          STA A  M+40      ;4
                  * THE FOLLOWING BRANCH TO C1 THROUGH C9 SINCE BRANCHES TO
                  * B? THROUGH B8 WOULD BE OUT OF RANGE
C00F 91 01          CMP A  M+1       ;3
C011 2F 23          BLE    C1
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00596 C013 91 02            CMP A  M+2
00597 C015 2F 22            BLE    C2
00598 C017 91 03            CMP A  M+3
00599 C019 2F 21            BLE    C3
00600 C01B 91 04            CMP A  M+4
00601 C01D 2F 20            BLE    C4
00602 C01F 91 05            CMP A  M+5
00603 C021 2F 1F            BLE    C5
00604 C023 91 06            CMP A  M+6
00605 C025 2F 1E            BLE    C6
00606 C027 91 07            CMP A  M+7
00607 C029 2F 1D            BLE    C7
00608 C02B 91 08            CMP A  M+8
00609 C02D 2F 1C            BLE    C8
00610 C02F 91 09            CMP A  M+9
00611 C031 2F 1B            BLE    C9
00612 C033 7E C051          JMP    J5
00613 C036 7E C0E7 C1       JMP    B1
00614 C039 7E C0EC C2       JMP    B2
00615 C03C 7E C0F1 C3       JMP    B3
00616 C03F 7E C0F6 C4       JMP    B4
00617 C042 7E C0FB C5       JMP    B5
00618 C045 7E C100 C6       JMP    B6
00619 C048 7E C105 C7       JMP    B7
00620 C04B 7E C10A C8       JMP    B8
00621 C04E 7E C10F C9       JMP    B9
00623                  * THE FOLLOWING (PRE BLE B10) IS NECESSARY SINCE THE BLE INSTRUCTION
00624                  * USES TWO S COMPLEMENT ARITMETIC WHICH RESTRICTS THE
00626                  * SIZE OF AN 8 BIT NUMBER TO +127 & THE SIZE OF A RAMP SPEC TO +126
00628 C051 86 01   J5       LDA A  #01
00629 C053 97 27            STA A  M+39
00630 C055 7F 0028          CLR    M+40
00631 C058 09               DEX
00632 C059 96 28   AS       LDA A  M+40
00633 C05B 4C               INC A
00634 C05C 97 28            STA A  M+40
00635 C05E 08               INX
00636 C05F 91 0A            CMP A  M+10
00637 C061 2F 03            BLE    C10
00638 C063 7E C069          JMP    K1
00639 C066 7E C114 C10      JMP    B10          ;B10 IS OUT OF RANGE FOR BRANCH
00640 C069 91 0B   K1       CMP A  M+11
00641 C06B 2F 03            BLE    C11
00642 C06D 7E C073          JMP    K2
00643 C070 7E C119 C11      JMP    B11          ;B11 IS OUT OF RANGE FOR BRANCH
00644 C073 86 02   K2       LDA A  #02
00645 C075 97 27            STA A  M+39
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C077 7F 0028 P1     CLR    M+40
C07A 09             DEX
C07B D6 2B          LDA B  M+43
C07D 27 06          BEQ    P4
C07F 7A 002B        DEC    M+43
C082 7E C08F        JMP    ASD2
C085 D6 2A   P4     LDA B  M+42
C087 27 16          BEQ    P6
C089 7A 002A        DEC    M+42
C08C 7A 002B        DEC    M+43
C08F 96 28   ASD2   LDA A  M+40
C091 4C             INC A
C092 97 28          STA A  M+40
C094 08             INX
C095 91 0C          CMP A  M+12
C097 2F 03          BLE    C12
C099 7E C077        JMP    P1
C09C 7E C11E C12    JMP    B12      ; B12 IS OUT OF RANGE FOR A BRANCH
C09F 86 03   P6     LDA A  #03
C0A1 97 27          STA A  M+39
C0A3 7F 0028        CLR    M+40
C0A6 09             DEX
C0A7 96 28   ASD3   LDA A  M+40
C0A9 4C             INC A
C0AA 97 28          STA A  M+40
C0AC 08             INX
C0AD 91 0D          CMP A  M+13
C0AF 2F 68          BLE    B11
C0B1 91 0E          CMP A  M+14
C0B3 2F 5F          BLE    B10
C0B5 91 0F          CMP A  M+15
C0B7 2F 56          BLE    B9
C0B9 86 04          LDA A  #04
C0BB 97 27          STA A  M+39
C0BD 7F 0028        CLR    M+40
C0C0 09             DEX
C0C1 96 28   ASD4   LDA A  M+40
C0C3 4C             INC A
C0C4 97 28          STA A  M+40
C0C6 08             INX
C0C7 91 10          CMP A  M+16
C0C9 2F 3F          BLE    B8
C0CB 91 11          CMP A  M+17
C0CD 2F 36          BLE    B7
C0CF 91 12          CMP A  M+18
C0D1 2F 2D          BLE    B6
C0D3 91 13          CMP A  M+19
C0D5 2F 24          BLE    B5
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C0D7 91 14            CMP A   M+20
C0D9 2F 1B            BLE     B4
C0DB 91 15            CMP A   M+21
C0DD 2F 12            BLE     B3
C0DF 91 16            CMP A   M+22
C0E1 2F 09            BLE     B2
C0E3 91 17            CMP A   M+23
C0E5 2F 00            BLE     B1
C0E7 96 19      B1    LDA A   M+25        ;3
C0E9 7E C120          JMP     J           ;3
C0EC 96 1A      B2    LDA A   M+26
C0EE 7E C120          JMP     J
C0F1 96 1B      B3    LDA A   M+27
C0F3 7E C120          JMP     J
C0F6 96 1C      B4    LDA A   M+28
C0F8 7E C120          JMP     J
C0FB 96 1D      B5    LDA A   M+29
C0FD 7E C120          JMP     J
C100 96 1E      B6    LDA A   M+30
C102 7E C120          JMP     J
C105 96 1F      B7    LDA A   M+31
C107 7E C120          JMP     J
C10A 96 20      B8    LDA A   M+32
C10C 7E C120          JMP     J
C10F 96 21      B9    LDA A   M+33
C111 7E C120          JMP
C114 96 22      B10   LDA A   M+34
C116 7E C120          JMP     J
C119 96 23      B11   LDA A   M+35
C11B 7E C120          JMP     J
C11E 96 24      B12   LDA A   M+36
C120 C6 1F      J     LDA B   #$1F        ;2
C122 F7 8004          STA B   ORA         ;5
C125 C6 09      DEL   LDA B   #09         ;2
C127 78 0029    D1    ASL     M+41        ;6     )
C12A 5A               DEC B               ;2     )B*12=9*12=108
C12B 26 FA            BNE     D1          ;4     )
C12D 8C 0000          CPX     #0000       ;3
C130 4A               DEC A               ;2
C131 26 F2            BNE     DEL         ;4
C133 D6 00            LDA B   M           ;3
C135 F7 8004          STA B   ORA         ;5
C138 9C 25            CPX     M+37        ; IS MOTION COMPLETE ? 4
C13A 27 21            BEQ     Z2
C13C 96 27            LDA A   M+39
C13E 81 01            CMP A   #01         ;2
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C140 27 0F          BEQ    Z3       ;4
C142 81 02          CMP A  #02      ;2
C144 27 0E          BEQ    Z4       ;4
C146 81 03          CMP A  #03      ;2
C148 27 00          BEQ    Z5       ;4
C14A 81 04          CMP A  #04      ;2
C14C 27 0C          BEQ    Z6       ;4
C14E 7E C009        JMP    ASD0     ;3
C151 7E C059 Z3     JMP    ASD1
C154 7E C08F Z4     JMP    ASD2
C157 7E C0A7 Z5     JMP    ASD3
C15A 7E C0C1 Z6     JMP    ASD4
C15D C6 1F   Z2     LDA B  #$1F
C15F F7 8004        STA B  ORA
C162 39             RTS
             * THE FOLLOWING FREQUENCIES ARE APPROXIMATE ONLY
             * & ARE SLIGHTLY DIFFERENT FOR ACCELERATION & DECELERATION
             * F IS THE REQUIRED FREQUENCIES FOR ACCELERATION/DECELERATION
C163 86 FF   F      LDA A  #255     ;20.2HZ
C165 97 19          STA A  M+25
C167 86 81          LDA A  #129     ;39.9HZ
C169 97 1A          STA A  M+26
C16B 86 4A          LDA A  #74      ;69.3 HZ
C16D 97 1B          STA A  M+27
C16F 86 2F          LDA A  #47      ;108.6 HZ
C171 97 1C          STA A  M+28
C173 86 22          LDA A  #34      ;148.9 HZ
C175 97 1D          STA A  M+29
C177 86 13          LDA A  #19      ;261.4 HZ
C179 97 1E          STA A  M+30
C17B 86 0C          LDA A  #12      ;402.3 HZ
C17D 97 1F          STA A  M+31
C17F 86 09          LDA A  #9       ;548 HZ
C181 97 20          STA A  M+32
C183 86 07          LDA A  #7       ;677 HZ
C185 97 21          STA A  M+33
C187 86 05          LDA A  #5       ;917 HZ
C189 97 22          STA A  M+34
C18B 86 04          LDA A  #4       ;1115 HZ
C18D 97 23          STA A  M+35
C18F 86 03          LDA A  #3       ;1422 HZ
C191 97 24          STA A  M+36
C193 39             RTS
             * INIT. GIVES THE LENGTHS OF RAMPS
             * 360 DEGREES .= 360/.9*464/128*50.044=72564 STEPS
             * 25.7 DEGREES=5180.25 STEPS
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                * 50.044 IS APPROX. THE RATIO OF THE WORM GEAR
C194 86 1E      INIT    LDA A   #$1E
C196 97 00              STA A   M
C198 86 01              LDA A   #01
C19A 97 01              STA A   M+1
C19C 86 02              LDA A   #02
C19E 97 02              STA A   M+2
C1A0 86 05              LDA A   #05
C1A2 97 03              STA A   M+3
C1A4 86 0C              LDA A   #12
C1A6 97 04              STA A   M+4
C1A8 86 14              LDA A   #20
C1AA 97 05              STA A   M+5
C1AC 86 20              LDA A   #32
C1AE 97 06              STA A   M+6
C1B0 86 32              LDA A   #50
C1B2 97 07              STA A   M+7
C1B4 86 53              LDA A   #83
C1B6 97 08              STA A   M+8
C1B8 86 76              LDA A   #118
C1BA 97 09              STA A   M+9
C1BC 86 28              LDA A   #40
C1BE 97 0A              STA A   M+10
C1C0 86 52              LDA A   #82
C1C2 97 0B              STA A   M+11
C1C4 86 01              LDA A   #01
C1C6 97 0C              STA A   M+12
C1C8 86 2A              LDA A   #42
C1CA 97 0D              STA A   M+13
C1CC 86 52              LDA A   #82
C1CE 97 0E              STA A   M+14
C1D0 86 75              LDA A   #117
C1D2 97 0F              STA A   M+15
C1D4 86 21              LDA A   #33
C1D6 97 10              STA A   M+16
C1D8 86 33              LDA A   #51
C1DA 97 11              STA A   M+17
C1DC 86 3F              LDA A   #63
C1DE 97 12              STA A   M+18
C1E0 86 47              LDA A   #71
C1E2 97 13              STA A   M+19
C1E4 86 4E              LDA A   #78
C1E6 97 14              STA A   M+20
C1E8 86 51              LDA A   #81
C1EA 97 15              STA A   M+21
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C1EC 86 52              LDA A  #82
C1EE 97 16              STA A  M+22
C1F0 86 53              LDA A  #83
C1F2 97 17              STA A  M+23
C1F4 39                 RTS
                * ACCELERATION
C1F5 BD C194    ACCEL   JSR    INIT     ;INIT MAY NOT BE REQUIRED (I.E. THIS IS A CHECK)
C1F8 96 2C              LDA A  M+44
C1FA 97 00              STA A  M

C1FC 7F 002A            CLR    M+42
C1FF 7F 002B            CLR    M+43
C202 CE 00C8            LDX    #200
C205 DF 25              STX    M+37
C207 39                 RTS
                * DECELERATION
C208 96 2C      DECEL   LDA A  M+44
C20A 97 00              STA A  M
C20C 7F 0001            CLR    M+1
C20F 7F 0002            CLR    M+2
C212 7F 0003            CLR    M+3
C215 7F 0004            CLR    M+4
C218 7F 0005            CLR    M+5
C21B 7F 0006            CLR    M+6
C21E 7F 0007            CLR    M+7
C221 7F 0008            CLR    M+8
C224 7F 0009            CLR    M+9
C227 7F 000A            CLR    M+10
C22A 7F 000B            CLR    M+11
C22D 7F 000C            CLR    M+12
C230 7F 002A            CLR    M+42
C233 7F 002B            CLR    M+43
C236 CE 00C8            LDX    #200
C239 DF 25              STX    M+37
C23B BD C000            JSR    ASD
C23E BD C194            JSR    INIT
C241 39                 RTS
C242 CE 0053    DEC32   LDX    #83      ;32+51
C245 DF 25              STX    M+37
C247 86 33              LDA A  #51
C249 97 28              STA A  M+40
C24B CE 0033            LDX    #51
C24E 86 04              LDA A  #04
C250 97 27              STA A  M+39
C252 96 2C              LDA A  M+44
C254 97 00              STA A  M
C256 BD C0C1            JSR    ASD4
```

```
C259 BD C194        JSR    INIT
C25C 39             RTS
               * USED FOR DECELERATION FROM 100 HZ. < F. MAX. <400 HZ.
C25D CE 0053 DEC5   LDX    #83      ;78+5
C260 DF 25          STX    M+37
C262 86 4E          LDA A  #78
C264 97 28          STA A  M+40
C266 CE 004E        LDX    #78
C269 86 04          LDA A  #04
C26B 97 27          STA A  M+39
C26D 96 2C          LDA A  M+44
C26F 97 00          STA A  M
C271 BD C0C1        JSR    ASD4
C274 BD C194        JSR    INIT
C277 39             RTS
               * USED FOR DECELERATION FROM F. MAX. < 100 HZ.
C278 7F 8005 PIA    CLR    CRA      ;SET CRA FOR DDRA
C27B C6 1F          LDA B  #$1F     ;BITS 0-4 O/P,5-7 I/P
C27D F7 8004        STA B  DDRA
C280 C6 04          LDA B  #04      ;SET CRA FOR ORA
C282 F7 8005        STA B  CRA
C285 BD C194        JSR    INIT
C288 BD C163        JSR    F
C28B 39             RTS
               * U P HAS BEEN INITIALIZED
               * BITS 0+1 = ROTATION, 2= O/P TO HP, 3+4 = TRANSLATION, 5= I/P FROM HP,
               * 6+7 ARE NOT PRESENTLY USED.
               * BITS 0-4 ARE ACTIVE ON THE POSITIVE TRANSITION, BIT 2
C28C 39        DELAY RTS            ;5
               * R8D (ROTATE FOR 8 DETECTORS)
               * PRODUCES 25.7 DEGREES ROTATION
               * ACCEL.+ DECEL. = 400 STEPS
               * A BIT PATTERN IS NOT REQUIRED FOR 25.7 DEG.
C28D CE 143C R8D    LDX    #5180    ; 5180 STEPS = 25.7 DEGREES
C290 DF 25          STX    M+37
C292 CE 12AC        LDX    #4780    ; 4780*M+12
C295 DF 2A          STX    M+42
C297 BD C000        JSR    ASD
C29A 39             RTS
               * START SENDS A SIGNAL TO THE HP
               * AND WAITS FOR A SIGNAL IN RETURN
C29B C6 1B     START LDA B  #$1B    ;TELL HP SCAN IS READY TO START & SET
               *                     ROT. & TRANSL. BITS HIGH
C29D F7 8004        STA B  ORA
C2A0 BD C28C        JSR    DELAY
C2A3 C6 1F          LDA B  #$1F
```

```
00943 C2A5 F7 8004          STA B  ORA
00944 C2A8 F6 8004 J3       LDA B  ORA
00945 C2AB C4 20            AND B  #$20
00946 C2AD 27 F9            BEQ    J3       ; WAIT FOR HP SIGNAL
00947 C2AF 39               RTS
00950                 * TRANSLATION - A SIGNAL IS SENT TO THE HP
00951                 * EVERY M+50 MOTOR STEPS.  M+51,M+52 AND M+53
00952                 * DETERMINE THE DELAY BETWEEN MOTOR STEPS
00953                 * DELAY BETWEEN STEPS
00954                 * =M+52((M+53*12)+8) + 42 +M+51*12 +56
00955 C2B0 C6 01     TRANSL LDA B  #01
00956 C2B2 D7 2E            STA B  M+46
00957 C2B4 08               INX             ; PULSES IN INX = # OF PULSES FOR TRANSLATION
00958 C2B5 96 34     LOOP6  LDA A  M+52     ;PRODUCES DESIRED DELAY BETWEEN PULSES  3 CYCLES
00959 C2B7 7A 002E          DEC    M+46     ;6
00960 C2BA 26 17            BNE    N2       ;4
00962 C2BC C6 1B            LDA B  #$1B     ;2) SEND SIGNAL TO HP
00963 C2BE F7 8004          STA B  ORA      ;5)
00964 C2C1 BD C28C          JSR    DELAY    ;9+5)
00965 C2C4 C6 1F            LDA B  #$1F     ;2) 42
00966 C2C6 F7 8004          STA B  ORA      ;5)
00967 C2C9 F7 8004          STA B  ORA      ;5)
00968 C2CC D6 32            LDA B  M+50     ;3) HP SIGNAL EVERY M+50 STEPS
00969 C2CE D7 2E            STA B  M+46     ;4)
00970 C2D0 7E C2E8          JMP    J1       ;3)
00971                 *
00972 C2D3 78 0029 N2       ASL    M+41     ;6) TIME TO J1 IS THE SAME EITHER WAY
00973 C2D6 78 0029          ASL    M+41     ;6)
00974 C2D9 78 0029          ASL    M+41     ;6)
00975 C2DC 78 0029          ASL    M+41     ;6) 42
00976 C2DF 78 0029          ASL    M+41     ;6)
00977 C2E2 78 0029          ASL    M+41     ;6)
00978 C2E5 78 0029          ASL    M+41     ;6)
00979 C2E8 09        J1     DEX             ;4
00980 C2E9 27 23            BEQ    J2       ;4
00981 C2EB D6 33            LDA B  M+51     ;3
00982 C2ED 78 0029 J4       ASL    M+41     ;6 )
00983 C2F0 5A               DEC B           ;2 )B*12=M+51*12
00984 C2F1 26 FA            BNE    J4       ;4 )
00985 C2F3 D6 2C            LDA B  M+44     ;3
00986 C2F5 F7 8004          STA B  ORA      ;5
00987 C2F8 BD C28C          JSR    DELAY    ;9+5
00988 C2FB C6 1F            LDA B  #$1F     ;2
00989 C2FD F7 8004          STA B  ORA      ;5
00990 C300 D6 35     LOOP5  LDA B  M+53     ;2
00991 C302 78 0029 J6       ASL    M+41     ;6 )
```

```
00992 C305 5A               DEC B              ;2 ) M+53*12
00993 C306 26 FA            BNE    J6          ;4 )
00994 C308 4A               DEC A              ;2
00995 C309 26 F5            BNE    LOOP5       ;4
00996 C30B 7E C2B5          JMP    LOOP6       ;3
00997 C30E 39          J2   RTS
01000                       END
```

Symbol Table

```
ORA   8004  DDRA   8004  CRA  8005  M      0000  ST     602C  JMPE   605C  ST1    60BC  K5     60EA
MC1   61A6  JMPE2  61B5  ST3  620C  JMPE3  6241  ST4    62AC  JMPE4  62E1  ST5    634C  JMPE5  6381
C1    C036  C2     C039  C3   C03C  C4     C03F  C5     C042  C6     C045  C7     C048  C8     C04B
ASD1  C059  C10    C066  K1   C069  C11    C070  K2     C073  P1     C077  P4     C085  ASD2   C08F
ASD3  C0A7  ASD4   C0C1  B1   C0E7  B2     C0EC  B3     C0F1  B4     C0F6  B5     C0FB  B6     C100
B9    C10F  B10    C114  B11  C119  B12    C11E  J      C120  DEL    C125  D1     C127  Z3     C151
Z6    C15A  Z2     C15D  F    C163  INIT   C194  ACCEL  C1F5  DECEL  C208  DEC32  C242  DEC5   C25D
R8D   C28D  START  C29B  J3   C2AB  TRANSL C2B0  LOOP6  C2B5  N2     C2D3  J1     C2E8  J4     C2ED
J2    C30E  JMPE1  6107  ST2  6179  ASD    C000  ASD0   C009  C9     C04E  J5     C051  C12    C09C
P6    C09F  B7     C105  B8   C10A  Z4     C154  Z5     C157  PIA    C278  DELAY  C28C
```

*** SOME PROGRAM LINE NUMBERS MAY BE MISSING DUE TO DELETION OF SOME COMMENTS. THIS WAS DONE FOR FORMATTING PURPOSES.

*** APPENDIX D - CONTAINS READ SUBROUTINE ***

Motorola M68SAM Cross-Assembler

```
           8004    ORA     EQU    $8004
           8004    DDRA    EQU    $8004
           8005    CRA     EQU    $8005
           8006    ORB     EQU    $8006
           8006    DDRB    EQU    $8006
           8007    CRB     EQU    $8007
           0000    M       EQU    $0000
                           NAM    READ
      C300                 ORG    $C300
                   * 4 TYPES OF INFORMATION WILL BE
                   * SENT FROM THE CAMAC.
                   * (1) 8 BIT DATA SENT THROUGH A SIDE
                   * BITS B7-BO (HIGH-LOW)
                   * (2) 9 BIT ADDRESS FOR DATA (512 BYTES RAM)
                   * B SIDE BO (HIGH), A SIDE B7-BO (LOW)
                   * (3)9 BIT ADDRESS FOR STARTING PROGRAMS
                   * IN RAM (AS IN (2)) AND
                   * (4) 16 BIT ADDRESS FOR STARTING PROGRAMS IN ROM
                   * PIA A SIDE B7-BO IS SENT TWICE.
                   * THE FIRST BYTE SENT IS THE HIGH
                   * BYTE OF THE ADDRESS.
                   * THE TYPE OF INFORMATION
                   * BEING SENT IS ENCODED IN B SIDE
                   * BITS 1&2.
                   * TYPE 1 =B1=B2=O
                   * TYPE 2 =B1=O, B2=1
                   * TYPE 3 =B1=B2=1
                   * TYPE 4 =B1=1, B2=O
      C300 7F 8005         CLR    CRA        ;SET CRA FOR DDRA
      C303 7F 8007         CLR    CRB        ; SET CRB FOR DDRB
      C306 7F 8004         CLR    DDRA       ; A SIDE I/P'S
      C309 C6 F8           LDA B  #$F8       ; B SIDE BITS O-2 I/P,
      C30B F7 8006         STA B  DDRB       ; BITS 3-7 O/P
      C30E C6 E6           LDA B  #$E6       ; SET CRA
                   * CRA = 1110 0110
                   * B5=1 CA2 = O/P;
                   * B4=B3=O HANDSHAKE MODE, CA2 GOES
                   * LOW ON READ ORA & RETURNS HIGH
                   * WHEN IRQA1 IS SET BY CA1;
                   * B2=1 ORA; B1=1 +VE TRANSITIONS OF
                   * CA1 SET IRQA1; BO=O DISABLES IRQA1
                   * INTERRUPT BY CA1.
      C310 F7 8005         STA B  CRA
      C313 C6 FD           LDA B  #$FD       ; SET CRB
```

```
                      * CRB = 1111 1101
                      * B5=1 CB2= O/P; B4=1 CB2 FOLLOWS B3
                      * AS CHANGED BY WRITE CRB INSTRUCTION;
                      * B3=1 NORMALLY HIGH, ACTIVE HIGH TO LOW;
                      * B2=1 ORB; B1=0 -VE TRANSITION OF CB1 SET
                      * IRQB1; B0=1 ENABLES IRQB MPU INTERRUPT
                      * BY  CB1.
C315 F7 8007                 STA B  CRB
C318 FE C31E                 LDX    READ
C31B FF A000                 STX    $A000     ; VECTORIZE JBUG SO THAT IRQ WILL
C31E B6 8005 READ            LDA A  CRA       ; CAUSE A JUMP TO READ SR
C321 84 80                   AND A  #$80
C323 27 F9                   BEQ    READ      ; CHECK IF MESSAGE
C325 B6 8006                 LDA A  ORB       ; HAS BEEN SENT
C328 B7 01F6                 STA A  M+502
C32B 84 04                   AND A  #04       ; $01F6= SCRATCH LOCATION
C32D 26 10                   BNE    B2
C32F B6 8006                 LDA A  ORB       ; IS B2=0 ??
C332 84 02                   AND A  #02       ; B2=0
C334 26 26                   BNE    B1        ; IS B1=0 ??
C336 B6 8004                 LDA A  ORA
C339 A7 00                   STA A  X         ; B2=0, B1=0
C33B 08                      INX
C33C 7E C31E                 JMP    READ
C33F B6 8006 B2              LDA A  ORB
C342 84 01                   AND A  #01       ; B2=1
C344 B7 01F4                 STA A  M+500     ; MASK BITS SIDE B 1-7
C347 B6 8004                 LDA A  ORA       ;$01F4= SCRATCH LOCATION
C34A B7 01F5                 STA A  M+501
C34D FE 01F4                 LDX    M+500     ; $01F5= SCRATCH LOCATION
C350 F6 01F6                 LDA B  M+502
C353 C4 02                   AND B  #02
C355 26 03                   BNE    B91       ; IS B1=0 ??
C357 7E C31E                 JMP    READ
C35A 6E 00   B91             JMP    X         ; B2=1, B1=0
C35C B6 8004 B1              LDA A  ORA       ; B2=1, B1=1
C35F B7 01F4                 STA A  M+500     ; B2=0, B1=1
C362 B6 8005 REA             LDA A  CRA
C365 84 80                   AND A  #$80
C367 27 F9                   BEQ    REA
C369 B6 8004                 LDA A  ORA
C36C B7 01F5                 STA A  M+501
C36F FE 01F4                 LDX    M+500
C372 6E 00                   JMP    X
                             END
```

Symbol Table

ORA	8004	DDRA	8004	CRA	8005	ORB	8006	DDRB	8006	CRB	8007	M	0000
READ	C31E	B2	C33F	B91	C35A	B1	C35C	REA	C362				

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