Tomographic Imaging of Matter using Primary and Secondary X- and Gamma-Radiation.

By

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A thesis submitted to the Faculty of Science of the University of Surrey for the degree of Doctor of Philosophy.
To my family.
"Writing a book is an adventure: it begins as an amusement, then it becomes a mistress, then a master, and finally a tyrant."

Sir Winston Churchill.
Acknowledgments.

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Abstract.

Gamma rays may interact with matter by a variety of processes, many of which give rise to secondary radiations. This thesis examines the possibility of performing tomographic imaging by means of these secondary photons using low-cost apparatus.

The techniques are compared with each other and with transmission tomography, which plays such an important role in modern diagnostic imaging. The progress of industrial tomography is reviewed as are techniques of investigation using gamma ray scattering in both industry and medicine. Some new applications of a simple gamma ray CT scanner have been performed.

A method of determining the spatial distribution of pure beta emitters in matter by performing tomographic imaging using the bremsstrahlung radiation produced by the beta particles has been demonstrated. This technique has been shown to permit imaging at depths in material greatly exceeding the range of beta particles in matter.

All the imaging techniques using secondary radiation have displayed two principal limitations: long scanning times and poor quantitative accuracy. The low scanning rate results from the small number of secondary photons that are detected. The major contributing factors to poor accuracy are attenuation and the noise produced by unwanted in-scattering.

The possible applications for secondary photon imaging have been briefly outlined and some suggestions for future work are included. Although techniques based upon imaging using secondary radiation will not be able to compete with transmission CT in the vast majority of applications, they may prove valuable in a range of specialised fields.
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Chapter 1.

Introduction.

The potential of ionizing radiations for imaging was recognized soon after their discovery by Roentgen in 1895. The ability of X-rays to penetrate matter was soon appreciated and their use revolutionized diagnostic medicine. Since those early years, a very wide range of applications for sources of radiation have been found and today their employment is invaluable in many fields.

The primary reasons for the usefulness of ionizing radiations are their ability to pass through material and the ease by which they may be detected. For most inanimate objects, large doses of radiation may be employed without adverse effect and thus techniques involving radiation are non-destructive. In medical applications the benefits obtained from measurements made using radiation generally greatly outweigh the risks to health associated with the dose to the patient, and the employment of radiation techniques has proved vital to many people.

Despite the amount of adverse publicity that ionizing radiations have received recently, and the appearance of new non-destructive techniques (such as ultrasound and NMR) the versatility of radiation based measurements will ensure their continued employment in the future.

The work described in this thesis is primarily concerned with the use of ionizing electromagnetic radiations (X- and γ-rays) to image objects of industrial interest. The simplest form of imaging using radiation is radiography, in which a 'shadow picture' is formed by the passage of a beam of radiation through the object onto an X-ray sensitive film. This method is very widely used in both industry and medicine as a result of its speed, simplicity and low cost. Radiographs are, however, two dimensional representations of three dimensional objects. As a result they suffer from a loss of depth information.
along the beam path resulting from the superposition of images formed by successive plane sections of the material under examination.

Over the past two decades increasing attention has been paid to other forms of imaging loosely described as 'tomography', which involve the formation of an image of a single plane or slice through the object. These techniques generally employ collimated detectors to provide the required degree of spatial resolution and frequently utilise computers to reconstruct a two dimensional image from a set of one dimensional measurements.

The technique of computerized tomography (CT) has had a major impact upon diagnostic medicine and has also been used effectively in a range of industrial applications. CT techniques may be broadly divided into two classes, transmission tomography and emission tomography. In transmission tomography a beam of X- or γ-rays is detected after passing through the object and the imaged parameter is the linear attenuation coefficient of the object. In emission tomography a radioactive substance is generally introduced to the object of interest and the resulting image maps the spatial distribution of the radionuclide.

Transmission CT images provide useful information about macroscopic objects because the linear attenuation coefficient, generally denoted by μ, is determined principally by the photon energy and physical parameters of density and atomic number of the material. The linear attenuation coefficient is, in fact, the sum of the partial cross-sections due to a range of interaction processes, as shown in fig. 1.1. I₀ represents the initial beam intensity and Iₓ the transmitted intensity, and the various σ terms represent the respective cross-sections for the attenuation processes, Compton scattering, pair production, Rayleigh scattering, the photoelectric effect and at high energies various nuclear interaction products.

As shown in fig 1.1, each of the individual interaction processes may give rise to secondary radiations, which are, in principle, detectable. Since the dependence of the partial cross-section term on density and elemental composition
varies substantially from interaction to interaction (see chapter 2), measurements of the secondary radiations from an object may be more sensitive than measurements of the transmitted intensity.

Since visual information is generally easy to interpret, where spatial variations in physical parameters of interest are of primary concern it is natural to use the detected radiation to form an image. A straightforward way of doing this is to perform spatially resolved measurements using a detector system capable of rejecting events resulting from interactions other than those of interest. Such an image may essentially constitute a secondary photon radiograph with resulting loss of information in one dimension; the use of focused collimation, however, results in a tomographic image being formed as a result of interactions within a single plane of interest. The technique of computerized tomography, described in chapter 3, provides an alternative method of imaging. The availability of comparatively powerful low-cost personal computers has made CT economically favourable, and CT techniques (which do not require the complicated collimation systems of focal-plane methods) generally demonstrate higher counting rates for a given detector size and source strength than non-reconstructive tomography.

Transmission CT (TCT) is particularly fast technique, multi-beam systems being capable of imaging in times of minutes with radioisotope sources and seconds with X-ray tubes. The main reason for the speed of TCT is that the utilization of emitted photons is very high, typically a few percent of the incident count rate. In comparison, other methods detect only the small fraction of photons which, after undergoing the appropriate interaction, are emitted into the detector field of view. As a result TCT and conventional radiography are the only viable techniques for imaging rapidly changing systems or for routine monitoring of large numbers of samples.

There exist, however, a host of applications in which high speed imaging is not required, and as a result it is of value to consider imaging
techniques which are based upon detection of secondary radiation.

Chapter 2 describes these interaction processes in more detail, and chapter 3 describes the basic principles of reconstructive tomography. An overview of the industrial applications of transmission CT is given in chapter 4 which includes several practical applications of the technique using a simple low-cost laboratory apparatus.

In chapter 5 the uses of Compton scattering in industrial gauging and diagnostic medicine are described and the techniques of imaging using Compton scattered radiation are outlined. The results of experimental reconstructive Compton scatter imaging are shown and the chapter closes with a brief review of the prospect of imaging using coherent (Rayleigh) scattered photons.

Chapter 6 deals with X-ray fluorescence tomography which is an elemental sensitive imaging technique. Some experimental techniques are demonstrated and some drawbacks to the method are outlined. Chapter 7 discusses bremsstrahlung tomography for the imaging of pure beta emitters in matter and briefly describes the practical limitations of the method. The final chapter sums up the various techniques and discusses their rival merits and concludes with suggestions for further work.
Figure 1.1: Secondary Radiation Emitted from a Gamma Ray Beam.
Chapter 2

The Interactions Between Gamma Rays and Matter.

2.1 Introduction.

If an external radiation beam is to be used to obtain information about an object, some interaction between the radiation and the object must occur. The nature of the interaction may involve absorption or scattering of the primary beam, and may also involve the production of secondary radiation.

This chapter will outline the four basic modes of photon interaction which are of practical importance in analysing matter on a macroscopic scale; these are Rayleigh scattering, photoelectric absorption, Compton scattering and pair production. Other interactions can occur with the atomic nucleus and nuclear scattering processes are briefly described. The cross-section for nuclear interactions is, however, generally prohibitively small for their useful employment for the interrogation of matter.

The attenuation of gamma radiation is of particular importance in the interrogation of matter, and a section has been given to studying the definitions of attenuation coefficients. The contributions of the various interaction processes to the attenuation coefficient are briefly discussed.
2.2 Rayleigh Scattering.

The process of coherent (elastic), or Rayleigh scattering consists of the scattering of photons by bound electrons without the excitation or ionization of the parent atom [Dyson 1973].

The basic principle of Rayleigh scattering is well explained in terms of the classical theory of the electron. When an electromagnetic wave is incident upon an electron, the electron is set into vibration at the driving frequency. The vibrating charge reradiates the electromagnetic radiation with its original frequency, the angular distribution of radiation is determined by the behaviour of the electron as an electric dipole. This process is referred to as Thomson scattering.

In the Thomson model the electron is treated as essentially free, and momentum conservation during the process is neglected. This classical explanation is valid, however, as the electron is loosely bound to the atom, allowing the momentum of the incident photon to be absorbed by the atom as a whole. The differential cross-section for this process can be shown to be (eg. Dyson 1981):

$$\frac{d\sigma_{\text{Th}}}{d\Omega} = \frac{r_e^2 (1 + \cos^2 \theta)}{2}$$

where \(r_e\) is the classical electron radius \((e^2/m_e c^2)\) and \(\theta\) is the scattering angle.

When the scattering occurs as a result of interactions between photons and atoms rather than single electrons, account must be taken of the coherent nature of the scattering process. The scattered radiation from all the electrons within the atom is in antiphase with the incident radiation, causing the scattered contributions to be coherent. The effect of this
coherence may be taken into account by adding an atomic scattering 'form factor' \( F(q,Z) \), the square of which represents the probability that the recoil momentum is transferred to the atom without any energy loss. The differential Rayleigh scattering cross-section may thus be written:

\[
\frac{d\sigma_{\text{ray}}}{d\Omega} = \frac{r_0^2(1+\cos^2\theta).|F(q,Z)|^2}{2}
\]

(2.2.2).

The form factor \( F(q,Z) \) may be defined as:

\[
F(q) = Z \int e^{i\mathbf{q} \cdot \mathbf{r}} \rho(r) d^3r
\]

(2.2.3)

where \( q = \lambda^{-1}\sin(\theta/2) \) for a photon of wavelength \( \lambda \), and \( \rho(r) \) is the electron density distribution.

The value of \( F(q,Z) \) falls with both increasing energy and scattering angle due to the reduced probability of momentum transfer occurring without energy absorption by the atom. As the value of the atomic number, \( Z \), increases the Rayleigh scattering cross-section increases due to the increasing binding energy of the inner electrons (\( \sim Z^2 \)) enabling a greater number of electrons to participate in the coherent scattering process. The two main characteristics of coherently scattered radiation are:

(I) that the scattered photons are of the same energy as the incident beam, and

(ii) the scattering is strongly forward peaked.

These two characteristics of Rayleigh scattering result in it being of minor importance as gamma ray attenuation process. However, it is the coherent scattering from planes of atoms which give rise to the well-known crystal diffraction patterns.
2.3 The Photoelectric Effect.

The photoelectric effect is essentially an absorption process in which the incoming gamma ray interacts with a bound electron. The gamma ray energy is completely transferred to the electron which is then ejected from the atom with an energy $E_e$, given by:

$$E_e = E_\gamma - E_b$$  \hspace{1cm} (2.3.1)

where $E_\gamma$ is the gamma ray energy and $E_b$ is the binding energy of the ejected electron.

The value of photoelectric cross-section is characterized by discontinuities at low energies due to effects of absorption edges, showing sudden increases in value as the photon energy exceeds the binding energy of the electron shells of the absorber atom.

For a given electron shell, the photoelectric absorption cross-section is a maximum when $E_\gamma$ is just greater than the electron binding energy. Thus for gamma radiation above the K-edge, the most prominent photoelectric interaction is with the K-shell electrons. The photoelectric effect is strongly dependent upon atomic number, and is the dominant form of interaction at low energies, becoming progressively less important above about 100 keV.

Exact theoretical treatment of the photoelectric cross-section $\sigma_p$ has proved difficult, however a number of formulae have been employed to describe the photoelectric process using simplifying assumptions. The Born approximation has been used (Heitler 1954) to give an approximate equation for photons with energy well above the K-absorption edge:
\[
\sigma_{\text{BA}} = 4\pi^2 \frac{Z^2 \alpha^2}{n^2} \frac{(mc)^2}{E_{\gamma}^2} \sigma_{\text{T}}. \tag{2.3.2}
\]

where \(\sigma_{\text{BA}}\) is the Born approximation photoelectric cross-section for \(s\)-state electrons, \(\alpha\) is the fine structure constant, \(n\) the principal quantum number, \(Z\) the atomic number and \(\sigma_{\text{T}}\) is the Thomson cross-section.

This formula assumes that the electron is ejected by a plane wave and neglects screening and relativistic effects. The formula displays the strong \(Z\) and \(E_\gamma\) dependence exhibited by the photoelectric process, but is otherwise a rather poor approximation, due to the inconsistencies between the assumptions made in the Born approximation and the the basic theories of photon-electron interactions.

A more accurate semi-theoretical model of the photoelectric effect has been described by Jackson and Hawkes [1981]. This treatment has overcome some of the drawbacks associated with the Born approximation method, and indicates that it is not possible to separately factorize the terms in \(Z\) and \(E_\gamma\).

2.4 Compton Scattering.

The Compton scattering process is essentially a collision between a gamma photon and a free electron, in which energy and momentum are transferred to the electron, giving rise to a scattered photon with reduced energy. By simple energy and momentum conservation principles it can be shown that the energy \(E'_{\gamma}\) of the scattered photon is given by:

\[
E'_{\gamma} = \frac{E_\gamma}{1 + (E_\gamma/m_c^2)(1 - \cos \theta)} \tag{2.4.1}
\]

The free electron cannot absorb all the energy of the gamma photon due
to the law of conservation of momentum, and the maximum energy transfer occurs when the gamma photon is backscattered through 180°. For small scattering angles, very little energy is transferred and $E'_{\gamma} \approx E_{\gamma}$.

The differential cross-section for Compton scattering was derived by Klein and Nishina [1929] using the Dirac equation applied to a free electron and is given by:

$$
\frac{d\sigma_{\text{KN}}}{d\Omega} = \frac{r_{e}^{2}}{2} \left[ \frac{1}{1+\varepsilon(1-\cos \theta)^{2}} \right] \left[ (1+\cos^{2} \theta) + \frac{\varepsilon^{2}(1-\cos^{2} \theta)}{1+\varepsilon(1-\cos \theta)^{2}} \right] \quad (2.4.2)
$$

where $r_{e}$ is the classical electron radius, $\varepsilon^{2}/m_{e}c^{2}$ and $\varepsilon = E_{\gamma}/m_{e}c^{2}$.

This formula shows the cross-section falling with increasing scattering angle, the decrease in cross-section with angle becoming more pronounced as $E_{\gamma}$ increases. This leads to the Compton scattered radiation being strongly forward peaked at high photon energies.

As $E_{\gamma}$ falls, equation (2.4.2) reduces to the classical differential Thomson cross-section as given in equation (2.3.2).

The total cross-section for incoherent scattering for an electron is given by integrating equation (2.4.2) to give:

$$
\sigma_{\text{KN}} = 2r_{e}^{2} \left[ \frac{1}{\varepsilon^{2}} \left( 2\varepsilon(1+\varepsilon) - \ln(1+2\varepsilon) \right) + \frac{\ln(1+2\varepsilon) - \ln(1+2\varepsilon)^{2}}{2\varepsilon} \right] \quad (2.4.3).
$$

Hubbell [1969] has tabulated values of the Klein-Nishina cross-section over the energy range 10 keV to 100 GeV.

If all the electrons in an atom of atomic number $Z$ are considered free, the incoherent scattering cross-section for the atom is given by:

$$
\sigma_{\text{inc}} = Z \sigma_{\text{KN}}, \quad (2.4.4).
$$

In practice allowance must be made for the bound nature of atomic electrons. The effect of electrons being bound is to reduce the
differential cross-section for incoherent scattering to below that predicted by the Klein-Nishina formula. At small scattering angles or low photon energies, the electron may recoil with an energy less than its binding energy, and hence the electron is not ejected from the atom, remaining in a bound state. The Compton scattering from bound electrons is an inelastic process as there is some kinetic energy lost due to changes in the internal energy of the atom.

The effect of binding is normally taken into account by the inclusion of the incoherent scattering factor $S(q, Z)$. The limiting behaviour of this factor is given by:

$$S(q) \rightarrow 0 \quad \text{as} \quad q \rightarrow 0$$

$$S(q) \rightarrow Z \quad \text{as} \quad q \rightarrow \infty$$

The differential Compton scattering cross-section for an atom thus becomes:

$$\frac{d\sigma}{d\Omega}_{\text{Com}} = \frac{d\sigma}{d\Omega}_{\text{K,N}} S(q, Z).$$

Compton scattering is the most probable form of gamma ray interaction over a wide range of energies up to several MeV.

2.5 Pair Production.

The pair production process involves the creation of a positron and electron within the intense Coulomb field near an atomic nucleus. The total energy of the positron-electron pair is equal to that of the parent photon.

Pair production can only occur when the photon energy exceeds $2m_e c^2$. 
(i.e. 1.022 MeV) which is the rest energy of the two particles. The full description of the process requires the quantum field theory developed by Dirac, which considers the process as involving the excitation of electron from a negative energy state by the photon. The 'hole' produced in the negative energy state by the loss of the electron is observed as a positron. To conserve both energy and momentum during the pair production process, the presence of the nuclear Coulomb field is required. No change in the nucleus or its surrounding electrons occurs during the process, except for the absorption of the momentum of the photon.

Pair production plays an important role in gamma ray attenuation at energies of several MeV, the cross-section increasing with photon energy above threshold and approximately with the square of the atomic number.

2.6 Nuclear Interactions.

Although unimportant in gamma ray attenuation, nuclear scattering is of interest in the understanding of the interactions of radiation with matter, and may give rise to information about the nuclear structure of matter if the scattered radiation is detected. The interactions may be divided into nuclear Thomson scattering and nuclear resonant scattering.

Nuclear Thomson scattering is the name given to the coherent scattering which occurs due to the interaction between the nucleus (which may be considered as a point charge) and the electric field associated with an incident gamma ray. The cross-section is the same as that shown in equation (2.2.1), except with the nuclear mass replacing the electron mass in \( r_0 \). As a result of the large difference in nuclear and electron masses, the overall interaction probability is very low, nevertheless, the
scattered radiation is strongly forward peaked and as a result may be
detected at small angles [Davy 1953].

A second nuclear scattering process is known as nuclear resonant
scattering and is dependent upon the properties of nuclear excited states.
The basic principle of the interaction is the absorption of a gamma ray by
a nucleon, the nucleon then being raised to an excited state. The nucleus
may then emit a gamma ray of the original energy as it deexcites. The
cross-section for the process is exceedingly small as the nuclear states in
question have very narrow widths. At moderate photon energies, nuclear
resonant scattering is only generally observable when the nucleus is
irradiated from a source which is decaying (by beta emission or electron
capture) to the excited state of interest. At higher energies nuclear
resonant scattering becomes more probable due to the greater width of gamma
ray lines resulting from energetic reactions with high recoils, giving
cross-sections of a few millibarns [Reibel 1960].

The process of nuclear resonant scattering is of principal practical
importance in the field of Mössbauer spectroscopy, in which the nucleus is
bound into a crystal lattice, thereby reducing recoil effects effectively
to zero. In such situations with low temperatures and low gamma ray
energies, the interaction probability between a gamma ray emitted by a
nucleus within the lattice and another nucleus of the same species
contained within the lattice is very high. The technique of Mössbauer
spectroscopy has been of considerable interest in the field of solid state
physics, although the technique lies outside the scope of this work.

Above the nuclear binding energy (~8 MeV), high energy photons can
participate in (γ,n) and (γ,p) reactions. The generation of energetic
particles by this means along the beam path of the incident photons could,
in principle, be used for mapping the isotopic constitution of objects in
favourable cases. This is the nuclear analogue of XRF tomography.
2.7 The Attenuation of Gamma Rays in Matter.

2.7.1 Introduction.

When passing through matter, gamma rays may interact via a range of scattering and absorption processes. The removal of photons from a beam of gamma rays is commonly described in terms of an attenuation coefficient. Measurements of attenuation coefficients may be employed as a method of analysing materials because the value of the attenuation coefficient is dependent upon the density and elemental composition of an object.

2.7.2 Definitions of Attenuation Coefficients.

The linear attenuation coefficient, \( \mu \), is defined as the probability of the photon interacting per unit length in a material and may be related to the total interaction cross-section, \( \sigma_{\text{tot}} \) by:

\[
\mu = N \sigma_{\text{tot}}
\]  

2.7.21.

The number of atoms per unit volume, \( N \), is given by:

\[
N = N_A \rho / \Lambda
\]  

2.7.22

where \( N_A \) is Avogadro's number and \( \rho \) the mass density of the material. The mean free path \( \lambda \) is equal to the reciprocal of the attenuation coefficient.

The attenuation coefficient may be determined by measurement of the relative intensity of a narrow beam of gamma photons passing through an absorber.
Consider a collimated source of monoenergetic gamma rays, \( S \), producing a narrow beam of photons of intensity \( I_o \) as in fig. 2.7.2a. If \( x \) denotes the thickness of the absorber in the direction of the beam then the number of photons, \(-dl\) removed in a small element of thickness \( dx \) is given by:

\[
dl = -\mu I dx
\]

Integrating 2.7.23 over the thickness of the absorber gives:

\[
I = I_o e^{-\mu x}.
\]

It must be noted that this formula is only valid for the case of a narrow beam of gamma rays, in which any scattered photon is invariably lost from the beam. In the broad beam case, allowance must be made for the secondary photons produced by the interaction processes contributing to the total detected flux, see fig 2.7.2b. This is normally described by a build-up factor, \( B \), giving:

\[
I = I_o Be^{-\mu x}.
\]

where \( B \) is a function of \( E, \rho, Z, \) and geometrical details.

It is often convenient to use the mass attenuation coefficient, \( \mu/\rho \) in place of \( \mu \). This enables 'mass thicknesses' (i.e. mass per unit area) of various materials to be used to study gamma ray attenuations. Conversion from mass attenuation coefficient to linear attenuation coefficient is achieved by multiplication of the mass attenuation coefficient by the mass density of the material in question.
Gamma Ray Attenuation

Figure 2.7.2a: Narrow Beam Geometry.

\[ I = I_0 \exp(-\mu x) \]

Figure 2.7.2b: Broad Beam Geometry.

\[ I = B I_0 \exp(-\mu x) \]

S-Source
D-Detector
2.7.3 The Mixture Rule.

When the absorbing material consists of compound or a mixture of substances, then the mass attenuation coefficient is given by a weighted sum of the attenuation coefficients of the constituent elements:

\[(\mu/\rho)_{\text{mix}} = \Sigma w_i (\mu/\rho)_i\]  

where \((\mu/\rho)_i\) is the mass attenuation coefficient of the ith element and \(w_i\) is its proportion by weight.

It is assumed in this formula that the interaction cross-sections for gamma rays are independent of the chemical bonding of the absorbing atoms. Hawkes [1982] has suggested that this assumption gives rise to errors of less than 2% for energies above 10 keV, which are more than 1 keV away from an absorption edge.

2.7.4 Factors Affecting The Value of the Attenuation Coefficient.

As outlined earlier in this chapter, at energies below the nuclear binding energy, gamma rays interact with matter via four main physical processes: the photoelectric effect, Rayleigh scattering, Compton scattering and pair production. The contributions made to the attenuation coefficient by each of these processes is a function of both photon energy and the atomic number of the element involved. In general the attenuation coefficient may be written:

\[\mu(Z,E_p) = N(\sigma_{\text{pe}} + \sigma_{\text{ray}} + \sigma_{\text{cp}} + Z\sigma_{\text{cdw}})\]
The Compton scattering component is multiplied by the atomic number \( Z \) because the Compton effect occurs with individual electrons rather than the whole atom.

The variation with \( Z \) and \( E_\gamma \) of the interaction cross-sections for the processes outlined above leads to substantial changes in both the value of the attenuation coefficient and the relative importance of the interaction processes if one considers a range of elements. Figure 2.7.4a shows the variations in the interaction cross-section for copper as a function of energy [Hubbell 1970], and figure 2.7.4b displays the comparative contributions of the photoelectric effect, Compton scattering and pair-production with varying \( Z \) and \( E_\gamma \) [Evans 1955].

Understanding of the relative importance of the gamma ray interaction processes is of crucial importance in optimising a system using gamma radiation to interrogate matter. From fig. 2.7.4b it will be noticed that at low energies and high atomic numbers the photoelectric effect is dominant. At low energies, therefore, the attenuation coefficient is strongly dependent on the elemental composition of the absorber because of the strong variation of \( \sigma_{\text{photo}} \) with \( Z \). At energies above a hundred keV, the Compton scattering process becomes dominant, and as a result the attenuation becomes strongly dependent on the electron density of the material. At very high energies the pair-production becomes dominant and the atomic number of the absorber again becomes important.
Gamma ray attenuation processes for copper (Hubbell 1970)
Relative Importance of the Gamma-ray Interaction Processes

- Photoelectric effect dominant
- Pair-production dominant
- Compton effect dominant
Chapter 3.

The Theory of Computerized Tomography.

3.1 Introduction To Computerized Tomography.

Almost immediately after their discovery by Roentgen in 1895, X-rays were being employed as a method of imaging the human body. Since that time, X-radiography has proved one of the most important tools in diagnostic medicine. Applications have also been found in many industrial fields as a means of non-destructive testing.

Over the past few decades many improvements have been made in techniques of obtaining data using radiography (Cullinan 1972). The use of image intensifiers, higher intensity tubes, improved collimators and contrast media have increased both the speed and versatility of medical radiography.

Despite the many recent improvements, the basic principle of radiography displays several limitations as an imaging process. A conventional radiograph is essentially a superimposition of a series of shadows cast by the various structures within the object. Some features of the object's internal structure may be revealed if they have absorption coefficients differing greatly from the surrounding material. However, the technique remains a two-dimensional representation of a three-dimensional object, in which depth information is absent and much detail is obscured by the confusion of overlapping material. To obtain greater detail from X-ray transmission measurements, it is necessary to reduce the ambiguity in the radiographs caused by the interference between the structures at different depths within the object.
The process used to reduce the information loss is called tomography, a term loosely derived from the Greek word 'tom' meaning slice. The first tomographic measurements were performed by Bocage (1921), this method utilised an exposure performed using a moving X-ray source and recording film [Herman 1980]. This focal plane tomography works on the basis that only one layer within the object is not blurred out as a result of the motion. The effect of this technique is to produce a sharp image of the plane of interest on which is superimposed the blurred background from the rest of the object. This method has proved valuable in medical radiography, although has not found employment in industrial fields [Gilboy 1982]. Although the detail of the plane of interest is not totally obscured by other structures, the image may be seriously degraded by the blurring effects of the other planes.

The difficulties involved in conventional tomography have been largely overcome by the use of reconstructive or computerized tomography (CT). The developments giving rise to modern computerized tomography illustrate the absence of cross-fertilisation of ideas which sometimes occurs in the scientific world. The mathematical basis for tomographic reconstruction was first developed by Radon (1917), who proved that it was mathematically possible to reconstruct an image of an object from a series of projections. However, the first practical technique for reconstructing a 2-dimensional image from a series of 1-dimensional measurements was made in astrophysics by Bracewell (1956). Bracewell independently developed a method based upon Fourier analysis which was later shown to be equivalent to Radon's technique. The reconstruction of an image from a series of views has also been achieved in electron microscopy [De Rosier 1968] and optics [Rowley 1969]. The first worker to use tomographic reconstruction in association with the detection of transmitted ionizing radiation was Cormack (1963). He utilised a $^{60}$Co source in association with a Geiger-Müller detector to
image an aluminium and wood phantom, positioned by hand within the fixed radiation beam.

It is, however, largely to Hounsfield's work at E.M.I. [Hounsfield 1968, 1972, 1973] that modern CT imaging owes its current importance as a practical diagnostic tool. The Hounsfield brain scanner incorporated an X-ray source and detector and on-line minicomputer and has revolutionised medical imaging. The importance of the CT technique in diagnostic medicine was recognised by the 1979 Nobel Prize for medicine being awarded to Cormack and Hounsfield.

3.2. The Theory of Computerized Tomography.

The basic configuration of the tomographic scanners discussed here consists of a source of radiation (either a radioisotope or an X-ray generator), a detector and a scanning bed on which the test object is situated. In most medical applications the source and detector are mounted on a gantry which rotates around the patient. For industrial applications with small samples, however, it is more convenient to move the object whilst keeping the source-detector apparatus stationary.

The most simple 'first generation' scanners use a single radiation source emitting a single tightly collimated beam aligned with a detector. To increase the scanning speed, later scanners have employed multiple or fan beam sources and detector arrays [Seeram 1982], enabling scan times to be reduced by two orders of magnitude.

To understand the basic principles of image reconstruction from projections, some initial definitions are required. Consider figure 3.21; in this diagram an object with a density function $f(x,y)$ is contained within a region of interest $R$, $f(x,y)$ being zero outside this region.
Figure 3.2: Coordinate system for parallel beam geometry.
Consider a ray from the source to the detector, defined by the coordinates $r$ and $\theta$, where $\theta$ is the angle of the ray with respect to the $y$ axis and $r$ is the distance from the origin and:

$$
x = r\cos \theta \\
y = r\sin \theta.
$$

(x.2.1)

If the path length of the ray is $s$, given by:

$$s = x\cos \theta + y\sin \theta$$

(3.2.2)

then the "raysum" $p(r, \theta)$ may be defined by the Radon transform:

$$p(r, \theta) = \int f(x, y) \cdot ds$$

(3.2.3)

For gamma ray transmission tomography $f(x, y)$ represents the linear attenuation coefficient $\mu(x, y)$ of the object, enabling a physical meaning to be given to the quantity $p(r, \theta)$ in terms of the detector response. For a narrow beam of intensity $I_0$, the transmitted beam intensity, $I$, is given by:

$$I = I_0 \exp(-\int \mu(x, y) \cdot ds).$$

(3.2.4)

Thus it can be seen that the detected intensity gives a measurement of the line integral of the attenuation coefficients through the object along the beam direction. Combining (3.2.3) and (3.2.4) we have:

$$p(r, \theta) = -\ln (I/I_0).$$

(3.2.5)
Hence it may be seen that the value of the raysum is proportional to the natural logarithm of the detected counts. In emission tomography, the value of \( p(r, \theta) \) is related to the activity of the radionuclide at the point \((x, y)\) within the sample.

A set of raysums taken at a constant angle, \( \theta \), is called a projection or profile. It is the aim of CT reconstruction to determine the values of \( f(x, y) \) for the points within the image region to a useful degree of accuracy using a finite number of projections and raysums, the calculated values being denoted by \( f(x, y) \).

The tomographic reconstruction process can be considered mathematically as the inversion of the Radon transform (3.2.3). Writing in polar coordinates:

\[
f(r, \theta) = \frac{1}{2\pi} \int_{0}^{\infty} \int_{0}^{2\pi} \frac{p(r', \phi)}{r \cos(\theta - \phi)} \, dr' \, d\phi \quad (3.2.6)
\]

where \( p(r', \phi) \) is the raysum at \( r' \) and \( \phi \).

In theory, the values of \( f(r, \theta) \) can only be determined for an infinite number of raysums. In practice, however, a finite number, \( N \), of raysums are recorded and in consequence a finite number of values of \( f(r, \theta) \) are calculated. If we divide the image region, \( R \), into a \( N \times N \) grid, then it is possible to assign a value of \( f(r, \theta) \) to each of the \( N^2 \) picture elements (pixels). The photon beam has finite dimensions, so each of the calculated values for \( f(r, \theta) \) will be an average over a volume element or voxel.

It is sometimes convenient to approach the reconstruction problem using Fourier theory. The one-dimensional Fourier transform of (3.2.3) may be written:

\[
P_i(R, \theta) = \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} f(x, y) e^{-2\pi i r x} \, dr \, ds. \quad (3.2.7)
\]
\[
\hat{f}(r, \theta) = \mathcal{F}\{f(x,y)\} = F_2(R, \theta)
\]  

(3.2.9)

\(
F_i(R, \theta) \) is the two dimensional Fourier transformation of \( f(x, y) \). Thus the Fourier coefficient \( f(x, y) \) equals the Fourier coefficient of the projection at the same angle, if \( \mathcal{F} \) is the Fourier transform operator then:

\[
f(x, y) = \mathcal{F}^{-1}\{F_1(r, \theta)\; 0 \leq \theta \leq \pi\}.
\]  

(3.2.10)

The principle of inverting the Fourier transform has not been widely employed in CT due to the process being computationally wasteful.

The practical methods of solving \((3.2.5)\) may be divided into analytical and iterative methods, both techniques have been successfully utilised in tomographic reconstruction (Kouris 1982, Brooks 1976). These techniques will be briefly discussed in the following sections.


Intuitively, the most obvious means of estimating the values of \( f(r, \theta) \) is by using the method of back-projection, in which the value of each raysum is distributed evenly across all the points along the ray in the image plane.

The back-projected, \( b(r, \theta) \), image may be expressed as:

\[
b(r, \theta) = \int p(r \cos(\theta - \phi)) d\phi
\]  

(3.3.1)
This process is shown for two projections in figure 3.3. Clearly this method of approximation overestimates the value of \( f(r, \theta) \) in the low density regions of the image; if \( f(r, \theta) \) is a delta function at the origin then for an infinite number of projections, then the values of \( b(r, \theta) \) will be of the form of a \( 1/r \) distribution. If a finite number of projections are employed, then a star-shaped distribution results; this 'star artifact' must be eliminated if good quality images are to be formed.

The most common method of removing the star artifact is by means of a filtering process, in which the projection data is convolved with some function prior to back-projecting. The filtered projections contain negative components, which, when back-projected tend to remove the star artifact, a variety of filters may be used for this purpose [Kouris 1982].

Mathematically this process can be envisaged by writing (3.3.1) as a Fourier transform [Folkard 1983]:

\[
b(r, \theta) = \int \int P_i(R, \phi)e^{2\pi i R R e^{i \phi}} dR d\phi \tag{3.3.2}
\]

where \( P_i(R, \phi) \) is the Fourier transform of (3.2.3).

Taking the 2-dimensional Fourier transform of the back-projected image gives:

\[
B(R, \phi) = \frac{P(R, \phi)}{|R|} \tag{3.3.3}
\]

Hence it may be seen that the back-projected image is the true image with the Fourier amplitudes divided by the spatial frequency, confirming the conceptual argument above, because the back-projection of a delta function at the origin gives a point spread function with a \( 1/r \) distribution. If we choose a filter function of the form:

\[
H(R) = |R|.A_{\theta}(R). \tag{3.3.4}
\]
Two Profiles.

Figure 3.3: Backprojection.
where \(A(R)\) is a slowly varying function applied to the basic filter. \(H(R)\) is the filter modulation transfer function and the Fourier transform of \(H(R)\) is known as the impulse response function, \(h(r)\).

The filtering process consists of multiplying the Fourier transform of the back-projected image by the modulation transfer function of the filter to give the Fourier transform of the filtered image, \(C(R,\phi)\):

\[
C(R) = H(R).B(R,\phi). \tag{3.3.5}
\]

Combining (3.2.9), (3.3.3) and (3.3.4), gives:

\[
C(R) = A_2(R).F(R,\phi) \tag{3.3.6}
\]

thus by taking the transform of both sides:

\[
c(r,\theta) = f(r,\theta) \ast\ast a(r) \tag{3.3.7}
\]

where \(\ast\ast\) represents the operation of the two-dimensional convolution. The term \(a(r)\) represents the point spread function of the technique, giving an indication of the unsharpness of the image.

As the operations of back-projection and filtering are linear it is possible to filter the projection data and use these data for reconstructing the image. The projection data will then contain negative values which will cancel out the undesirable positive components in the back-projected image. The filtered projection data, \(p'(r,\theta)\), are formed by convolving \(p(r,\theta)\) with the filter function \(h(r)\) giving:

\[
p'(r,\theta) = p(r,\theta) \ast h(r) \tag{3.3.8}
\]
Back-projecting the filtered projections gives:

\[ b'(r, \theta) = \int \int p(r, \theta) h(r \cos(\theta - \phi) - x) \, d\phi \, dx. \]  

(3.3.9)

The process of convolution or filtered back-projection is one of the most popular methods of reconstruction as the technique is both fast and easy to implement.

3.3.1 The Choice of Filter Function.

The choice of filter function is important in order to obtain the optimum performance of the reconstruction technique. The optimum filter will depend on the purpose of the imaging experiment, and to a certain extent the preference of the experimenter.

Equation (3.3.4) indicates that the best filter to use is \(|R|\), but this filter tends to amplify high frequency noise, so it is not used in practice. To overcome this difficulty it is necessary to impose some form frequency cutoff at \(R'\), however, by reducing \(|R|\) at high frequencies the accuracy of the reconstruction is impaired, producing ringing at the edges of the reconstruction [Foster 1981]. Thus the selection of the filter function is a compromise between noise reduction and image accuracy.

Three commonly used filter functions are those of Bracewell and Riddle, Shepp and Logan and Hanning. The Bracewell filter [Bracewell 1967] is defined as:

\[ H(R) = \begin{cases} \frac{|R|}{2R'} & \text{if } |R| < R' \\ 1 & \text{if } |R| = R' \\ 0 & \text{if } |R| > R' \end{cases} \]  

(3.3.1.1)

The limiting spatial frequency is usually selected as:
\[ R' = 1/2s \quad (3.3.1.2) \]

where \( s \) is the step length in parallel beam tomography, as sampling at a spacing \( s \) irretrievably loses all information on frequencies above \( 1/2s \) [Foster 1981]. The modulation transfer function and impulse response function obtained using the Bracewell filter are shown in figure 3.3.1a and 3.3.1b. The Bracewell filter gives good reconstruction in the absence of noise but its discontinuous derivatives at spatial frequencies \( R' \) and 0 makes it likely to give rise to the Gibb's phenomenon or ringing at sharp boundaries.

The ringing effect may be reduced by using a smoother function such as the Shepp and Logan \( H_{SL} \), or Hanning \( H_{HR} \) filter:

\[
H_{SL}(R) = \frac{2R'}{\pi} \sin(R) \quad (3.3.1.3) \quad H_{HR}(R) = \frac{1}{\pi} \frac{1 + \cos(R/R')}{2} \quad (3.3.1.4).
\]

These filters reduce the effect of ringing but give somewhat poorer spatial resolution than the Bracewell filter.

### 3.4. Iterative Reconstruction Techniques

Iterative techniques are another popular family of methods of solving the reconstruction problem. The basis of the technique is to divide the image region into a square array of pixels, the aim being to determine the values of the image density, \( f(x,y) \). An arbitrary starting value is assigned to each pixel, and a series of 'pseudo raysums' are calculated from this initial array. Comparison is made between the pseudo raysums and those observed in the genuine projection data, corrections are then applied
Figure 3.3.1.

(i) Modulation Transfer Function and
(ii) Impulse Response Function of the Bracewell Filter (Foster 1981)
to the pixel values in order to improve the match between the actual and calculated raysums. Each successive attempt is termed an iteration, and the process is repeated until the matching accuracy is consistent with the errors in the measured projections. The optimum number of iterations has been found to be typically 5-10 [Brooks 1976].

A wide variety of iterative techniques have been used, employing different methods and sequences of correction. The simplest mode of correction is to calculate all the pseudo-projections at the start of the iteration, then to apply all the corrections to all the pixels simultaneously. This method is known as the Iterative Least Squares Technique (ILST), and was first used by Bracewell [1956]. A damping factor must be applied to all corrections, otherwise the technique may never converge due to the high degree of over-correction inherent in the method. The name for the technique is derived from Bracewell's original mode of damping, which aimed to produce the best least squares fit to the observed data after each iteration.

Hounsfield's original EMI brain scanner [Hounsfield 1972] used a form of ray by ray correction, nowadays known as Algebraic Reconstruction Technique (ART), the name having been used by Gordon [1970], who utilized the technique in the field of electron microscopy. The basis of the method is to calculate all the raysums for a given projection, then to apply the appropriate correction to all points, and then repeat for each projection. This technique has been found to converge quite rapidly at the early stages of reconstruction, however it has been found to be unsuitable for applications containing a high degree of noise [Brooks 1976].

A third method of iterative reconstruction technique, first developed by Gilbert in electron microscopy, is the Simultaneous Iterative Reconstructive Technique (SIRT). In this technique, all raysums through each pixel are calculated, the appropriate correction is then applied to
that pixel and then the process is repeated for all image pixels. SIRT is slower than either ILST or ART because of the number of corrections that need to be performed.

The method of correction employed may be either multiplicative, in which the degree of correction applied to each pixel is in proportion to its density, or additive, in which the correction is in proportion to the pixel's weighting factor. The weighting factors are assigned to take into account the contribution of each pixel to a given raysum. For a single narrow ray the weighting factor is zero for most pixels. A value of 1 may be assigned to each pixel which lies within a ray otherwise the value assigned may depend on the distance of the pixel centre from the ray. The former process is simplest to implement but sometimes less accurate than the latter.

A number of constraints are normally placed upon the calculated values of \( f(x,y) \) in order to increase the processing speed. These include disallowing values of density which are negative or above a certain limit.

Comparison of the different iterative techniques has shown that ART is the fastest method, but that ILST and SIRT are more accurate in the presence of noisy data. As a result it has been suggested that the optimum technique would be to employ ART at the beginning of reconstruction, moving to ILST after the first few iterations (Brooks 1976).
3.5 Emission Tomography.

The technique of X-ray CT relies upon the transmission of a beam of radiation through the object under study. Tomographic methods are also valuable in determining the spatial distribution of radionuclides within matter using measurements of the radiation emitted by the radionuclide, a technique known as emission tomography.

In the case of transmission CT (see fig 3.21) the imaged parameter is \( \mu \), the linear attenuation coefficient distribution of the object under investigation. If we consider fig 3.21 and replace the external radiation source by a distribution \( A(x,y) \) of a gamma emitting isotope, then the countrate detected by a tightly collimated detector viewing along a line \( s \) is given by:

\[
A(r, \theta) = \int_{x} A(x,y) \exp\left\{- \int_{x'} \mu(x',y')ds\right\} ds
\]

where \( \mu(x',y') \) represents the linear attenuation coefficient at a point \( (x',y') \). This equation assumes that the detector is sufficiently far from the object for the solid angle subtended by all points \( (x,y) \) at the detector to be approximately the same.

The two integrals are not separable, and so some method of estimating the attenuation is necessary if accurate values of \( A(r, \theta) \) are to be obtained. In addition, in almost all practical situations a high count-rate is required, and thus the detector is mounted as near to the object as possible, and hence some correction must also be made for solid angle variation between points within the object.
3.5.1 Solid Angle Correction.

The solid angle or geometrical factor refers to the variation in calculated source activity resulting from the change of the detector collimator field of view with distance from the detector. This is easily corrected for if the spatial distribution of the source is known prior to scanning, but clearly this is not generally the case as emission tomography is often performed in order to determine the spatial distribution of the radionuclide.

The geometrical factor can be minimised if an opposed detector geometry, i.e. two identical detectors placed diametrically opposite each other, is employed to detect the emitted photons. This method has been used in vivo (Arimizu 1969, Tothill 1971) with some success.

The operation of the technique can be seen from Fig 3.5.1 for a point source at the centre of a circular object. The output from the two detectors will be given by:

\[ I_1 = I_0 \exp(-\mu(r+x)) \]  
\[ I_2 = I_0 \exp(-\mu(r-x)) \]

where \( I_1 \) and \( I_2 \) are the respective detector outputs and \( I_0 \) is the detector output in the absence of absorber. If we take the geometric mean (i.e. the square root of the product) of the two detector outputs:

\[ (I_1 I_2)^{1/2} = I_0 \exp(-\mu r) \]

Expression (3.5.1.3) is independent of \( x \) and thus of source to detector distance. The second detector may be omitted if a single detector is used.
to take opposing views, although this requires a doubling of the scanning
time.

3.5.2 Attenuation Correction.

For most of the energies used in emission tomography, the attenuation
of the emitted photons is considerable. The effect of photon attenuation on
uncorrected emission tomograms is to cause serious errors in the
determination of activities within the scanned volume, and spurious spatial
distributions may also result [Kay 1974]. Scans obtained without
attenuation correction are of value in some qualitative applications, but
for accurate quantitative imaging, some method of attenuation correction is
necessary.

No exact analytical method of correcting for attenuation exists,
however several approximate techniques for correction have been employed.
These may be broadly divided into three groups:

Pre-correction techniques – applied prior to reconstruction.

Post-correction techniques – applied after reconstruction.

Intrinsic techniques – applied during reconstruction.

Pre-correction techniques have, so far, been the easiest to employ
[Budinger 1977, Kay 1975], but they have proved of limited value in
situations where the sources are distributed in large, non-uniform volumes.
Many pre-correction techniques employ simplifying assumptions such as a
uniform attenuation coefficient throughout the image. Some methods [Kuhl
1976] involve a correction matrix obtained from phantom studies being
stored in the memory of the computer. Uncorrected phantom data may then be
multiplied by the correction matrix prior to reconstructing the image. The
use of a phantom study provides the best results for highly non-uniform
distributions.
If two or more gamma rays are emitted by the distributed source then
the attenuation of any pair of gamma ray will not be identical and the
ratio of the count rates of the two photons may be used to indicate the
attenuation [Cline 1972], if their emitted intensity ratio is known.

Post-correction techniques have proved more accurate than pre­
correction methods [Chang 1979], but have not always given satisfactory
performance. The techniques may check for self-consistency within the
measured data if uniform attenuation is assumed, and form a corrected
image using an iterative technique to improve the agreement between
calculated and observed data. If the volume contains a high degree of non­
uniformity in attenuation coefficient then the assumptions about the image
data will be incorrect and thus may lead to inconsistent results.

Intrinsic techniques have been reported to possess considerable
potential for giving accurate quantitative values for radionuclide
distribution [Gullberg 1982, Webb 1985], but have not been widely employed
to date. The basis of the method is to use a transmission CT scan performed
at the same energy as the emission scan, and then use the transmission data
to correct the emission projections during an iterative (ILST)
reconstruction process. The technique requires additional time and
computing power as two sets of measurements and reconstructions have to be
performed, and involves increased patient dose in medical imaging.

The various forms of attenuation correction have been reviewed by
Choudhary [1987]. Most forms of correction employ a number of simplifying
assumptions such as uniform attenuation.
3.5.3 Scattering Correction.

Compton scattering is the dominant attenuation process for photons of the energies most commonly encountered in emission tomography. Scattered photons have not traversed a straight line between the source and the detector and thus are a source of unwanted noise, causing a loss of sharpness, reduction in image contrast and overestimate of isotope activity. In-scattered photons may also cause a ring of apparently increased activity about small distributions of an isotope if an attenuation correction is employed, since some of the 'attenuated' photons will be recorded.

In principle scattered photons may be rejected by utilising appropriate energy discrimination, since Compton scattered photons have an energy lower than the primary radiation. However, in practice most of the detectors employed are of the scintillator type with relatively poor energy resolution, and are unable to resolve the small-angle scattered photons from the transmitted radiation. Thus the scattered photons cannot be rejected without losing a considerable number of valid counts.

Several methods have been employed to reduce the effect of scattering on emission tomograms, these include:

(a) Using a secondary energy 'window', set to accept photons of a slightly lower energy than the primary photons. The assumption is made that the number of counts recorded in the scattering window is directly proportional to the number of scattered events within the primary window. The number of events recorded within the scattering window may then be multiplied by the relevant factor and subtracted from the primary count. This approach requires a system capable of performing dual energy measurements and causes an increase in computation time. If necessary a separate scan may be performed to form an image using the scattering
window, this second image may then be subtracted from the initial tomogram. However, this method doubles the scanning time unless data in the two windows is collected simultaneously.

(b) Using a 'deconvolution' method [Axelsson 1984]; in this technique the effect of scattering is assumed to cause a uniform loss of sharpness in the image. If this loss of sharpness can be determined, then the blurring may be deconvolved during reconstruction using an appropriate function. In practice the blurring due to scattering is not constant with source depth and thus the correction will not be totally accurate.

(c) Computer simulations may be used to model the effects of scattering using Monte Carlo techniques [Beck 1982]. An image may be formed using the computer model, and this may be subtracted from the recorded emission tomogram. The accuracy of this technique is, of course, limited by the accuracy of the scattering model. In addition this method of correction is costly in terms of computer time.

(d) An 'effective attenuation coefficient' may be employed during the attenuation correction, this coefficient is lower than the true linear attenuation coefficient to allow for the scattered photons which are recorded in the primary energy window [Larsson 1980, Choudhary 1987]. This simple method improves image uniformity by eliminating the bright ring around 'hot-spots', but is not very effective in reducing the loss of image contrast caused by scattering.
3.6 'Stimulated Emission' Tomography.

As described in the previous chapter, an incident beam of ionizing radiation may be used to produce secondary photons within an object. These secondary photons may be produced by Rayleigh or Compton scattering, X-ray fluorescence or pair production. Measurements of the intensity of the secondary radiation can be reconstructed to form an image of the distribution of the secondary emitters in a manner similar to that employed with the more usual forms of emission tomography.

The spatial resolution of such images may be determined by using a collimator to limit the detector field of view. However, considerably higher counting rates may be achieved by employing an uncollimated large-volume detector in association with a narrow interrogating beam. Using this latter technique a line source of emitted photons is produced along the beam path; spatial resolution is determined by the beam width. The raysums are thus approximate line-integrals of the emitted flux along the beam, and may be reconstructed in the same manner as other tomographic images.

Stimulated emission tomography suffers from the same geometrical, scattering and attenuation difficulties as emission tomography, however in general the problems are somewhat more complex as:

(i) The beam will have finite divergence through the object.

(ii) In the absence of a detector collimator a larger number of scattered photons will be detected.

(iii) Both the stimulating and secondary radiation will be attenuated within the object.

Some difficulties encountered with these modes of scanning will be described in chapters 5 and 6.
Chapter 4.

Industrial Applications of Transmission Tomography.

The success of transmission CT scanners in the medical field soon led to investigations of the possibility of their employment in industrial applications. Early work by Foster [1981] showed that whilst a typical medical CT scanner (EMI CT5005), produced good images for some objects, for others severe problems were encountered. Before examining the reasons for these difficulties it is valuable to consider the desirable qualities of a medical CT scanner.

4.1 Properties of Medical CT Scanners.

For the purpose of diagnostic imaging it is a primary consideration to ensure that the risk to the patient is as low as possible. With ionizing radiations, this entails the minimisation of the patient dose. For a radiation dose of comparable magnitude to that involved in a conventional radiograph (1-10 mSv) approximately $10^7$-$10^8$ photons may be delivered during a CT scan, corresponding to about $10^5$ photons per pixel.

In addition to dose considerations, in order to produce a high quality CT image it is important that the object either remains stationary or (as in the case of dynamic imaging of the heart) moves in a cyclic manner. As a result it is normally necessary to perform medical scans over as short a period as possible; to achieve this medical CT scanners need to utilise an X-ray source to obtain an adequate photon intensity. Counting time is further reduced by using multiple detector arrays and fan-beam geometry.

Since medical scanners cost a great deal of money (typically about
it is also important to make maximum use of scanner time and thus it is helpful to ensure that the time taken for data processing does not greatly exceed that required for scanning.

The primary aim of medical CT scanners is to give a high quality image of the patient, thus enabling the radiologist to make the correct diagnosis. Obtaining quantitative attenuation values within the object is of lower importance and as a result the reconstruction algorithms used in medical CT work sometimes contain features which may reduce quantitative accuracy to enhance contrast.

Medical scanners are normally categorized by a 'generation', which refers to the source-detector arrangement. The various categories of scanner are listed in table 4.1, the motion column refers to the X-ray tube and detectors except in the case of fourth generation scanners, which have a rotating tube placed within a stationary 360° ring of detectors. Scan time refers to a typical single slice through the human body.

<table>
<thead>
<tr>
<th>Gen.</th>
<th>Detectors</th>
<th>Beam</th>
<th>Motion</th>
<th>Scan Time</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Single</td>
<td>Parallel</td>
<td>Translate and rotate</td>
<td>Minutes</td>
</tr>
<tr>
<td>2</td>
<td>Several (3-60)</td>
<td>Fan</td>
<td>Translate and rotate</td>
<td>30-100 sec</td>
</tr>
<tr>
<td>3</td>
<td>Multiple (~700)</td>
<td>Fan</td>
<td>Rotate</td>
<td>1-8 sec</td>
</tr>
<tr>
<td>4</td>
<td>Multiple (~2000)</td>
<td>Fan</td>
<td>Tube only rotates</td>
<td>1-5 sec</td>
</tr>
</tbody>
</table>

The early commercial scanners (eg EMI 5005) were of the second generation type, using up to 60 detectors. At the present time, both third and fourth generation scanners are commercially available as neither geometry has demonstrated marked superiority. Third generation scanners possess two disadvantages when compared with fourth generation systems:
(i) The detectors at the centre of the arc do not view the source through air and are thus cannot be continuously recalibrated. As a result detector drift may occur.

(ii) Each detector views a limited number of volume elements and as a result the failure of a detector leads to a ring artifact in the final image.

In fourth generation scanners, continuous detector calibration can be performed; the result of a detector malfunction is also reduced as the effect is averaged across the whole image. However, the detectors used in fourth generation scanners employ larger apertures than in third generation systems, leading to a loss in spatial resolution and increased acceptance of scattered radiation, resulting in less accurate determination of density.

Most industrial operators of X-ray CT have preferred 3rd generation systems, due to their superior density resolution.

4.2 The Suitability of Medical CT Scanners for Industrial Tomography.

As outlined above, the design constraints of a medical CT machine compels the use of an X-ray source and multiple detectors, leading to a machine which is very expensive both in initial outlay and maintenance costs. Medical CT scanners are optimised to image objects similar to the human body, i.e. with a reasonable degree of circular symmetry and consisting of low density, low atomic number (Z < 20) material. As a result, good quality images of biological objects such as tree trunks have been obtained using medical scanners (Foster 1981). Square objects containing material of high density (e.g. reinforced concrete), however, displayed severe artifacts when scanned using the EMI medical scanner, in addition the
scanner gave grossly incorrect attenuation values within such objects.

The medical scanners clearly perform their designed function of diagnostic imaging of human patients very well. In addition they may be valuable in industrial applications which require similar performance characteristics to those used in diagnostic medicine, i.e. rapid imaging of fairly uniform circular objects of low atomic number. However, for the majority of industrial purposes, medical scanners are far from ideal, being expensive, bulky and producing poor results for all but a small range of objects. For most industrial applications use of a purpose designed instrument offers considerable improvements in performance at lower cost, if some sacrifice in measurement time is acceptable.

4.3. Industrial Tomography.

The industrial uses of CT are wide ranging, and as a result the optimum design of any scanner will vary depending on the desired application. It is, however, possible to set down some optimisation criteria for scanner design based on a range of considerations.

It should be noted that although the range of sizes and composition of industrial objects varies considerably more than is the case in medical radiography, inanimate objects possess several properties that considerably ease the requirements of scanner design:

(i) In contrast to people, the vast majority of industrial objects are able to absorb very high radiation doses without ill-effect, thus in most non-medical applications the number of photons that can be employed is limited solely by the time available for scanning.

(ii) The scanning time can often be much longer than for medical
scans, measurements lasting minutes or hours rather than seconds may be used, enabling radioisotope sources to be employed in some applications.

(iii) Normally industrial samples are static and thus no compensation for internal motion needs to be performed.

(iv) The ability to move the object rather than the source-detector assembly permits the use of relatively compact apparatus.

The design of industrial CT machines will be dependent upon the nature of the object, the desired accuracy, resolution and scanning time. In addition financial constraints may have an important effect on scanner configuration. The following section will outline some of considerations to be made for industrial CT.

4.4 Selection of Source for Transmission Tomography.

The type of source best suited to a transmission CT scan will depend on the nature of the object under investigation, its size and elemental composition.

At the time of writing the basic choice is between the use of a radioisotope or an X-ray generator. The use of an X-ray generator, employed universally in medical CT machines, does provide several useful features. These are:

(i) X-ray machines produce much higher counting rates than those obtainable using most available radioisotope sources. A typical X-ray tube employed in a commercial C.T. machine has an equivalent activity to a 500 TBq (15000 Ci) source, permitting scanning times orders of magnitude lower than with a system based around a practical radioisotope source.

(ii) By varying the kilovoltage applied to the tube, an X-ray set can
readily provide a range of photon energies. (iii) The X-ray machine presents no radiation hazard when switched off, thus removing some of the difficulties which may be encountered in storing and handling high activity radioisotopes.

Despite these advantages, the X-ray generator is far from perfect as a source for quantitative transmission tomography. X-ray machines and their associated power supplies are both expensive and bulky, and require regular maintenance. X-ray tubes are generally not able to attain energies above a few hundred keV. In addition the output of the tube may vary due to small voltage drifts, take-off angle or tube age and thus may require frequent recalibration.

The chief drawback of X-ray machines lies, however, in the continuous nature of the photon energy spectrum produced. The attenuation coefficient of a substance is a function of the energy of the incident radiation, and thus the lower energy radiation is preferentially absorbed by the outer regions of the object, an effect known as 'beam hardening'. Beam hardening produces image artifacts and can lead to misleading values for the attenuation coefficient of the object. The effects of beam hardening can be reduced by using a pre-calibration for the objects under scan. This method is only really effective with fairly homogeneous objects, and whilst being quite useful in medical CT, is of very limited value in most industrial applications.

Radioisotopes may be used as convenient sources of effectively monoenergetic gamma photons, thus disposing of the beam hardening problem and making absolute determination of attenuation coefficients within the object straightforward. As outlined above their principal disadvantage is that their brightness is several orders of magnitude less than that obtainable with typical X-ray sources, leading to proportionately longer scanning times. However, since scans of industrial objects lasting a few
hours are commonly quite feasible this disadvantage is not too serious. Most radiation sources are inexpensive when compared to X-ray machines, and they are compact and easily transportable, although at high energies (>500 keV) the massive shielding required can become somewhat inconvenient.

4.5 Selection of Gamma Ray Energy for Transmission CT.

Gamma ray sources with useful half-lives are commercially available (e.g., from Amersham International plc) in the energy range 6 keV - 2 MeV. The correct choice of which energy to employ in any given application is dependent upon several factors, but chiefly on the dimensions and composition of the object. Over the range of energies under consideration here, the value of attenuation coefficient falls with increasing energy and in general higher energies are required for imaging large objects or those with high densities. In addition it should be remembered that high energy sources can generally be produced with higher useful activities than with low energy gamma sources due to the lower amount of self-absorption occurring within the source.

The selection of the optimum source energy for gamma ray tomography has been studied in detail by Foster [1981] and Folkard [1983].

If a slab of material of thickness $x$ is placed in a narrow beam of monoenergetic gamma rays and the number of transmitted gamma ray is written as $I$, then the value of the linear attenuation coefficient, $\mu$, is given by:

$$\mu = -\frac{1}{x}\ln\left(\frac{I}{I_0}\right)$$

(4.51)
where $I_0$ is the number of incident gamma rays. Neglecting the effects of background, the thickness of absorber for which the error in $\mu$ is a minimum may be determined as follows:

The uncertainty in $\mu$ is:

$$\delta \mu^2 = \left( \frac{df}{dx} \right)^2 \delta x^2 + \left( \frac{df}{dI} \right)^2 \delta I^2 + \left( \frac{df}{dI_0} \right)^2 \delta I_0^2 \quad (4.52)$$

$$= \left[ \ln \left( \frac{I}{I_0} \right) \right]^2 \delta x^2 + \frac{1}{x^2} \left[ \frac{\delta I}{I} + \frac{\delta I_0}{I_0} \right]^2 \quad (4.53).$$

Using $\delta I^2$ and $\delta I_0^2 = I$ and $I_0$ respectively and $I = I_0 \exp(-\mu x)$ gives:

$$\delta \mu^2 = \left[ \ln \left( \frac{I}{I_0} \right) \right]^2 \delta x^2 + \frac{1}{x^2} \left[ \frac{1}{I} + \frac{1}{I_0} \right] \quad (4.54)$$

$$\delta \mu^2 = \frac{1}{I_0 x^2} \left( 1 + \exp(\mu x) \right) + \mu^2 \left( \frac{\delta x}{x} \right)^2 \quad (4.55)$$

It is convenient to determine the fractional error in $\mu$ as a function of the number of mean free paths, $\lambda$ ($\lambda = 1/\mu$):

$$I_0 \langle \delta \mu \rangle^2 = \frac{1}{\langle \mu \rangle} \left[ 1 + \exp(\mu x) \right] + I_0 \langle \delta x \rangle^2 \quad (4.56)$$

The ideal thickness for minimum variance in attenuation coefficient occurs when:

$$\frac{d(\delta \mu^2)}{dx} = -2 \left[ 1 + \exp(\mu x) \right] + \mu \exp(\mu x) - 2\mu \delta x^2 = 0 \quad (4.57)$$

giving the optimum value of $\mu x$ when:

$$\exp(\mu x) = \frac{2 + K(\mu x)^2}{\langle \mu x \rangle - 2} \quad \text{where } K = (\delta x/x)^2 \quad (4.58).$$
This is true when \( \delta x = 0 \), i.e. when \( \langle \mu x \rangle = 2.2 \) [Folkard 1983]. This shows that the best estimates of attenuation coefficient are obtained with the thickness of the sample is equal to about two mean free paths.

It should be noted, however, that the above theory assumes a constant number of incident photons, whereas in transmission tomography it is conventional to maintain a constant number of transmitted photons (for reasons of noise uniformity). Hence we should use \( I \) and not \( I_0 \) in (4.56):

\[
I(\delta \mu)^2 = \left( \frac{1 + \exp(-\mu x)}{\mu x^2} \right) + \left( \frac{\delta x}{x} \right)^2
\]  

(4.59)

For values of \( \mu x \) much greater than unity this formula tends to:

\[
I(\delta \mu/\mu)^2 \propto (1/\mu x)^2 + (\delta x/x)^2
\]  

(4.510)

This formula implies that, to obtain the most accurate value of \( \mu \), the thickness use should be as great as possible. This result is misleading, however, as in practice, the scan must be completed in a finite time. In fact, the number of incident photons is effectively fixed and thus the result of (4.58) is valid. The optimum energy of the gamma ray may only be determined roughly by this formula because the thickness of the object will vary from raysum to raysum during the scan. For a uniform circular object, the optimum diameter is equal to three mean free paths [Gilboy 1984] as shown in the following section.
4.6 Optimisation of Scanning Parameters.

Foster [1981] derived equations to predict the uncertainties associated with reconstruction from projection data containing noise. The model used assumes the geometry to be that of parallel beam tomography, and provides useful guidelines for the optimising of scanning experiments. The following symbols are used:

- **Scan Diameter** $D$. Reconstruction is performed on a square with side $D$.
- $N \times N$ pixels are used, each with width $w$.
- There are $S$ raysums per projection performed at a spacing $t$ ($t=D/S$).
- $M$ equally spaced projections are used over 180 degrees.

In general the pixel size is set to be the same as $t$, which ensures that all the raysums are contiguous and therefore all parts of the chosen section are scanned. The ideal relationship between $S$ and $M$ has been examined by several workers [e.g. Klug 1972, Folkard 1983], giving the optimum as in the range $\pi S/2 > M > S/4$.

If the object under investigation is a cylinder of homogeneous material with attenuation coefficient, $\mu$, and is irradiated by point source emitting $A$ gamma rays per second, the overall run time for the scan can be estimated using the expression derived as follows.

A detector of intrinsic efficiency, $\varepsilon$, is placed a distance $D$ from the source, if the detector collimator is of diameter $t$ then, in the absence of an absorber, the number of detected photons per second, $v$, will be:

$$v = \frac{A \pi t^2}{16D^2} \quad (4.61.)$$

The mean time for the detection of $n$ photons for a single raysum taken at a distance $x$ from the centre of the object is:
\[ \langle T_{\text{Ray}} \rangle = \frac{n}{\nu} \int_{D_D}^{D} \exp\left[\mu(D^2 - 4x^2)^{1/2}\right] \, dx. \] (4.62.)

\[ = \frac{n \sinh(\mu D)}{\nu \mu D} \] (4.63.)

If \( L \) raysums are used, where \( L = M \times S \) and \( M = \pi S/2 \), the time to complete the scan using a single beam is given by:

\[ \langle T_{\text{scan}} \rangle = \frac{8\pi M^2 \sinh(\mu D)}{\mu D_\Sigma} \] (4.64.)

This equation gives a useful 'rule of thumb' for estimating scan times. For example, using a 3.7Gbq source of \(^{241}\text{Am}\) on a sample of water of 100mm diameter using 1mm steps and 1000 cts/raysum a scanning time of about 30 minutes is obtained. In practice this constitutes a minimum time as finite time would be required for scanner motion and reconstruction, in addition the source to detector distance will generally exceed \( D \) due to the length of collimators and requirement of obtaining 'air values' for rays passing on either side of the object.

Foster [1981] showed that, for filtered back-projection, the absolute uncertainty in attenuation coefficient due to the effect of finite counting statistics, is given by:

\[ \Delta \mu = k(2\pi/nD_\Sigma)^{1/2} \] (4.65)

where \( k \) is a constant determined by the filter used.

Combining expressions (4.64) and (4.65) gives an expression for the relative uncertainties in measuring \( \mu \), caused by photon statistics:

\[ \frac{\Delta \mu}{\mu} = 4\pi k n^2 \cdot \frac{D_\Sigma}{\mu T_\Sigma} \cdot \frac{\sinh(\mu D)}{(\mu D)^3} \] (4.66.)
A plot of the function \( \frac{\sinh(\mu D)}{(\mu D)^2} \) is shown in figure 4.6a, showing a shallow minimum at \( \mu D = 3.0 \), corresponding to an optimum object size of three mean free paths. Due to the different sample geometry this result is not in precise agreement with expression (4.58), but since the minimum of fig 4.6a is very shallow, the disagreement between the two results is not very significant. Klyuev [1980] quoted a value of 4\( \lambda \) as the optimum sample thickness, but fig 4.6a shows that within the thickness range of 1 to 6 mean free paths the exact thickness value is not crucial.

The optimum source energy required for imaging using the three mean free paths rule is shown for some common materials in fig 4.6b.

Two important variables have not been fully discussed in the above section. The first is the detector efficiency, \( \varepsilon \), which is a function of gamma ray energy. The detector efficiency depends on a number of factors including detector size, geometry, mean atomic number and the nature of casing/window materials. In order to be detected, a gamma ray must interact within the detector, and thus the detector efficiency is limited by its attenuation coefficient and dimensions. High density, high Z materials are thus best suited to gamma ray detection [Knoll 1979].

In general the low energy efficiency of a detector is limited by penetration through the end-window, and at higher energies the probability of a photon interacting within the detector begins to fall. The maximum intrinsic efficiency for aluminium end window NaI(Tl) detectors similar to the device used in this work typically occurs at about 60 keV.

Improved efficiency at high energy may be obtained by using large detectors, but use of a large detector also increases the weight of the required shielding and the larger dimensions enables the interception of more background radiation. Ideal detectors for transmission tomography are smaller high Z types such as CsI, preferably with length in the beam axis direction greater than the diameter. The use of photodiodes in place of
Plot of "sensitivity factor" $[(\sinh x)/x^3]^{1/2}$; $X$ is the diameter of the test object expressed in mean free paths.
Optimum Thickness for some materials as a function of Photon Energy.
bulky photomultiplier tubes may also allow the use of increasingly compact
detectors for tomographic applications, which reduces the size and mass of
shielding required.

Low energy sources may be preferred in some applications because of
the greater difference in attenuation coefficients between different
elements at low energies. For studying fluid flow, for example, it is
possible to greatly increase contrast by doping the fluid with a solution
containing an element with a K absorption edge just below the gamma energy
being used. In such applications the large difference in \( \mu \) which ensues may
outweigh the effect of using a gamma energy slightly different from the
statistically determined 'optimum' value.

An important factor for consideration when selecting the optimum
source for a given application is available source brightness. This is of
particular relevance when using low energy sources, where source
brightness is particularly limited by self-absorption. The attenuation of
60 keV gamma rays in americium is approximately 300 times as great as that
of 1.25 Mev gamma rays in cobalt. As a result a bright \(^{57}\)Co source can be
constructed by increasing the source length along the collimator axis to as
much as 30mm (\( \mu_{\text{co}} = 0.047 \text{ mm}^{-1} \) at 1.25 MeV) whereas in an \(^{241}\)Am source
over 99% self-absorption occurs in 0.5mm.

Ursin [1988] reported obtaining more accurate determination of
densities using 662 keV radiation than with \(^{241}\)Am in studying rock cores,
due to the much higher statistical accuracy obtainable using high activity
\(^{137}\)Cs sources, despite the higher energy radiation being further away from
the three mean free path criterion.
4.7 Overview of Industrial Applications of Industrial Tomography.

The potential of CT as an industrial tool was recognised soon after the introduction of the EMI medical scanner and industrial CT has been employed in a wide range of applications since the late 1970s [Reimers 1984a].

The first worker to use a gamma ray source for transmission tomography was probably Sweeney [1974], working at the Lawrence Livermore Laboratory. His scanner utilised a high activity (1.5 TBq) source of $^{153}$Ir, which emits a series of gamma rays in the range 296-612 keV, in conjunction with a scintillator detector. His initial scans of such objects as a lens assembly, obtained using scanning times of a few hours, were somewhat disappointing. This work, however, paved the way for much more powerful scanner systems which could be applied to a wide range of industrial applications.

Kruger et al [1978] at Los Alamos started work into investigation of the suitability of X-rays, γ-rays and neutrons for non-medical CT. Much of the initial work was devoted to computer simulations with the aim of optimising the scanner configuration, but a simple scanner was built and used to image reactor coolant pipes using a 50μm wide beam of gamma radiation from a 37GBq (1 Ci) $^{152}$Ir source. Stress-corrosion cracks less than 0.3 mm wide were detected using this machine [Kruger 1982].

Among the first companies to design an industrial CT machine using gamma sources to perform scanning on a commercial basis was Scientific Measurement Systems (SMS) of Austin Texas [Hopkins 1981]. The scanner employed $^{137}$Cs (59 GBq) and $^{192}$Ir (7.4 TBq) sources used in conjunction with plastic scintillator detectors operated in the photon counting mode. A total of thirty one detectors were employed in a fan-beam arrangement; the different projections were obtained by rotating the sample in the beam,
although the detector array was also stepped over a small angle to fill in additional ray paths. Using the SMS machine to image wooden poles used for supporting electrical power cables, the internal structure of the poles was found to be clearly visible. Reinforced concrete structures were also imaged successfully using this device. SMS also produced a commercial scanner using the $^{192}$Ir source, capable of scanning objects up to 100 kg in mass.

Recent years have also seen increasing usage of CT in on-line testing, for example in steel production, where product wastage due to poor quality control is often excessive. The ability of gamma ray techniques to be employed on-line with hot materials can produce considerable economic benefit [Taylor 1989]. Sweany and Abdou [1987] have produced a fourth generation CT machine using 128 detectors for use with Bethlehem Steel in order to determine the product dimensions during manufacture.

Some of the largest objects to have been studied using CT techniques have been rocket engines. American Science and Engineering (AS&E) working in conjunction with Lockheed Missiles and Space Company (LSMC) and Hercules Inc. and Thiobel Corp. have produced large scale systems capable of imaging objects up to 2 m in diameter with spatial resolution of a few millimetres [Burstein 1982]. The X-ray source used was a Varian Linatron L-6000 accelerator, located one one side of the scanner bed, and to limit scattering to a minimum the beam was both pre- and post-collimated. This source is capable of producing bremsstrahlung X-rays with energies up to 16 MeV. The accelerator provides a practical, if expensive, method of generating very intense beams of the high energy radiation necessary when scanning large, high density objects. A fan beam and multiple scintillator detectors were employed, the system has been used to study a 1.9 m diameter Trident C4 rocket engine. The Aerojet Strategic Propulsion Company (ASPC)
of Sacramento California have demonstrated an even larger system capable of imaging a 2.3 m diameter Peacekeeper ICBM stage 2 motor.

4.7.1 Industrial CT Outside the USA.

One of the first groups in Europe to use CT for non-medical applications was at the University of Surrey. Foster [1981] and Folkard [1983] employed an EMI CT5005 scanner at the King Edward VII hospital at Midhurst, Sussex, to scan a range of industrial and archaeological objects. Difficulties encountered when imaging some objects led to the construction of a radioisotope based scanner, which was utilised for the scanning of a variety of items, including low density inorganic foam using 17 keV radiation from an $^{241}$Am source, dry cell batteries using the main 60 keV gamma from $^{241}$Am and a transformer using 122 and 344 kev gamma radiation from $^{152}$Eu.

Much of the transmission CT work at the University of Surrey has been carried out using a 3.7 Gbq (200 mCi) $^{241}$Am source in conjunction with a collimated NaI(Tl) detector, the scanner, described in detail below was a simple single beam-machine. Useful results were obtained when imaging gas fluidized beds [MacCuaig 1985] and in determining the fluid contents of porous rock core samples [Gilboy 1984] with the purpose of improving the recovery of oil from offshore oil reservoirs. Typical object sizes were 30-120 mm, and with spatial resolutions of between 1 and 2 mm useful images were obtained with scan times varying between 20 minutes and 5 days.

An early application of an X-ray CT scanner for non-medical purposes was by the Royal Armament Research and Development Establishment (RARDE) at Sevenoaks, Kent [Hunt 1981]. Thorn/EMI developed the system which resembles the EMI medical scanner except that a fixed source and detector was
employed in conjunction with a moving object. The X-ray tube was uprated from 150 to 450 keV to image objects of up to 155 mm diameter. Scanning of the object takes approximately one minute and reconstruction of a 320 x 320 pixel image about four minutes. A similar device has been developed by the French Atomic Energy Commision (CEA), using a single CsI detector and a 420kVp X-ray source [Huet 1979, 1982].

In West Germany the applications of CT for non-destructive testing were first studied at the Bundesanstalt für Materialprüfung (BAM), Berlin. Just as at the University of Surrey, initial studies were performed using an EMI CT5005 scanner, but later a purpose-built industrial machine was constructed using 31 plastic scintillator detectors in conjunction with either an X-ray or isotopic gamma source [Reimers 1984b]. The device is capable of use with objects of 1 metre diameter weighing up to 1000 kg and has been routinely used for commercial scanning for such varied objects as mummified heads, ceramic insulators and thick-walled steel pipes.

Some of the smallest objects to have been scanned using CT are the shells of tropical fresh water snails. Elliot [1982] working at the London Hospital imaged the snails with a modified scanning microradiography apparatus using Cu K X-rays (88 keV). Images were obtained with spatial resolutions as small as 15 μm in about 2½ minutes. Investigations of 'microtomography' have also been carried out by Sato et al [1981, 1984] for studying electronic chips with dimensions of the order of 250 μm using 7 keV X-rays. Spatial resolutions of 15 μm were achieved in around 8 minutes using 16 projections and 128 raysums. These images are heavily underdetermined and considerable improvements in quality could be obtained using a greater number of projections.

Over the past decade a number of Japanese workers have been developing the CT technique in a wide range of applications. Onoe et al [1984], have developed a portable X-ray CT scanner for measuring the annual rings of
live trees. The CT technique is particularly well suited to this area of study, the subject is circular and of low density, allowing good images to be formed quite quickly. In addition the non-destructive nature of CT is particularly valuable, and can lead to pre-selection of the correct timber for applications which require large wood sections free of defect. The choice of a small 40-120 kVp X-ray tube rather than a radioisotope source enabled faster scanning and the loss of accurate quantitative attenuation coefficients was not found to be important. Three NaI(Tl) scintillation detectors were employed in a fan-beam geometry.

The technique of CT can be used to obtain time-average density measurements of dynamic systems such as the void fractions of liquids flowing in aluminium and steel pipes. In experimental trials, low spatial resolution (4mm) scans of a steel pipe 4mm thick and 61mm diameter have been imaged successfully using 3.7 MBq (100 μCi) $^{137}$Cs source in a time of a couple of hours [Tsumaki et al 1984]. A commercial device using a fan beam, multiple detectors and a source activity of 37 GBq (1 Ci) would enable such scans to be performed in seconds.

Japanese workers have also made use of CT scanning in the nuclear industry. A 420 kVp X-ray based machine, built by the Toshiba Corporation has been used for the detection of stress corrosion cracks in the welds in a nuclear power plant [Miyoshi et al 1987]. An improved device utilising a linac X-ray source has been proposed.

A similar device to that designed by AS & E has been produced by Hitachi in Japan for study of solid fuel boosters for the Japanese space program. The system is a second generation type using 80 BGO detectors and a 15° fan beam of bremsstrahlung radiation produced by a Mitsubishi 12 MeV LINAC, filtered to reduce beam-hardening. The system is capable of imaging a section of a 1.8m diameter rocket engine in about 7 minutes [Kanamori 1989].

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The use of CT in crack detection in steel has been studied by Sawicka and Tapping (1987), using a first generation scanner incorporating a single CsF detector and 440 GBq (12 Ci) $^{60}$Co source; both source and detector were fitted with collimators 165 mm in length and a spatial resolution of 0.5 mm was obtained. Typical scans required that $2 \times 10^{8}$ photons be detected when imaging $17 \times 20$ mm$^2$ steel billets, which enabled density resolutions of the order of 2% to be obtained; this allowed cracks to be detected with widths of as little as 45 μm. These measurements were reported to be both faster and less labour intensive than traditional techniques involving microscopy.

4.6 Conclusions and Future Trends in Industrial CT

The technique of transmission CT has been demonstrated in a wide range of industrial applications, and has proved successful in imaging objects ranging in size from sub millimetre to over 1 metre. Many early workers used scanners designed for medical use, but the high cost and limitations in both accuracy and versatility of medical X-ray based systems precludes their universal employment. Nevertheless, some large companies which require to scan large numbers of fairly uniform objects rapidly may still employ medical scanners (Hayes 1988).

A growing number of workers have sought to design machines for industrial use, and specialised machines for specific applications have been demonstrated. Some large engineering firms (e.g. Hitachi in Japan) have designed CT machines for performing quality control sampling. The number of companies operating such machines is likely to increase, and the availability of relatively low-cost LINACS (~£100,000 for a 10 MeV device), enables systems to be devised for scanning large scale objects over a metre
In diameter. Such machines would undoubtably be expensive, but for large companies manufacturing expensive products the machines could prove economical in the field of quality control.

Despite these advances, only a handful of companies (e.g., SMIS at Guildford, Surrey) actually produce industrial CT apparatus, and it is probable that many workers in the field of non-destructive evaluation remain unaware of the potential of CT. It is reasonable to assume a growth in the market for CT machines will occur, and some devices for use with large (>50cm) or small (<1cm) objects will have to be purpose-built, but it is also probable that a market for a general purpose multi-beam system will soon exist. Such a system should be able to incorporate any of a range of different radioisotope sources for different applications. However, for reasons of shielding it is not practical for a single device to cover the full range of required energies (say 60–1250 keV). A small-scale sixteen detector device may cost less than £150,000, which although representing quite considerable capital outlay, is not beyond the reach of large engineering firms. It is the opinion of the author that as CT becomes a more widely accepted NDE technique in industry the demand for CT devices will increase.
4.9 The Scanner Design.

The basic scanner system used for most of the transmission tomography work was designed at the University of Surrey, and has been used successfully for several years. The scanner system is of the first generation type, with a single detector and a collimated radiation source producing a single fixed beam. Two degrees of motion are required for scanning, linear translation and rotation. The object is placed upon a rotating table which can be moved perpendicularly across the radiation beam. The scanner system consists of four basic parts:

(i) The scanner bed and its drive motors.
(ii) The source-detector arrangement.
(iii) The counting electronics.
(iv) The scanner control system.

These components will be described in the following sections. During its lifetime the scanner has undergone several modifications, these will be outlined below. Figure 4.9a is a photograph on the system in its most recent form, figure 4.9b is a block diagram displaying the important features of the scanner design.

4.9.1 The Scanner Bed.

The gamma ray CT scanner was designed by Foster and Gilboy (Foster 1981), the original aim was to build the scanner in the departmental workshops, but eventually commercially produced components were used. These were quite economical and have since proved reliable in service. The
Figure 4.9a: Photograph of transmission CT scanner.
Diagram of Apparatus for Parallel Beam C.T. Scanning

Figure 4.9b:
The original cost of the scanner bed and stepper motors, purchased from Time and Precision Engineering at Basingstoke in 1979, was £1256.

The scanner bed consists of two stepper motor driven transports, providing linear and rotational motion in the horizontal plane, and electronic units to convert the TTL pulses from the controller into currents to drive the motors.

The slide assembly consists of an extruded aluminium frame 0.8m in length, containing a central lead-screw along the long axis. The dovetail base of the scanner bed houses a sliding carriage, the moving face of which has been laminated with Nylatron GS bearing pads. The stepping motor at one end of the base operates a lead-screw of 1mm pitch, rotation of which moves the carriage along the bed. The screw rotates by 1/400 of a revolution with each pulse received by the stepping motor, and therefore the minimum step length of the system is 2.5 microns.

To prevent damage to the bed by the carriage running into the end of the frame, a safety system is incorporated. Running along the length of the frame and parallel to the screw is a rod containing two moveable collars. Contact with either collar activates a microswitch which disables the stepper motor. The screw is then rotated manually to release the release the microswitch and reactivate the stepper motor. The maximum linear traverse between the two collars is 550mm, which constitutes an upper limit on the size of object that can be scanned.

The rotary table consists of a 120mm aluminium cylinder with a roller bearing. The minimum angular step is 0.01 degree and, under Digiplan CD70 control the rotation speed is 20 degrees per second. The maximum loading is 50 kg.
The motors are driven by Digiplan CD70 stepper motor drives mounted in a 19" rack, using a PM1200 24 volt, 7 amp supply.

Operation of the scanner bed is performed via a microcomputer controller, which allows the user to move or rotate the scanner through specified linear/angular increments up to 150mm/180 degrees.

4.9.2 Sources and Collimators for Transmission Tomography.

As discussed earlier, the selection of the gamma ray source best suited for a given application is in part based on consideration of the mean free path of the gamma rays in the object to be studied. At the University of Surrey, the most commonly used source to date has been $^{241}$Am, with a principal gamma ray energy of 59.54 keV. The energy of this source is comparable with the mean energy of medical CT X-ray sources, and has proved valuable for scanning a wide range of low density objects. The low energy of the primary gamma ray makes source shielding very simple and allows for safe and convenient handling. In addition, ready availability (from Amersham plc.), low cost (about £650) and 432 year half life make the use of $^{241}$Am economically favourable.

In transmission tomography the spatial resolution is defined principally by the detector collimation. The collimator thickness along the beam axis must be sufficient to reduce the total number of source photons passing through the collimator material to a small fraction (ideally considerably less than 1%) of that entering through the collimator orifice. The collimator thickness is thus determined by the attenuation coefficient of the collimator material at the required gamma energy. The detector shield should also be sufficient to reduce the background count-rate to a
value less than the minimum count-rate through the object.

The collimation of the source is also of importance for accurate determination of attenuation coefficient, as narrow beam geometry is assumed in equation 3.2.4. Scattered photons, entering the detector along a path other than along the source-detector axis, will produce a reduction in contrast and inaccurate values for attenuation coefficient. It is thus necessary to collimate the source to produce a beam of size similar to that of the detector collimator acceptance.

Table 4.9.2 shows the attenuation coefficient for lead at various gamma energies and the length of collimator required to give 99.9% attenuation (about 7 mean-free-paths). The values in the table are interpolated from those of Hubbell [1969].

Table 4.9.2 Attenuation Coefficients and Lengths for Lead Collimators.

<table>
<thead>
<tr>
<th>Energy (keV)</th>
<th>Isotope</th>
<th>( \mu / \text{mm}^{-1} )</th>
<th>Min. Length of Collimator</th>
</tr>
</thead>
<tbody>
<tr>
<td>60</td>
<td>(^{241}\text{Am})</td>
<td>5.01</td>
<td>1.4mm</td>
</tr>
<tr>
<td>100</td>
<td>(^{153}\text{Gd})</td>
<td>5.91</td>
<td>1.2mm</td>
</tr>
<tr>
<td>300</td>
<td>(^{192}\text{Ir})</td>
<td>0.43</td>
<td>16mm</td>
</tr>
<tr>
<td>660</td>
<td>(^{137}\text{Cs})</td>
<td>0.123</td>
<td>57mm</td>
</tr>
<tr>
<td>1250</td>
<td>(^{60}\text{Co})</td>
<td>0.058</td>
<td>120mm</td>
</tr>
</tbody>
</table>

It can thus be seen that high energy sources require long and bulky collimation systems, whereas low energy sources such as \(^{241}\text{Am}\) are relatively easy to collimate.

The most commonly used source was an \(^{241}\text{Am}\) bead source, 5mm in diameter with an activity of 7.4 GBq, obtained from Amersham International plc. This was the brightest commercially available \(^{241}\text{Am}\) source, emitting \(5.5 \times 10^7\) 59.54 keV gamma photons per second per steradian.

For most transmission measurements performed using \(^{241}\text{Am}\) the source
was mounted in a brass holder placed within a hollow cylindrical lead shield, with a 15mm diameter central orifice. The source is placed in a brass housing, which is held onto a brass holder by means of a grub screw. The holder is then inserted into the lead shield, and is held in place by a second grub screw. Various cylindrical collimators may be inserted into the lead holder using tweezers. The collimators are 10mm in length and are constructed of lead, wrapped in brass holding rings. The beam exit diameter may be chosen from 0.5, 1, 2 or 3mm, according to the desired spatial resolution. The dose rate at the surface of the source holder was measured to be 6 μSv/hr falling to 1 μSv/hr at 0.5m, thus permitting safe manual handling of the holder for the short periods required whilst setting up the scans. The high dose rate from the collimator exit aperture could be reduced to a satisfactory level by placing a 2mm thick lead sheet over the end of the collimator. A photograph of the collimator system is shown in fig 4.9.2.

A second collimator system designed by Folkard (1983) was employed for some work, this system used 50mm long steel collimators to give ribbon or pencil beams with widths from 0.1 to 5mm. The effect of the longer collimator was to greatly reduce scattering but at the expense of lower counting rates as a result of the greater source-detector distance.

Both source holders could be placed in a perspex sleeve for mounting on the optical bench system. The sleeve was designed to permit the mounting of a HeNe laser coaxially with source exit beam, to enable rapid beam alignment with the detector collimator.

4.9.3 The Detector System.

The detector most commonly employed was a 50mm diameter NaI(Tl) scintillator, manufactured by Oakfield Instruments. This detector had a
Figure 4.9.2: Collimators used in association with $^{241}\text{Am}$ source.
measured intrinsic full energy peak efficiency of 70% and a FWHM energy resolution of 13% at 60 keV. The high efficiency and relatively low cost of the sodium iodide made it a useful, although not ideal, detector for transmission tomography work, where high resolution is not of great importance. The principal disadvantage of the detector was that the useful volume of the detector (determined by the width of the collimated beam and gamma penetration) was quite small compared to the actual volume, leading to an excessively bulky shielding system. In addition, the unused detector volume was a source of noise as it served to detect background radiation.

For transmission tomography applications, the detector should be of sufficient size to stop most of the incident radiation, and thus should have dimensions of the order of 5 mean free paths (about 2.5mm in NaI(Tl)). Increasing the detector size slightly above this value may be beneficial in increasing the peak to Compton ratio, it is clear, however, that the size of detector used was very much greater than optimum. A small CdTe detector (1mm deep by 5mm diameter) was tested for possible tomography applications but its peak to Compton ratio was significantly inferior to the sodium iodide device. Compact detectors with high efficiencies such as CsI(Tl) coupled to photodiodes will probably prove valuable in the construction of detector arrays for CT scanning in the future. A 10mm diameter by 3mm deep CsI(Tl) detector has been shown to give 400 times reduction in background count compared to the 50mm by 50mm NaI(Tl) detector [Gooda 1988].

The NaI(Tl) detector was mounted in a specialised holder, of 5mm thick brass lined with 5mm of lead. This amount of shielding was highly effective in stopping background radiation of about 60 keV $^{241}$Am, though proved insufficient for use with higher energy $^{137}$Cs or $^{60}$Co sources.

The front of the detector assembly was designed to accept a wide range of collimators with apertures of both circular and rectangular cross-section. The collimator width could be selected from a range between 0.5
The both the source holder and the detector holder were mounted on an optical bench, which permitted rapid and accurate source-detector alignment.

4.10 Scanner Electronics.

The purpose of the scanner electronics was to:

(i) Amplify the detector output pulses.
(ii) Discriminate between pulses arising from the detection of transmitted gamma rays from the source and those arising from background or scattering.
(iii) Accumulate the valid counts.
(iv) Transmit the measured count-rate at the end of each raysum.

An Ortec spectroscopy amplifier was employed for the first stage of the system; a type 572 low noise amplifier with pulse pile-up rejecter was used initially. This device has a gain adjustable from 1 to 1500 and has switch-selectable shaping time. Amplifier pile-up time was insignificant for count rates of a few thousand per second when using a 2 μs shaping time. This amplifier subsequently developed a malfunction and an Ortec 575 amplifier was used instead. This much simpler model proved a satisfactory alternative when used in conjunction with the sodium iodide detector. Typical settings were gain of 30x and 2 μs shaping time when used with the NaI(Tl) detector operated at +700V.

4.10.1 The Single Channel Analyser.

In order to ensure that only pulses due to transmitted gamma photons were being counted, it was necessary to employ some kind of energy
discrimination. One method would have been to record the amplified output with a multichannel analyser (MCA), however such devices are expensive and suffer from severe ADC dead-time limitations at high counting rates. It was thus cheaper and more effective to use a single channel analyser (SCA) to define an energy 'window' around the full energy peak. Selection of the full energy peak rather than acceptance of the entire pulse height spectrum reduces the effect of background, electronic noise and in-scattering, producing superior image contrast and more accurate values for the measured attenuation coefficients.

The type of SCA used was an Ortec 550; lower and upper thresholds for pulse height were set using helipots. A TTL pulse is delivered from the SCA every time a pulse arrived between the two discriminator levels.

Setting of the upper level (ULD) and lower level (LLD) of the discriminator was achieved by displaying the amplified pulse height spectrum from the detector on an MCA. A region of interest around the full energy peak corresponding to the appropriate gamma ray energy was defined by feeding the output from the SCA to the MCA gate input, operating in coincidence mode. The SCA discriminator levels were varied until only pulses within the full energy peak region were accepted. An alternative method employed when difficulty was experienced with the MCA gate was to use a precision pulser to supply pulses of a height equivalent to the lowest channel of the region of interest on the MCA to the SCA. The SCA was then connected to a counter-timer and the LLD varied until the counter-timer just ceased recording. Pulses equivalent to the highest channel of the MCA region of interest were then supplied to the SCA and the ULD varied until the counter-timer just ceased to count. By this method only pulses occurring within the defined full energy peak would be recorded.
4.10.2 The Dual Counter-Timer.

A Canberra 2071 dual counter-timer with IEEE interface was purchased at a price of £1106. Switches on the front of the device enable it to be operated for a preset time or for a preset number of counts. The number of counts and the counting time is shown on a liquid crystal display. Setting is achieved using thumbwheels mounted on the front of the device; the maximum number counts that can be recorded is $10^6$. In transmission mode the device is normally set to read the time for a preset number of events, which enables uniform statistical uncertainties to be obtained for all raysums which ensures uniform image statistics.

The counter-timer was connected to the Hewlett Packard (HP) microcomputer controller using the IEEE interface, which enables the counter-timer to transmit the value of each raysum (normally the collection time) to the HP. The HP controller then resets the counter-time ready to record the next raysum.

4.10.3 The High Voltage Supply.

The high voltage to the photomultiplier of the sodium iodide detector was supplied using a Nuclear Enterprises NE4701 unit, purchased at a cost of £480. The device is a standard single NIM width unit supplying several milliamperes over a voltage range from $-5000$ to $+5000$ V. Normal operating potential for the sodium iodide detector photomultiplier was $+700$ V.

All the electronic units were mounted in a 19" NIM rack with a filtered mains power supply.
4.11 The Scanner Control System.

The scanner was originally controlled using a Motorola M6800 microcomputer, which served to drive the scanner and store the data as it was collected [Foster 1981]. The scan control program was written in machine language and was stored on the University PRIME computers, being transferred to the M6800 prior to scanning. This system worked quite well, but suffered from several limitations, most notably the absence of its own floppy disk unit. This resulted in the total number of raysums that could be collected being restricted by the amount of available random access memory (RAM). Since the M6800 possessed only 16K, of which 4k were required to store the scan control program the largest size of image that could be stored was 60 x 60 pixels. The M6800 determined the collection time for each raysum using its internal clock. All reconstruction was performed using the SNARK75 [Herman 1975] program run on the University of Surrey PRIME computers. The M6800 was too limited to perform image reconstruction and possessed no graphics facilities for image display.

In 1980 an Apple II Plus microcomputer was obtained, which with its colour monitor and twin disk drives cost £1900. The Apple II had 48k of RAM and some limited graphics facilities. The scanner program was written in Applesoft BASIC with key parts of the routine written in Apple machine code [Folkard 1983]. The time for each raysum was now recorded for each raysum by the use of an external counter-timer. The Apple was capable of reconstructing the image during scanning, the reconstruction routine was rather slow (several hours) being written in BASIC; however, this was not thought too great a disadvantage due to the generally long scanning times. The Apple subsequently developed a hardware fault, repair was impossible as the original suppliers had since gone into liquidation.

For a time scanning was continued using the M6800 and later a BBC
microcomputer was employed for data collection, although reconstruction was still performed using the University PRIME computers using the SNARK program.

4.11.1 The Hewlett-Packard Personal Computer.

As related above, the scanner system was still dependent on the use of University PRIME computers, for both reconstruction and image display. If the functions of scan control, reconstruction and image display could be performed by a single microcomputer, the scanner system could then become self-sufficient and portable.

MacCuaig [1986] purchased a Hewlett-Packard 9836 personal computer for about £18000 including peripherals. This device was able to perform all the necessary computing, allowing the scanner to be transported and operated outside the University of Surrey.

The Hewlett-Packard (henceforth referred to as the HP) 9836C is based around the Motorola 68000 16/32 bit microprocessor. The system consists of a high resolution 12" colour monitor, computer with 1 Mbyte internal memory, a 15 Mbyte Winchester hard disk, twin 5½" floppy disk drives (528 kbyte) and 'inkjet' printer. An inbuilt HP-IB IEEE-78 interface was used to communicate with the Digiplan stepper motor control unit and the counter-timer.

The scanner control program for the HP was written in PASCAL by Neil MacCuaig. The program is used to set up the scanner, operate the scan, and reconstruct the image for display and analysis. The total system including scanner and controller is thus self-sufficient and easily portable in a medium-sized car, taking two persons about half an hour to reassemble.
4.11.2 The IEEE Interface.

The IEEE-488 bus system is widely employed as an interface; its principal advantages are in ease of programming and in the simplicity of connection between controller and device which it permits.

The IEEE-488 specification defines the procedure for communication between a set of IEEE compatible devices. It determines the electrical connections and signal levels along with the 'handshake' procedures which are necessary for successful communication between devices.

The bus system is based on a highway which is connected to all devices in parallel. The controller (in this case the HP) 'supervises' the other elements in the system, which may be 'talkers' (i.e. send data) or 'listeners' (i.e. receive data) or both. The controller decides which components are 'talking' and which are 'listening' at any one time, as well as being able to send and receive data itself. Each device is identified by a different 'primary address'; data is transmitted in ASCII format, with the most significant character first. Eight data lines are employed by the interface, along with three handshake lines to determine whether a device is ready to transmit valid data (DAV), receive data (NRFD), or if it has successfully collected the data (NDAC).

The HP controller is interfaced to the Digiplan motor drive unit enabling translation and rotation of the scanner table, and to the Canberra 2071 dual counter-timer. The counter-timer transmits values of either time or counts to the HP during scanning, and is then is reset between each successive raysum. After receiving the raysum data, the HP sends a series of command pulses to the Digiplan unit, and current pulses are then sent from the stepper-motor drive to cause a linear movement of the desired amplitude. At the end of each projection, the projection data is convolved with the Bracewell filter function and the filtered data is back-projected across the image array. Thus over successive projections the image is
gradually reconstructed and displayed on the HP monitor. A series of command pulses are then transmitted from the HP to the Digiplan unit, causing the rotary table to be moved through the required angular interval.

Upon scan completion both the reconstructed image and the projected data are filed onto the Winchester disk for subsequent analysis.

4.12 The Scanner Control Program.

The scanner control program was written by MacCuaig (1986) in Hewlett-Packard Pascal. The program was designed using experience gained with the previous control systems, and was written to allow easy scanner operation with the following features:

(1) Movement of the scanner turntable, both linear and rotational motion being permitted, the magnitude of the movement being entered via the HP keyboard.

(2) Setting up of the scan parameters prior to scanning.

(3) Establishment of a background count correction prior to scanning.

(4) Graphical display of the raysums during data collection.

(5) Reconstruction of the image during the scanning process.

(6) Display of the partially reconstructed image.

(7) Filing of the projection and reconstructed image data on completion of the scan.
4.12.1 Operation of the Transmission Tomography Scanner.

To perform a transmission scan using the scanner, the SCA window is first set so that only events within the chosen full-energy peak from the detector are recorded by the counter-timer, as described above.

With the source temporarily removed from the holder, the object is placed upon the turntable, and the HeNe alignment laser is mounted in the source holder sleeve to ensure that the beam is concentric with the position of the source collimator aperture. The laser beam is then aimed into the detector collimator orifice by adjusting the position of the source holder using the screws attached to the optical bench riders. These enable fine adjustments to be performed quickly and accurately. With the beam pointing directly into the detector, maximum count-rate and minimum scanning time will be obtained. The laser is then removed and the collimated source is remounted in the sleeve.

With the source and detector aligned, the size of the linear motion of the scan is chosen, this value must just exceed the largest horizontal dimension of the object to allow 'through air' values to be obtained at the end of each projection. The raysum values through air are used to normalise the data, by determining \( I_0 \) in equation (3.2.4). This repeated sampling of air value reduces the effect of detector drift or source decay during a long scan. Care must be taken to ensure the attainment of air values when scanning highly asymmetric objects. The normal method of checking the correct scan amplitude is by moving the object through its full motion with the laser operating in place of the source, enabling the existence of ray paths through air to be observed visually. Alternatively, the source may be placed in the holder and the time for the final raysums of test projections may be checked against a pre-recorded air value taken with the object removed.
The scan parameters are determined by a compromise between the required spatial and density resolutions and the total scanning time. Spatial resolution is determined by the collimation, with step size normally being set equal to the detector collimation width. The number of projections is selected to be of the same order of the number of raysums as discussed in earlier in the chapter. To gain an estimate of the overall scanning time a quick trial projection may be carried out using a very low number of counts per raysum. The number of counts per raysum is actually determined by the required resolution in attenuation coefficient, this being limited by counting statistics.

The scan parameters are entered into the HP by way of a menu which requires the number of raysums, step size, number of projections, scan title, mode of scanning and scan reference number to be entered.

To correct for the effect of room background radiation, the background option allows the storage of a 'background raysum', obtained by counting with the collimator aperture blocked with a lead shield. The background count may then be subtracted from all raysums. This can be of importance when objects of high attenuation are being scanned, as in such cases the contribution of room background by reducing the estimate of attenuation coefficient leads to inaccurate image determination, especially at the image centre. If the background is principally caused by source photons 'leaking' through the detector collimator, a more accurate procedure is to take background 'projections' with the detector collimator completely blocked.

Two possible data collection modes are available using the SCANNER program, sequential and non-sequential. In sequential mode the angular steps are taken at regular increments equal to 180° divided by the number of projections. In non-sequential mode the object is rotated through large
angular intervals for the first few projections, subsequent projections are then performed to fill in the 'missing' angles. The non-sequential mode of scanning is normally preferred for two reasons:

(i) The image builds up rapidly in the early stages of reconstruction and a recognisable image is normally formed by about the 12th projection. This enables gross features to be seen, as well as indicating errors in the setting up of the scan, thus saving time.

(ii) The non-sequential reconstruction routine is less vulnerable to artifacts resulting from source decay or detector drift. This can be of particular value when performing very long scans.

The total scanning time for the two modes is the same. The non-sequential scanning process, originally devised by MacCuaig, is useful for any slow scanning system which can reconstruct and display the simultaneously with collecting data. The filtered back-projection method of image reconstruction enables the image to be built up projection by projection.

On completion of a scan, two files are created on the hard disk, the first is the image file containing the the reconstructed data. During the course of the research the program was modified to begin saving the image at the end of each projection in the final third of the scan so as to save the partially reconstructed image in case of a system crash. The second file is the scan data file containing the projection data. This file is stored in binary format by the HP.
4.13 Difficulties Encountered During Scanning.

One problem commonly encountered in small-scale fixed beam CT scanners is that of ensuring that the centre of rotation of the subject is at the centre of the projection. The usual method of positioning is by using a pin at the object centre and then aligning using a laser concentric with the source. The SCANNER program, however, uses a software technique to centre the projections. The routine is based on the method explained by Foster [1981], which shifts the projections so that their centre of gravity is at the centre of rotation. Without the centering routine, badly positioned objects would give rise to the 'ears' artifact illustrated by fig 4.13a.

Many of the scans displayed the 'corners artifact', in which the corners of the image plane having an anomalous attenuation coefficient. This results from the number of pixels across the diagonals of the image exceeding the number of raysums. However, if a clear ring of zero density air values can be observed around the subject in the image, a circular region of interest may be drawn so as to exclude the image corners for determining the pixel histograms. In some imaging routines the anomalous corner values are set to zero.

Several errors were discovered in the scanner program during the progress of this project. The program functioned correctly in non-sequential transmission mode, but failed to correctly image non-symmetric objects in sequential or emission modes. Non-circularly-symmetric objects could not be reconstructed correctly other than in the non-sequential transmission mode. Fig 4.13b shows an image of two circular objects after reconstruction using the faulty program. At first these difficulties were discovered in emission scanning and the errors were thought to be due to the centering routine. This routine was then removed from the program, but
Figure 4.13a: The 'Ears' Artifact.

Figure 4.13b: Incorrectly Reconstructed Image of Two Circular Objects.
the reconstruction errors persisted. It was subsequently discovered that the errors resulted from stepping through the wrong increment or back-projecting at the incorrect angle. These errors were eventually corrected and the system now performs correctly in both sequential and non-sequential modes.

For the first stage of the research, the scanner performed reliably. In the latter stages, however, the system became subject to frequent crashing. The scanner would stop in the midst of a projection, the data appeared correct and the failures occurred at random. Originally the reason for this 'silent death' failure was thought to be due to mains-voltage spikes, as other apparatus within the laboratory began to malfunction over the same period. A mains filter was purchased and initially enabled a scan of 8 days' duration to be performed over the Christmas vacation. The silent death crash later began to recur, however.

When the crash occurred the HP suspended operation, apparently awaiting a signal, either data from the counter-timer or a signal from the Digiplan motor interface. The counter-timer was ruled out as the source of the problem because it could be operated manually from front panel switches, which is not possible when under HP control. As a result the Digiplan motor interface was dispatched for maintenance and a replacement was kindly loaned by Time and Precision Ltd, Basingstoke. To check that the motor drives were not at fault, the translation/rotation motor drives were interchanged. The scanner then performed continuous scanning for a period of two weeks without failure. When the original Digiplan interface was replaced after servicing, a failure occurred during the rotation phase of a scan thus verifying the interface-rotation drive connections as the most likely source of failure. Close attention to the wiring enabled subsequent scanning to be performed reliably.

A further source of scan failure observed was non-rotation of the
turntable. This was caused by the stepper motor drive cable fouling on other parts of the apparatus. The effect of such a failure produces a characteristic set of ring artifacts, caused by the same projection being backprojected at all angles. Fig 4.13c shows this artifact encountered when scanning a perspex phantom drilled with holes from 0.5 to 10mm in diameter.

4.14 The Reconstruction Program.

A program called RECON was also written by MacCuaig, was employed to enable a data file to be reconstructed. Unfortunately the data saving routine in SCANNER did not file the data correctly in non-sequential mode leading to a considerable loss of experimental data prior to the error being discovered and corrected. The reconstruction algorithms are basically the same as those in SCANNER, and the images produced by the two programs were identical once the programs had been successfully 'debugged'. Reconstruction time for a 60 projection, 45 raysum image would be of the order of five minutes.
Figure 4.13c: Artifact Resulting from Rotation Failure.
4.15 Image Display.

An image display program in PASCAL was written by MacCuaig to operate on the HP. The program was based upon the CIA (Computer Image Analysis) package written by Foster for the University of Surrey PRIME computers. The DISPLAY program contains the following features.

**SHOW**
Displays the reconstructed image on a default window. A colourscale is the default setting.

**WIND**
Varies the 'window' i.e. the dynamic range in $\mu$ displayed on the greyscale.

**LINES**
Draws a line section through the image and plots it.

**CIR**
Draws a circular region of interest on the image.

**STATS**
Displays the image statistics (mean, s.d. and s.d. of the mean) within a given region of interest.

**PAPER**
Prints out a (low quality) grey level image on the HP inkjet printer.

**SIGMA**
Displays the image on a SIGMA 5674 terminal fitted with a COTRON monitor and Polaroid camera for rapid hard copy.

The program could display the image as a 'greyscale' in any of four colours, otherwise a multi-colour image could be displayed. Such images though aesthetically pleasing were somewhat difficult to interpret because of the non-systematic variation of colours with $\mu$.

The CIA program on PRIME could, in principle, be employed for image display. However it proved difficult to transfer data from the HP to the PRIME systems, in addition the SIGMA terminal gave distorted images and sometimes proved unreliable. Use of the PRIME systems became exceedingly
time-consuming with the withdrawal of funding by SERC of the 'system E' PRIME computer. In future it would be advantageous to develop a rapid data transfer system from HP to PRIME to permit both image reconstruction (using the highly versatile SHARK program), and image display using CIA, which includes several useful features such as three-dimensional plotting and projection stacking, which are not available on the HP.

 Transmission Tomography Experiments.

During the course of this research a wide range of objects were scanned using the transmission tomography technique and the apparatus described above. Some of the scanning work displays the versatility of the CT method.

4.16 Scanning Using $^{241}$Am (60 keV).

As outlined above the most commonly used isotope for transmission work was $^{241}$Am. The energy (59.54 keV) of the gamma rays was suitable for objects of 30-100mm in diameter with densities less than about 3 gcm$^{-3}$. 
4.16.1 Test Scanning.

To demonstrate the scanner, a 50mm square, 10mm deep carbon cuboid was examined. The results of the three scans are shown in table 4.16.1

<table>
<thead>
<tr>
<th>Raysums</th>
<th>Projections</th>
<th>CPR</th>
<th>Background Sub</th>
<th>μ/mm⁻¹</th>
<th>σ/mm⁻¹</th>
<th>σ/mm⁻¹</th>
</tr>
</thead>
<tbody>
<tr>
<td>90 x 1mm 60 x 3'</td>
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<td>No</td>
<td>0.030</td>
<td>0.006</td>
<td>0.00057</td>
<td></td>
</tr>
<tr>
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<td>Yes</td>
<td>0.031</td>
<td>0.006</td>
<td>0.00051</td>
<td></td>
</tr>
<tr>
<td>90 x 1mm 60 x 3'</td>
<td>1000</td>
<td>No</td>
<td>0.029</td>
<td>0.0017</td>
<td>0.00015</td>
<td></td>
</tr>
<tr>
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<td>1000</td>
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<td>0.00006</td>
<td></td>
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<td>0.0294</td>
<td>0.00063</td>
<td>0.00002</td>
<td></td>
</tr>
</tbody>
</table>

It can be noticed that the use of greater numbers of counts per raysum (CPR) leads to lower variance in the pixel values and to a lower uncertainty in the value of μ as shown by the reduction in the standard deviation of the mean (σμ). The inclusion of a background correction leads to a slight increase in the measured μ value. The results were obtained using the DISPLAY program, a circular region of interest being defined in the central portion of the square. Fig 4.16.1a and b show images of the square using 1000 and 10000 counts per raysum respectively. In each case a lines section has been drawn (bottom left) through the image and a pixel histogram (top left) is shown. In fig 4.16.1a the greater degree of noise on the image due to poorer statistics can be seen in the line section, this is also clearly demonstrated by the greater widths of the peaks in the pixel histograms.
Figure 4.16a: Scan of carbon square using 1000 counts per raysum.
Counts per Raysum.

Figure 4.16b: Scan of carbon square using 10000
4.17 Dynamic Gamma Ray Scanning.

In most gamma ray tomography applications the object is static, and single-beam gamma ray CT is too slow to scan all but the most glacial of processes. However, low speed gamma ray scanning can be used to obtain time-averaged density determinations of dynamic systems. So far the technique has been demonstrated in the study of gas fluidized beds [MacCuaig 1985]. A further application, also from the field of chemical engineering has been the study of particulate flow through a conical hopper.

This study was performed in association with T. Verghese of the University of Cambridge chemical engineering department. The aim of the study was to determine the degree of voidage in a range of sands as they flowed through an orifice in a conical hopper. The void percentage is inversely proportional to the attenuation coefficient, and the tomographic method has the ability to display any regions of high average voidage within the flowing sample. Without the use of computed tomography, measurements of voidage had to be performed indirectly by measuring small pressure changes within the flowing sand. The method of pressure determination is limited to fine sands, since with coarse material air can percolate too freely into the voids to sustain measurable pressure differences. The CT method is totally non-invasive and involved the minimum of modification to the original apparatus.

The original hopper was constructed of aluminium set in a square cross-section block of perspex. The hopper varied in diameter from about 120mm at the top to approximately 10mm at the exit orifice. An unfortunate feature of the original hopper design was the fact that the perspex block was secured together by four 5mm diameter brass screws. The screws caused a high degree of attenuation, and in preliminary tests these screws resulted
in an excessively long scanning time.

In order to overcome this disadvantage a new hopper was designed by Thomas Verghese; this design used four 10mm cylindrical perspex supports in place of the screwed perspex block arrangement. This simple modification reduced the scanning time to approximately half its previous value.

To enable flow to be maintained throughout the scanning run, it was necessary to continually refill the hopper with sand. A method of extracting the sand falling from the orifice and return it to a container also had to be designed. Some difficulty was incurred in devising a system capable of operating whilst the hopper moved and rotated on the scanner bed.

The eventual system involved a 25mm diameter plastic pipe about 1 metre in length being attached to a 50mm top diameter plastic funnel placed about 10mm below the exit orifice of the hopper. To facilitate rotation and translational motion, the usual optical bench system of the scanner was replaced by two smaller optical benches separated by 70mm. The detector assembly was placed on one bench adjacent to the scanner bed. The source was placed on a second bench separated from the scanner, allowing the exit pipe to move freely between the two benches. To prevent damage to the scanner by sand entry, the scanner was covered in thin PVC sheeting.

The continuous flow of sand was facilitated by the simple, if arduous technique of continually refilling the hopper by hand. Early tests demonstrated that this action could be performed at a speed exceeding the rate of flow of sand from the hopper. The necessity of refilling the hopper limited scanning time to approximately one hour due to the fatigue of the operator. The scanner is shown in operation in fig 4.17a.

Measurements of voidage were performed at several different heights, from the top of the hopper down to the exit orifice. It was necessary to determine density changes of the order of a few per cent.
Figure 4.17a: Scanning sand flowing through a hopper.
achievable in run times of 45-60 minutes using 600-1200 counts per raysum.

A major difficulty encountered during the scanning was that of exceedingly frequent scan crashes. The failure rate became extremely high when using flowing sand (one crash every few minutes), fortunately it was possible to restart the scanner by switching the mains supply to the Digiplan interface system off momentarily. One possible reason for the increase in failure rate of the scanner was the very large electric fields that resulted from the charges built up by the moving sand, which acted as a Van der Graaf generator. Sparks were frequently emitted from the aluminium hopper, indicating fields of several thousand volts per centimetre. In an effort to reduce the charge build-up, a series of earthing wires were connected to various parts of the apparatus, including the scanner bed, aluminium hopper and plastic piping. The reservoir bucket was placed on an aluminium plate which was earthed to the equipment rack. Using these precautions the failure rate was somewhat reduced, although it remained considerable. The problem of scanner crashing was subsequently solved by exchanging the motor drive connections to the Digiplan interface.

The second problem encountered during this work was a result of an error in the SCANNER program reconstruction algorithm, which incorrectly back-projected the projection data in the sequential scanning mode; this mode was necessitated by the limited rotation of the hopper emptying pipe. As a result of this error the perspex support pillars were incorrectly imaged, leading to a series of circular artifacts around the perimeter of the image as shown in fig 4.17b. The error was easily rectified, but the data files had not been filed in the correct format by the SCANNER program. A scan of the sample performed using the corrected program is shown in fig. 4.17c. The voidage fractions obtained using the corrected and uncorrected programs were consistent within experimental uncertainty.
Figure 4.17b: Uncorrected scan of hopper.

Figure 4.17c: Corrected scan of hopper.
The results of the scans clearly demonstrated the increase in voidage as the sand approached the orifice. The results were further checked by performing single transmission measurements at different heights on the hopper. The transmission method does provide a non-invasive measurement of density, but had to be performed in conjunction with a non-flowing hopper to compensate for any effects due to the different thicknesses of material through which the beam passes at different heights.

The tomographic scanning technique proved a useful way of studying voidage, and the results could be improved by performing more scans using the corrected reconstruction algorithm. The main disadvantage with the method was the long scanning time, which proved considerably more of a problem than usual due to the necessity of keeping the hopper full. X-ray CT would solve the problem of scanning time, although most medical CT machines scan with the object axis horizontal. In addition the beam hardening artifacts would make it difficult to determine accurately the presence of voids in the outer parts of the object. A multi-detector fan beam gamma ray system which would allow scanning in a few minutes without artifacts would be ideal for research applications. A 16 beam system is to be commissioned at the University of Surrey in the near future.

4.18 Scanning of the Gas Filter 'Candle'.

An object of interest to the University of Surrey chemical engineering department was a 'candle', a filtration device constructed from very low density silicon oxide ceramic fibres. The candle structure has a very fine pore size, so that the it may be used for the filtration of very fine dusts. The device is operated at temperatures up to 1000°C, filtration being
performed from the outside in. On completion of the filtration cycle, the dust caking is removed by reverse flowing pure gas through the system. The outer surface of the candle has been impregnated after foaming to prevent ingress of fine dust into the filter.

Scanning was performed using 1 mm collimation and 60 projections and 2000 counts per raysum. The results of the scan are shown in fig 4.18a, which clearly displays higher density at the outer parts of the device. A line section is shown in figure 4.18b, which again clearly shows the change in density of the material at the outer edge. This effect was believed to be a result of non-uniform impregnation of the outside of the object.

Three concentric regions of interest were drawn within the candle, the results are shown in table 4.18. Although the 60 keV energy was somewhat higher than the optimum for scanning the low density ceramic, density variations were easily shown using a 24 hour scan time.

<table>
<thead>
<tr>
<th>Region of Interest</th>
<th>No Points</th>
<th>Mean $\mu$ mm$^{-1}$</th>
<th>$\sigma$ mm$^{-1}$</th>
<th>$\sigma$(mean) mm$^{-1}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>1 (Outer)</td>
<td>206</td>
<td>0.0107</td>
<td>0.0024</td>
<td>0.00017</td>
</tr>
<tr>
<td>2 (Middle)</td>
<td>178</td>
<td>0.0071</td>
<td>0.0031</td>
<td>0.00029</td>
</tr>
<tr>
<td>3 (Inner)</td>
<td>155</td>
<td>0.0049</td>
<td>0.0047</td>
<td>0.00038</td>
</tr>
</tbody>
</table>

4.19 Rock Core Scanning.

Analysis of rock-core samples obtained by drilling into prospective oil-bearing rock is of major importance in the oil recovery industry. The oil content is of primary commercial interest, as are the flow processes which occur within oil reservoirs. Tomography provides a useful method of
Figure 4.18a: Scan of gas filter 'candle'.

Figure 4.18b: Line section through above image.
non-destructive core analysis, which is capable of measuring oil or water content and provides a valuable method of studying the flow of fluids within rock cores.

X-ray computed tomography has been used for several years to analyse rock cores and has given qualitatively good images (Wang 1984, Hunt 1988). The drawbacks of using conventional X-ray tubes in terms of beam hardening artifacts, and lack of penetration have been observed. As a result: the X-ray based systems are not always suitable for giving accurate quantitative results. The use of monoenergetic gamma rays allows absolute measurements to be performed (Nicholls 1984), although requiring much longer scanning times.

A dry rock-core was obtained from BP Sunbury Research Centre, this was imaged using 65 mm steps, 60 projections and 1200 counts per raysum. The core was then water saturated and rescanned. Fig.4.19 shows the saturated (top) and unsaturated rock cores. The increase in attenuation coefficient is clearly shown by the pixel histograms (left), the water surrounding the saturated core is also clearly visible. Using the various values of attenuation coefficient in the two images, the degree of water saturation could be determined.

Scanning time was between 12 and 16 hours, which is considerably too long for routine scanning, but clearly demonstrates the ability of gamma ray scanning to measure saturation density. The gamma ray scanner may also be used in order to check results of faster X-ray CT scans in situations where it is hard to distinguish real structure from artifacts. A low-cost multi-beam gamma ray scanner would prove a useful supplement to an X-ray CT device, especially if a range of gamma ray sources were available to permit the system to study objects too dense to be imaged using X-ray CT machines.
Figure 4.19: Scans of water (upper) and oil (lower) saturated rock cores.
4.20 Scanning Using $^{192}$Ir.

A 37 GBq $^{192}$Ir source was made briefly available by SMIS Ltd, and it was employed to study the ability of the gamma ray CT technique to determine small density changes in low Z objects using medium energy gamma rays.

The objects scanned were an apple, an orange and a 'Marathon' chocolate bar figs 4.20a, b and c respectively. All these objects have attenuation values close to that of those of water. The scanning energy chosen from this mult-energy emitter was 300 keV, at which energy $\mu_{\text{water}}$ is $0.0119 \text{ mm}^{-1}$ so that the objects were approximately 0.8 mean free paths in thickness, which is considerably lower than the optimum given in section 4.5. Despite of this very small changes in attenuation coefficient can be clearly determined. This was possible as an unusually large number of counts (100000) was taken per raysum allowing very low statistical uncertainties on pixel values to be achieved. The brightness of the source enabled scans with a resolution of 0.8 mm to be performed in times of 16 hours or less.
Figure 4.20a: Scan of apple using $^{122}$Ir.

Figure 4.20b: Scan of orange using $^{122}$Ir.
Figure 4.20c: Scan of 'Marathon' bar using $^{192}$Ir.
4.21 Examination of a Reinforced Concrete Test Sample

Using High Energy Gamma Ray Tomography.

The ability to determine the presence and position of crack defects in reinforced concrete is of particular interest in many civil engineering applications where cracking could lead to a significant loss in structural strength. High energy gamma ray tomography offers a potential non-destructive solution to the study of large scale concrete pillars.

ERA Technology requested a series of experimental trials to determine the capability of the CT technique using the University of Surrey scanner.

In 1989 ERA Technology delivered a test sample to the University of Surrey. The sample simulated a section of concrete pillar, and consisted of a squat cylinder of 300 mm diameter and 50 mm height. Within the sample a number of 10 mm diameter steel "reinforcing rods" had been placed. To simulate the presence of radial cracks, a number rectangular sections of expanded polystyrene and polythene sheeting with thicknesses between 3 mm and 0.1 mm had been positioned. The number and position of the reinforcing pins and simulated cracks was not disclosed prior to scanning.

The aims of the experimental work were as follows:

(i) To determine the optimum source for CT studies of large concrete samples.

(ii) To obtain estimates of the scanning time required to determine 0.1 mm cracks with available sources.

(iii) To indicate some of the difficulties which could arise in practice, especially in terms of source shielding requirements.
4.21.1 Theoretical Considerations

The direct imaging of 0.1mm concrete cracks in 16 hours (the preferred upper time limit) using sources of available activity is impractical using a single-beam system. The technique of gamma ray tomography is, however, capable of detecting the presence of objects smaller than the spatial resolution.

Consider a uniform sample with attenuation coefficient $\mu$, in which is placed a small object of volume $v$ and attenuation coefficient $\mu'$. If the object is then scanned tomographically using a raysum width $t$ and thickness $d$, then the sample may divided into volume elements of volume $V$, where $V = d \times t$. The measured attenuation coefficient $\mu_m$ of the voxel containing the object will be given by:

$$\mu_m = \frac{\mu'v + \mu(V-v)}{V} \quad 4.21$$

The presence of the object can just be detected when the difference in attenuation coefficient, $\mu - \mu_m$ exceeds the statistical variation between voxels. The presence of voids or cracks is equivalent to the existence of objects of effectively zero attenuation within the sample.

An estimate of the statistical uncertainty associated with a scanning time of 16 hours was be obtained using equation 4.66 applied to the concrete sample. If a step size of 2mm is used, then a 0.1mm crack will produce a 5 per cent reduction in attenuation coefficient if the vertical extent of the crack exceeds the slice thickness.

In the calculation the following values were used:

- Number of pixels 165². Detector photopeak efficiency 6%.
- Source activity 7.4GBq. $\mu_{1.25}$ = 0.078 cm⁻¹ (at 1.25 MeV).
This predicted an uncertainty in $\mu$ of around 8%, which indicates that detection of cracks of 0.2 mm or greater would be possible.

4.21.2 Experimental Work.

To facilitate scanning of the concrete sample, a 310 mm aluminium turntable mounting was constructed by the departmental workshops. The 50mm x 50mm NaI(Tl) detector was mounted in its usual brass and lead collimator holder. The first source available was a 56 MBq (1.5 mCi) needle of $^{137}$Cs, emitting 662 keV radiation. The usual detector collimator was found to be an inadequate shield against 662 keV gamma rays (allowing about 20% transmission) and as a result an extra 15 mm of lead was placed on the front of the detector to reduce the transmission to about 3.5%.

The needle source was placed in a 100 mm diameter lead holder, which had been machined with a 2 mm wide, 20 mm deep and 100mm long collimator to produce a ribbon beam. The detector collimator was 2 mm wide by 7 mm deep. An initial scan was performed using this low-activity source; 90 projections, each comprising of 164 2mm steps, were recorded with 100 counts per raysum. This scan produced a noisy image in about 36 hours. The central reinforcing pins were clearly visible but the large amount of image noise made it difficult to see the simulated cracks of less than 2 mm in width.

An eight day scan was subsequently performed over the Christmas vacation. The scanning parameters were as above except that the number of counts per raysum was increased to 500. The resulting image is shown in figure 4.21a top left. The 12 large 'cracks' (expanded polystyrene) are clearly visible and about half of the thinner (0.2mm) polythene sheet 'cracks' can be discerned. The thinnest (0.1mm) sheets cannot be seen with any degree of certainty. The central region of the image displays some
Figure 4.21a: Scans of reinforced concrete sample using $^{157}$Cs and $^{60}$Co sources.
structure, but this was thought to be an artifact due to the low penetration of the 662 keV photons through the concrete. Using the data of Hubbell [1969], the narrow-beam penetration of 662 keV through 300mm concrete was calculated to be 0.31% falling to 0.15% when the beam traverses two reinforcing rods. This latter value exceeds the optimum thickness for CT imaging as defined in section 3.5, which requires minimum transmissions in the range 0.25 - 36%.

These initial scans proved useful, but showed that a source of much higher activity would be required for practical scanning in periods of less than 16 hours. In addition a source energy higher than 662 keV is desirable.

4.21.3 Work with the $^{60}$Co Source

To perform scans of the concrete sample in a more practical time period, a 7.4 GBq (200mCi) $^{60}$Co source was employed. As well as having much higher activity than the $^{137}$Cs source used previously, the maximum attenuation through the sample using gamma rays with a 1.25 MeV mean energy was 0.79%. The disadvantage with the $^{60}$Co was the shielding problems which it presented. The air kerma rate from the unshielded source was 2.27 mGy/hr at 1 metre.

A new shield/collimator assembly was designed for this source. It consisted principally of a 250mm long lead cylinder, 115mm in diameter. This size was lower than the optimum, but was selected due to its ready availability and because the collimator holder had to be capable of being moved by one person in the laboratory. The cylinder was machined in two pieces which were designed to slide together in a vee shaped bed. The front part of the cylinder was designed to hold both the $^{60}$Co source (Amersham catalogue number CKC25) and any one of a series of cylindrical copper
collimators. Copper was chosen as it has fairly high density (and therefore attenuation coefficient) whilst being stiffer and easier to machine than lead. The collimators had a total length of 140 mm, consisting of a 25 mm diameter section, 115 mm in length with a 37 mm diameter end-piece, which projected from the front of the lead holder to facilitate easy collimator changing using 0.5 m handling tools. The collimators were drilled to give exit beam diameters of 1, 1.5, 2, 2.5 and 5 mm. Due to the difficulties of boring rather long, small diameter holes, the four narrower collimators were drilled to 3 mm diameter down the main section. The end piece was drilled to the required collimation diameter.

As a result of using the high energy of the $^{60}$Co gamma radiation, an improved shield/collimator was also designed for the NaI(Tl) detector. It consisted of a holder machined from the lead billet described above. The front section was designed to accept copper collimators, of similar design to those above, except with a total length of 125 mm. The rear part of the holder was designed to accept the NaI detector in its usual lead/brass mounting. Both the source and detector holders were mounted on vee-shaped beds constructed of aluminium, based with steel. These items were mounted on the standard optical bench system on the scanner bed to enable accurate beam alignment. A photograph of the holders and collimators is shown in fig 4.21b.

The attenuation coefficient of copper at 1.25 MeV is 0.47 cm$^{-1}$; thus the source collimator was 6.6 mean free paths leading to 99.86% attenuation in the source collimator with a further 99.7% attenuation in the detector collimator. A high degree of shielding was required due to the relatively high count-rate variations encountered with the beam passing through the steel reinforced region of the sample.

Prior to mounting the 200 mCi source, a preliminary investigation was made using a lower activity (0.8 mCi) $^{60}$Co source. The use of this source
Figure 4.21b: Holders and collimators for COCo source.
enabled the shielding to be tested and the source and detector collimators to be accurately aligned without incurring unnecessary radiation dose. The count-rate was measured using various collimator combinations to check that dead-time problems would not occur using the 200mCi source. Using the 2mm collimators, a full energy peak count-rate of 15 per second through air was observed. This corresponds to a count rate of less than 4000 per second using the high activity source with the beam passing through air, resulting in dead-time losses of below 1%.

The operator shielding provided by the source holder was considered inadequate for exposure times longer than about a minute, and thus a 200mm thick wall of lead bricks was constructed around the apparatus. A lead brick wall 50mm thick was also constructed behind the detector to ensure that the radiation dose rates in adjacent workshops would remain below prescribed safety limits.

The final dose rate at the computer console was measured to be 2.2 μSv/hr, which is well below the 7.5μSv/hr limit for undesignated laboratory workers.

### 4.21.4 Scanning Using the 200 mCi 60Co Source.

With the source in place the first scan of the concrete sample was performed. The aim was to scan the object in a time of eight hours. Extrapolation from a single trial projection indicated that 169 2mm steps and 90 projections could be performed in this time using 500 counts per raysum. The image from this scan is shown in fig 4.21a top right. A second scan was subsequently performed using 1000 counts per raysum, the image obtained is shown in fig 4.21a bottom left.

Examination of these images indicated that a certain amount of streaking was visible, this showed up especially clearly within the corner
artifacts. This streaking was due to underdetermination of the image as a result of using too few projections. The centre of the image also shows quite good uniformity, verifying that the structure visible in the scan performed using $^{24}C$ was an artifact.

Since the simulated cracks were placed radially within the concrete sample, the effect of radial artifacts was highly undesirable. As a result the scan was repeated using 180 projections. Due to the limitations of available time only 250 counts per raysum were taken, giving an overall scanning time of 8 hours. Nevertheless the image quality is quite good (fig 4.21a bottom right), the radial streaking artifacts have disappeared and most of the 0.2 mm 'cracks' could be discerned.

It should be noted that the smaller simulated cracks contain polythene rather than air, producing a smaller reduction in attenuation coefficient than a true void. The 0.2 mm 'cracks' produce a 6% lowering in $\mu$ rather than the 10% change for real voids.

4.21.5 Conclusions

The technique of high energy gamma ray tomography is well suited to the detection of small (up to 0.1mm) cracks in concrete samples, where all-round access to the sample is possible. The experimental work carried out was able to detect 0.2 mm 'cracks' in a period of 8 hours using a 7.4 GBq $^{60}Co$ source. A high activity source is essential to obtain adequate counting statistics in reasonable run times. For samples of 30cm diameter or greater a source with gamma ray energy above 1 MeV is required to obtain adequate penetration of the sample.

The results indicated a scanning time of about 16 hours would be required in order to detect 0.1 mm cracks. Whilst this period is rather
long for many applications, the use of a higher activity source and/or multiple detectors could considerably reduce scanning time. Using a ribbon beam to scan a larger slice thickness would result in increased counting rate; this technique would be of value if the cracks were expected to have considerable extent along the axis of the sample.

In practice the massive shielding required when using intense, high energy gamma ray sources results in some inconvenience; however the use of a uranium shield would give a considerable reduction in shielding bulk. The design of a tomographic system for in-situ scanning of reinforced concrete structures would have to include a substantial system for moving the heavy source-detector assembly around the sample under investigation. The system must be able to deal with samples of different diameters and possibly non-circular geometry. The engineering problems presented should prove soluble without excessive cost for sample diameters up to ~400 mm.
Chapter 5.

Gamma Ray Scattering in Materials Analysis.

Detection of scattered optical radiation is the basic principle behind that most useful of imaging devices, the human eye. In contrast, conventional X- and gamma-radiography rely on detection of transmitted radiation to form an image. In recent years, however, the scattering of gamma radiation has been usefully employed in the interrogation of matter.

As outlined in chapter 2, gamma radiation can be scattered by either a coherent (Rayleigh) or incoherent (Compton) process. Either process may be used to obtain information about objects of interest, but to date the Compton effect has found the greatest amount of use. Compton scattering is the dominant mode of interaction between gamma rays and matter over a wide range of the energies encountered in bulk materials analysis. As a result the study of the Compton scattered radiation provides an interesting alternative to transmission techniques as a means of interrogating material.

The principal advantages of the Compton scattering technique are:

(i) Access is only required to one side of the object.

(ii) The technique is sensitive to electron density and not to elemental composition. In some applications this feature of the Compton technique is particularly valuable.
The dependence of the Compton scattering intensity on density can be shown as follows:

Consider a small volume \( V \) of material of atomic number \( Z \), with \( N \) atoms per unit volume, under irradiation by a narrow beam of gamma radiation with intensity \( I_0 \). The Compton scattered intensity, \( I_{\text{Com}} \), at angle \( \theta \) to the incident beam is given by:

\[
I_{\text{Com}} \propto I_0 N Z V \frac{d\sigma_{\text{Com}}(\theta)}{d\Omega}
\]  

(5.1)

where \( \frac{d\sigma}{d\Omega}_{\text{Com}}(\theta) \) is the differential Compton cross-section per atom and \( N Z \) equals \( \rho_e \), the electron density. Equation 5.1 may be rewritten as:

\[
I_{\text{Com}} \propto I_0 N_{\text{Av}} Z \rho V \frac{d\sigma_{\text{Com}}(\theta)}{d\Omega}
\]  

(5.2)

where \( N_{\text{Av}} \) is Avogadro's number, \( A \) the atomic mass and \( \rho \) the mass density. Assuming \( Z/A \) is constant (approximately 0.5 for elements other than hydrogen), then \( I_{\text{Com}} \) is proportional to \( \rho \).

This chapter briefly reviews the applications of gamma ray scattering in diagnostic medicine and non-destructive evaluation. The sections include materials gauging using Compton backscattering, the determination of bone mineral density using scattering methods and the techniques and applications of Compton scatter imaging including some experimental demonstrations. It concludes with a brief outline of the potential uses of coherent scattering in imaging.
5.1 Industrial Gauging Using Compton Backscattering.

The technique of Compton backscatter gauging has been employed for some time in a range of applications. One of the earliest uses of backscatter gauging was in determining soil density [Cameron 1958]. The basic principle of the technique is straightforward: a beam of gamma rays from a collimated radioisotope source (usually $^{137}$Cs) is directed into the soil under investigation and the backscattered intensity is recorded at a detector which is well shielded from the primary radiation emitted from the gamma source.

Early gauges did not employ detector collimation or energy discrimination and as a result the relationship between detected count-rate and soil density was somewhat ambiguous [Smith 1968, Taylor 1972]. Detected counts resulted from background or from scattered photons with varied histories; for example multiply scattered photons contributed a significant fraction of the recorded counts. Some multiply scattered photons are of quite low energy (<100 keV) with the result that the photoelectric absorption process becomes important, leading to Z dependent count-rate effects. Improved instrumentation using scintillator detectors with their output discriminated to permit only photons within a narrow energy range has since been used, with a result that the dependence of the output signal on elemental composition is reduced [Taylor 1972].

A simplified expression for the backscattered intensity at a small detector may be obtained as follows. Consider a narrow beam of monoenergetic gamma rays of energy $E$, incident upon a sample of material of thickness $t$, density $\rho$, and with $N$ atoms per unit length. A collimated detector is placed so as to detect gamma rays scattered through an angle $\theta$, as in figure 5.1.
Figure 5.1: Principle of Backscatter Gauging.
The number of unscattered photons per second reaching a length element \( \delta x \) of unit area at a depth \( x \) is given by:

\[
I = I_0 \exp\left(-\frac{\mu}{\rho} \delta x \right) \tag{5.1.1}
\]

where \( I_0 \) is the primary photon flux incident upon the sample and \( \mu/\rho \) is the mass attenuation coefficient of the material at energy \( E \). The intensity of singly Compton scattered photons at an angle \( \theta \) due to \( \delta x \) is given by:

\[
\delta I^s(\theta) = I_0 N.Z \sigma(\theta) \exp\left(-\frac{\mu}{\rho} \delta x \right) \tag{5.1.2}
\]

where \( \sigma(\theta) \) is the differential Compton scattering cross-section at an incident energy \( E_0 \) and an angle \( \theta \). The scattered radiation will be further attenuated in its passage through the sample, and thus the flux at a detector \( D \) is given by:

\[
\delta I^s(\theta) = I_0 N.Z \sigma(\theta) \exp\left[\left(-\frac{\mu}{\rho} - \frac{\mu'}{\rho'}\right) \delta x \right] \exp\left(-\frac{\mu'}{\rho} \delta y \right) \delta x \tag{5.1.3}
\]

where \( \mu'/\rho' \) is the mass attenuation coefficient at the scattered energy \( E' \). As \( \theta \) tends to 180°, \( y \) approaches \( x \) and hence the backscattered flux is given by:

\[
I' = I_0 N.Z \sigma(180°) \int_0^{180°} \exp\left[\left(-\frac{\mu}{\rho} + \frac{\mu'}{\rho'}\right) \delta x \right] \, d\theta \tag{5.1.4}
\]

This leads to:

\[
I' = I_0 N.Z \sigma(180°) \left(1 - \exp\left[-\left(\frac{\mu}{\rho} + \frac{\mu'}{\rho'}\right) \rho t\right]\right) \tag{5.1.5}
\]
In most commonly encountered materials the theory is complicated by the presence of several elements, with the result that (5.14) becomes:

\[
I' = \frac{I_0 \sum_i N_i Z_i \sigma(180')_i (1 - \exp(-\sum_j [(\mu/\rho)_j + (\mu'/\rho)_j]pt_i W_j))}{\sum_j (\mu/\rho)_j + (\mu'/\rho)_j} W_j
\]  

(5.1.6)

the summation being performed over all elements and compounds, \( W_j \) is the fraction by weight of each compound. The backscattered intensity rises to a saturation value given by:

\[
I'_{\text{sat}} = \frac{I_0 \sum_i N_i Z_i \sigma(180')_i}{\sum_j (\mu/\rho)_j + (\mu'/\rho)_j} W_j
\]  

(5.1.7)

Once saturation is achieved, then the backscatter intensity is a function only of the mass attenuation coefficients and macroscopic differential scattering cross-sections of the sample material. At intermediate photon energies the dominant attenuation process is Compton scattering, so the detected flux at saturation is almost exclusively dependent on the macroscopic Compton scattering cross-section and thus on the electron density of the sample.

For samples of constant density and composition, Compton backscattering can be used for thickness gauging so long as the sample thickness is below that giving rise to saturation [Kato 1976]. Compton scatter gauges are most commonly used to determine sample density with objects beyond the saturation thickness, such as soil density measurement [Ertek 1984], rock density and water/oil saturation in bore-logging [Cameron 1971].
5.2 Gamma Ray Scattering in Bone Densitometry.

5.2.1 Introduction.

In recent years increasing interest has been shown in the subject of osteoporosis as an important health problem. Serious loss of bone mineral is the cause of considerable suffering in terms of fracture injury and the subsequent disability and deformity. The most common site of injury is the femoral neck. In 1981 the cost in femoral neck fractures to the British N.H.S. was estimated at some £48 million, involving the occupation of about 18% of all orthopaedic beds in the U.K. [Woolf 1987].

The principal sufferers of osteoporosis are post-menopausal women, who may lose 2-3% of their bone mass per year immediately following menopause [Aitken 1984]. Losses of bone mass of over approximately 20% are accompanied by significant increase in fracture probability. Steady bone loss occurs in both men and women as they advance beyond middle age, and as a result the topic is becoming of increasing interest to our ageing population.

Clearly the prevention of osteoporosis is highly desirable. In order to provide effective treatment, it is, however, first necessary to identify those members of the population affected. Several methods of doing this have been attempted, and many of these techniques have involved the use of X or γ radiation. A simple method of attempting to determine bone mineral density has been to use radiography [Aitken 1984], but such methods are somewhat subjective and their validity has been called into question [Mazess 1983a]. As a result a considerable amount of research has been carried out into more accurate non-invasive forms of bone mineral measurement.

The technique of dual-photon absorptiometry is becoming increasingly
accepted as a diagnostic tool; the principle is to measure the degree of
gamma-ray attenuation through bone to determine the mineral density. The
use of two energies is required to allow corrections to be made for the
effects of soft tissue covering the bone.

The most popular source for the dual photon absorptiometry is
$^{153}$Gd [Smith 1983] and this has been employed in several commercial
densitometers, eg. the Norland ND2600. A precision of about 3% has been
achieved using the dual photon absorptiometry technique [Le Blanc 1986].

Bone density may be determined from tomographic images, and where X-
ray CT scanners are available they can be usefully applied as diagnostic
tools for detecting osteoporosis, but the effects of beam polychromaticity
and scattering reduce the accuracy of the method. Despite this, successful
in vivo work has been performed using CT to obtain a precision of about
2.2% for a surface radiation dose of 16mGy [Banks 1986].

The CT technique has the advantage of producing images, which enable
the trabecular bone to be distinguished from the outer cortex, in addition
the appearance of the trabecular bone may give indications of its
structural strength. Partly as a result of their widespread availability,
CT scanners have been widely used to study osteoporosis, based on
measurements performed on vertebral trabeculae, although other structures
have also been scanned.

The chief disadvantage of the X-ray CT scanner for densitometry
studies is the effect of beam hardening, which may lead to systematic
errors in measurement of a few percent [Mazess 1983b]. This difficulty may
be reduced by performing the CT measurement at two kVp settings [Hawkes
1986], the results of the bone scans be compared with calibration scans of
bone equivalent material at both energy settings. The dual energy technique
does, however suffer from two major disadvantages: increased patient dose and longer scanning times, with increased errors due to patient movement.

A considerable reduction in patient dose can be obtained by using a radioisotope source in place of the X-ray tube. The use of gamma radiation also removes the errors which result from beam hardening and permits the use of much cheaper apparatus. Early studies such as those of Ruegsegger et al [1976], utilised an $^{125}$I (29 keV) source to study the distal radius. The low count-rate achieved with radioisotopes precludes the use of technique for the spine. Research with the technique still continues [Ruegsegger 1987], although the method has not proved popular.

Gamma ray scattering has also been proposed as an alternative technique, and several different methods are described below.

5.2.2 Bone Densitometry using Compton Scattering.

The use of Compton scattering as a means of evaluating bone mineral density was first suggested by Garnett et al [1973], the basis of the technique is to use the intensity of the Compton scattered radiation to give the electron density of the bone (see sections 5.1 and 5.2.1 above), which is proportional to its mass density.

The electron density $\rho_{e1}$ is given by:

$$\rho_{e1} = \frac{Z}{A} \cdot N_A \cdot \rho$$  \hspace{1cm} (5.2.1)

where $N_A$ is Avagadro's number. Since the value of $Z/A$ is roughly constant, then the number of scattered photons detected at a given angle will depend solely on the mass density of the material. For all elements other than
hydrogen, Z/A is approximately 0.5, and therefore, assuming the hydrogen content of bone does not vary widely, then the bone density may be determined by comparing the scattered count-rates with that from a sample of known density:

$$\rho_{\text{bone}} = \frac{(Z/A)_{\text{ref}} \cdot S_{\text{bone}} \cdot \rho_{\text{ref}}}{(Z/A)_{\text{bone}} \cdot S_{\text{ref}}} \quad (5.2.2).$$

If we use water as the reference material, then (Z/A)$_{\text{water}} = 0.556$ and (Z/A)$_{\text{bone}} = 0.537$ and the difference in (Z/A) between water and bone is only 0.35%.

In general fairly low photon energies (<100keV) have been employed in bone densitometry so as to give a reasonably high interaction cross-section and low patient dose. At such energies the beam attenuation within surrounding tissue is significant, and it necessary to correct for the attenuation of both the incident and scattered beam. The attenuation correction is complicated by the difference in energy between the primary and scattered radiation. One method of correction is to employ two radiation sources with energies chosen such that the energy of the second is equal to the energy of the scattered photons from the first [Webber 1976].

The technique is illustrated by figure 5.2.2, where the primary source photon energy is $E$ and the secondary source photon energy is $E_c$, given by the Compton scattering energy formula:

$$E_c = \frac{E}{1 + (E/m_c^2)(1 - \cos \theta)} \quad (5.2.3).$$

Firstly a measurement is made of the scattered radiation flux $S$, along $AB$.
Figure 5.2.2 Compton scattering densitometry.

(attenuation correction)
where $\mu$ and $\mu'$ are the linear attenuation coefficients at $E$ and $E'$ respectively. The apparatus is then rotated through $180^\circ$ and the number of scattered photons $S_z$ is given by:

$$S_z \propto \rho_e \cdot \exp\left(-\int_0^\beta \mu \, dx\right) \cdot \exp\left(-\int_0^\beta \mu' \, dx'\right) \tag{5.2.5}$$

Taking the product $S_1 \cdot S_2$ gives:

$$S_1 S_z \propto \rho_e^2 \cdot \exp\left(-\int_0^\beta \mu \, dx\right) \cdot \exp\left(-\int_0^\beta \mu' \, dx'\right) \cdot \exp\left(-\int_0^\beta \mu' \, dx'\right) \cdot \exp\left(-\int_0^\beta \mu' \, dx'\right) \tag{5.2.6}$$

Simplifying gives:

$$S_1 S_z \propto \rho_e^2 \cdot \exp\left(-\int_0^\beta \mu \, dx\right) \cdot \exp\left(-\int_0^\beta \mu' \, dx'\right) \tag{5.2.7}$$

Performing transmission measurements along $AA'(T_1)$ and $BB' (T_2)$ at $E$ and $E'$ respectively gives:

$$T_1 T_2 \propto \exp\left(-\int_0^\beta \mu \, dx\right) \cdot \exp\left(-\int_0^\beta \mu' \, dx'\right) \tag{5.2.8}$$

Thus:

$$\rho_e \propto \left[\frac{S_1 S_z}{T_1 T_2}\right]^{0.5} \tag{5.2.9}$$

The constant of proportionality for a given experimental set-up may be determined by using a standard material, usually water.

Webber and Kennet [1976] employed a Compton scattering densitometer
for in vivo measurements using $^{153}$Sm (103 keV) as the primary source and a
$^{170}$Tm (87 keV) secondary source with a scattering angle of 90°. An accuracy
of 22 kg m$^{-3}$ (~2%) for a soft tissue dose of 1.6 mSv was quoted. Kennet
[1976] discussed the sources of systematic error involved, including non-
identical geometry (i.e., the scattering volume is larger than that the
attenuating volume involved during transmission measurements) and finite
beam widths. The uncertainties produced were, of the order of 1%. A more
serious source of error is multiple scattering, which could cause
overestimates of bone density by up to 10%. Huddleston et al [1979, 1986]
suggested that empirical corrections could be usefully applied to reduce
the effect of multiple scattering, and that the use smaller angles
scattering make the effects of multiple scattering less serious.

5.2.3 Coherent and Compton Scattering Ratio.

When the spectrum of scattered radiation from a sample is studied
using a detector with high energy resolution, it can be seen that some of
photons are scattered by an elastic (coherent) process. Puumalainen et al
[1976] suggested that examining the ratio of coherent/Compton scattered
photons could be used to give useful information on bone mineral density.

The experimental arrangement for the coherent/Compton method is shown
schematically in fig. 5.2.3. The technique is insensitive to the presence
of overlying tissue since all photons will have traversed the same path,
assuming the variation in attenuation coefficient between elastically and
inelastically (Compton) scattered photons is not too great. The need for
good energy resolution entails the use of a hyperpure germanium (HPGe) or a
lithium drifted germanium (Ge(Li)) detector. The theory of the method is
outlined below.
Figure 5.2.3 Coherent/Compton densitometry.

(90 degrees)
Consider a narrow beam of gamma rays with photon energy $E$ and intensity $I_0$ incident upon an object under investigation. If the scattering volume $V$ contains material with $N$ atoms per unit volume, the coherently scattered intensity $S_{coh}$ is given by:

$$S_{coh} = I_0 \exp(-\int \mu(x) dx) \cdot N \cdot V \cdot \frac{d\sigma}{d\Omega}_{coh}(\theta) \cdot \exp(-\int \mu'(x) dx) \tag{5.2.10}$$

and the Compton scattered intensity is given by:

$$S_{com} = I_0 \exp(-\int \mu(x) dx) \cdot N \cdot V \cdot \frac{d\sigma}{d\Omega}_{com}(\theta) \cdot \exp(-\int \mu'(x) dx) \tag{5.2.11}.$$ 

If the value of the attenuation coefficient $\mu$ at the incident energy $E$ is approximately equal to the attenuation coefficient $\mu'$ at the Compton scattering energy $E'$, then $R_{cc}$, the ratio of coherent to Compton scattered photons is simply given by:

$$R_{cc} = \frac{S_{coh}}{S_{com}} = \frac{(d\sigma/d\Omega)_{coh}}{(d\sigma/d\Omega)_{com}} \tag{5.2.12}.$$ 

The linear attenuation coefficient $\mu_{scatt}$ due to scattering in a material of mass density $\rho$ and atomic mass $A$ is given by:

$$\mu_{scatt} = \rho \cdot (N_A/A) \cdot \sigma_{scatt} \tag{5.2.13}$$

where $N_A$ is Avogadro's number. For Compton scattering the cross-section is proportional to the atomic number and hence:

$$\mu_{com} \propto N_A (Z/A) \cdot \rho \quad \text{i.e. } \mu_{com} \propto \rho \tag{5.2.14}$$
The coherent scattering cross-section varies approximately with $Z^{2.5}$, and thus:

$$\mu_{\text{coh}} \propto N_A (Z^{2.5}/A) \rho \quad \text{i.e.} \quad \mu_{\text{coh}} \propto \rho Z^{1.5} \quad (5.2.15)$$

Thus since the coherent scattering differential cross-section varies with $Z^{2.5}$ and the Compton differential cross-section varies with $Z$, then their ratio $R_{cc}$ is a function of $Z^{1.5}$. Osteopenic bone contains less phosphorus and calcium than healthy bone and hence low values of $R$ indicates a loss in bone mineral.

A number of authors have carried out in vitro studies to determine the diagnostic potential of the coherent/Compton ratio technique. Kerr et al [1980] used a $^{153}$Sm source (103 keV) and a 27° scattering angle; a small scattering angle was used in an attempt to maximise the coherent count-rate as the coherent scattering differential cross-section is strongly forward peaked. The disadvantage of using small scattering angles, however, is that it becomes increasingly difficult to resolve the coherent and Compton scattered signals, and in practice curve-fitting computer routines are required. Gigante and Sciuti [1985] used a 135° angle in order to gain maximum sensitivity but at such large angles the coherent differential cross-section is very small leading to a very low coherent flux. In addition, large angle scattering gives rise to a substantial difference in the Compton and coherent scattered energies and hence the attenuation coefficients $\mu$ and $\mu'$ may differ by several percent (8% in muscle, 25% in bone at 60/50 keV), which tends to reduce the validity of expression 5.2.11. Long counting times and high patient doses would probably be required using this geometry for in vivo work [Mossop 1988]. In vivo coherent/Compton studies have been performed by Karellas et al [1983] using an $^{241}$Am source and a scattering angle of 71°. A precision of 3% and
accuracy 5% has been quoted for a tissue dose of 3 mSv and a measuring time of 15 minutes [Shukla 1985]. Measurements were performed on the os calcis (heelbone), which contains a large amount of trabecular (spongy) bone which is highly sensitive to mineral loss.

5.2.4 Bone Densitometry using Coherent Scattering.

Kerr et al. [1980] in their study of scattering methods in the determination of bone density concluded that the coherent/Compton ratio technique is less sensitive to small changes in bone mineral content than a technique involving measurement of coherent scattering alone. Consideration of equations 5.2.13 - 5.2.14 indicates that the ratio of coherent to scattered photons varies as $Z'^{-2}$. However, the coherent scattering signal varies as $pZ'^{-2}$, and hence greater sensitivity to changes in mineral content which will result in a change in density as well as mean atomic number.

As in Compton scattering densitometry, a correction has to be made for the effects of the attenuation of both the incident and scattered radiation. The situation is however simplified in the coherent scattering case, due to the primary and scattered photons having the same energy. The method of correction involves two scattering measurements and two transmission measurements, as illustrated in figure 5.24. [Mossop 1988].

With the source in position 1 a transmission measurement $T$, using detector 2 and a scattering measurement $S$, using detector 1 are performed where:

$$ S = I \exp \left( \int_{V} \mu(x) \, dx \right) \cdot \frac{d\Omega}{d\Omega_0} (\theta) N \cdot V \cdot \exp \left( \int_{V} \mu(x) \, dx \right) \quad (5.2.16) $$
Figure 5.2.4: Coherent scattering densitometry:

attenuation correction by transmission.

(Mossop 1988).
\[ T_1 = I_o \exp(-\int_0^l \mu(x)dx) \exp(-\int_0^l \mu(x)dx) \quad (5.2.17) \]

The source is then placed in the second position and a second pair of measurements is performed:

\[ S_2 = I_o \exp(-\int_0^l \mu(x)dx) \frac{d\sigma}{d\Omega} \langle \theta \rangle N.V. \exp(-\int_0^l \mu(x)dx) \quad (5.2.18) \]

\[ T_2 = I_o \exp(-\int_0^l \mu(x)dx) \exp(-\int_0^l \mu(x)dx) \quad (5.2.19) \]

Combining the above equations leads to:

\[ R_{tc} = \left[ \frac{S_1, S_2}{T_1, T_2} \right]^{0,e} = \frac{d\sigma(\theta)/d\Omega_{coh}} {\sigma_{coh}} \quad (5.2.20) \]

This method of correction is subject to the same geometrical drawbacks as in the Compton/transmission method, and a high resolution detector is required to resolve the coherent signal from the Compton scattered radiation. Photon energies in the range 60-110 keV have been employed in order to maximise sensitivity and minimise patient dose. The site of interest best suited for the coherent technique is the os calcis, with the advantages of high trabecular content and accessibility. The use of a distal site is also preferable to minimise the risk to the patient.

Mossop [1988] made extensive in vitro studies of the coherent scattering technique, employing \(^{241}\text{Am} (59.54 \text{ keV})\) and \(^{153}\text{Gd} (97-103 \text{ keV})\) sources and a scattering angle of 70°. The \(^{241}\text{Am}\) source was found to be superior to the \(^{153}\text{Gd}\) terms of lower patient dose and longer half-life.

The results obtained indicated that the improvement in sensitivity
over the coherent/Compton method was about 10%. However the coherent/transmission method requires the added complexity of making four measurements, and it was concluded that the coherent/Compton method would be preferred for in vivo work.

5.2.5 Bone Densitometry Using Gamma Ray Scattering: Conclusions.

It has been demonstrated that gamma ray scattering can be employed as a method of determining bone mineral content and thus may have some role to play in determining those members of the population likely to suffer injury as a result of osteoporosis.

The selection of the optimum method for bone densitometry must be based upon a number of parameters, including accuracy, patient dose, scanning time and cost. In addition the benefits of standardisation may lead to the adoption of a single technique by clinicians even if that technique is not the most accurate available.

Gamma ray scattering techniques provide a method of obtaining useful precision without great expense, and are less susceptible to giving misleading results to to overlying tissue than absorptiometry. The method of data collection is quite slow using isotopic sources, this drawback could be overcome by using filtered X-ray sources [Cooper et al 1985, Webster 1985]. However the resulting radiation doses still would be of the order of a few mSv, considerably more than with CT densitometry. CT also has the advantage of providing an image of the site of interest, which may reveal structure of clinical interest.

Dual-photon absorptiometry will probably gain increasing acceptance as a standard technique of bone densitometry, due to commercial availability of DPA machines and their low cost compared to that of CT scanners.
5.3. Other Applications of Gamma Ray Scattering.

Although bone density can be determined using various gamma ray scattering techniques, the scattering method may also be applied in several other fields of diagnostic medicine. The Compton backscattering technique has been studied to determine the applicability of the method in measuring the density of the lung [Odeblad 1956]. Measurement of lung density is complicated by the fact that the density within the lung is considerably less than the surrounding tissues of the chest wall. Scattering techniques provide a means of overcoming this 'masking' effect, by using the source-detector collimator arrangement to define a volume of interest within the lung.

Wolf and Munro [1985] used an $^{241}$Am source in association with lung phantoms and found that density changes of 4% were measurable. The precise positioning of the volume of interest was important to ensure that the count resulting from the chest wall is a minimum. Positioning would be crucial for in vivo work due to the high density of the ribs. Multiple scattering has also been demonstrated as a major source of error [Kouris 1980]. Correction for multiple scattering over the range of densities encountered in lung measurements is possible using two energy windows for data collection. The first window is set around the Compton scattering peak at the relevant angle, the second is set at a lower energy to record multiple scattering events.

The coherent to Compton ratio has also been employed to study the fat content of the liver in vitro [Puumalainen et al 1977], and the iodine content of tissue [Puumalainen et al 1979], using an $^{241}$Am source and technique similar to that used in bone mineral measurement. In both applications the results were reproducible to about 3%. The technique has also been used to study the compositions of alloy castings [Cooper et al]
1982] and in association with XRF for studying lead in bronze alloys
[Gigante et al 1985].

Holt et al [1983, 1984] have demonstrated the potential of the
coherent to Compton scattering ratios for the determination of the
concentration of some solutions of biological interest. An annular 185 GBq
(5 Ci) $^{241}$Am source was employed and a 2.5% change in the scattering ratio
was observed corresponding to a 0.2% change in concentration.

A disadvantage of the coherent/Compton scattering technique is that an
expensive high resolution detector is required to differentiate between
photons scattered by the two processes. Timms et al [1987] demonstrated
that the two energies may be separated by using an appropriately chosen
filter to absorb the coherent scattered radiation whilst transmitting most
of the Compton scattered photons with energy below the absorption edge of
the filter. This method allows coherent/Compton scattering measurements to
be performed using low cost scintillation crystal detectors.

The relative simplicity of the apparatus required for scattering
measurements implies scope for employment in a range of areas. The
technique may well be further exploited, especially in industrial fields
where high radiation doses may be used to obtain high statistical
accuracy.
5.4 Imaging With Compton Scattered Radiation.

The technique of Compton scattering densitometry yields information about the electron density of an object. In some applications there are advantages to imaging using the Compton scattering technique (Holt 1985). Such images essentially constitute maps of the electron density (which is closely related to mass density), and in some fields this can lead to useful information on the macroscopic properties of the object under study. Compton scatter imaging has been performed in a range of applications in both the medical and non-destructive evaluation (NDE) fields. Since the medical and NDE fields differ greatly in their requirements, the two applications will be reviewed separately. In some circumstances, imaging using the Compton scattered radiation provides a useful alternative to transmission CT, and a comparison between the two techniques is made at the end of this chapter.

5.4.1 Compton Scatter Imaging in Medicine.

There are certain applications in medicine where the knowledge of density of structures within the body is of greater importance than spatial resolution. In such applications, the possibility of scanning the body with a radiation beam and imaging using the scattered radiation has been considered by a number of authors. A particular area of interest has been in the estimation of radiation dose contours for radiotherapy planning. Compton scatter imaging has also been found valuable for studying the density of regions of low attenuation coefficient within high attenuation objects.

These techniques, have been rather loosely described as 'scattering tomography'. The first studies of such techniques were carried out by
A variety of methods have been used for Compton imaging, with varying degrees of success.

These methods may be characterised by the scanning mode used, and fall broadly into:

- Point by point scanning.
- Line by line scanning.
- Plane by plane scanning.
- Reconstructive tomography.

### 5.4.2 Point by Point Scanning

The basis of this mode of scanning is illustrated in fig. 5.4(1), a pencil beam of radiation is raster scanned across the object and a focused collimator is employed to define a volume element within the object.

Early workers used the Compton scattering method to produce images of longitudinal planes within phantoms and small animals. A spatial resolution of about 5mm and density precision of 5% were obtained (Lale 1959, Clarke 1965, 1969). Later work has been carried out with the aim of producing images of transverse sections. Since Compton scattering is the dominant gamma ray interaction process between about 200 keV and 2 MeV, in biological materials most workers have employed isotope sources such as $^{137}$Cs (662 keV) or $^{60}$Co (1.25 MeV), typical source activities being several Curies in order to minimise scanning times.

The system designed by Clarke et al (1976) was configured to image a phantom mounted on a mobile couch, above which was mounted a highly collimated $^{60}$Co source. Two large NaI(Tl) detectors were used; one, to record the transmitted beam, was placed vertically below the source, while the second detector was placed at an angle of about 45° to the beam axis in order to detect the Compton scattered photons from the phantom. The
collimator system mounted on both the source and detector enables a sensitive volume within the patient to be defined since, neglecting multiple scattering, only scattering events occurring at the collimator intersection will be recorded. Movement of the couch using a raster scanning motion causes the sensitive volume to vary in position within the patient, thereby imaging a plane within the object.

By determining the scattered count rate at many points during the scan, the electron density within the object may be mapped. This was achieved by using the photomultiplier output of the detector to modulate the brightness of a cathode ray tube spot. Density variations of 1% were detectable during phantom studies. The device was further tested using dogs as subjects, with the sensitive volume being reduced to 0.5 cc. To obtain good spatial resolution it proved necessary to correct for respiratory motion, a crude method employed for this purpose was to deactivate the scanner during respiratory excursions. Presumably some method of blocking the gamma ray beam would be used if human subjects were being scanned. The overall scanning time using this device ranged from 40 minutes to one hour, approaching the limit for practical scanning of people. The gamma ray dose during this scanning procedure would probably be quite considerable as the photon utilisation is low, but the method was conceived as a technique for optimisation radiotherapy planning and thus would produce a net reduction in patient dose.

This essentially point-by-point technique has been further studied by Battista and Bronskill (1981). Improvements in scanning time were obtained using a 25TBq (700Ci) $^{60}$Co source, allowing a 0.32m x 0.32m body section to be scanned in 15 minutes. Two detectors for scattered radiation were employed, a weighted average of the two detector signals was used in order to determine tissue density. The technique was used in vivo, and the performance of the scanner was compared with that of commercial CT

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METHODS OF COMPTON SCATTERING TOMOGRAPHY

(i) Point by Point Scanning  
(Lale 1959)

(ii) 'Flying Spot' scanning  
(Jacobs 1979)

Beam raster scanned across object  
Depth determined by beam energy.
(iii) Line by Line Scanning
(Farmer and Collins 1971)

(iv) Plane by Plane Scanning
(Guzzardi 1981)
(v) SCAT CAT (Brateman 1984)

(vi) Line by Line Reconstructive Tomography
scanners. A spatial resolution of 5 mm was obtained, along with uncertainty in tissue density of 4% for a maximum patient dose of 0.11 Gy (11 rads). The authors considered that they had approached the limit of practical Compton scanner design, but the performance fell well short of that possible using a modern CT machine.

An interesting variation of the point by point method of Compton scatter imaging has been studied by Jacobs et al [1979]. This method interrogated the object by raster scanning the beam from an X-ray machine and viewing the backscattered radiation using uncollimated large area detectors, see fig 5.4(ii). Spatial resolution was determined by the diameter of the scanning beam. The absence of detector collimation allowed a relatively high counting rate for a given dose compared with other methods, but the results were more akin to radiographs, and were not used to give quantitative results. The use of several different applied kilovoltages could be employed to vary the image depth to some degree, allowing a slice to be obtained and built up to give 3-dimensional information.

The same group studied an aluminium phantom obscured by an aluminium plate 13 mm thick and obtained reasonable images. The abdominal cavity of a guinea pig was also imaged using this technique; the images of soft tissue showed quite well in comparison to a conventional radiograph, in which the image is normally dominated by high density tissues. The authors quoted spatial resolutions of down to 1 mm for doses of about 100 μGy. No in vivo studies were reported, however. This 'flying spot' method of scanning can be readily achieved using readily available apparatus, and the use of large area uncollimated detectors overcomes the difficulty of low scattering intensity. However, the reported results were only qualitative and the method would presumably be affected quite strongly by multiple scattering.
A different mode of Compton scatter scanning was devised by Farmer and Collins [1971], instead of point by point scanning they employed an essentially line by line technique. To achieve this a tightly collimated source of $^{137}$Cs (initially 2 Ci, later 40 Ci) to direct a narrow pencil beam of gamma rays through the object under study as shown in fig. 5.4(ii). A Ge(Li) detector collimated to accept photons scattered through 60-120° was used. The energy of the scattered photons was in the range 226-402 keV.

The principle of the scanning technique was to use the energy of the scattered photon to determine its point of origin within the object. The ability of the Ge(Li) detector to resolve energy to within 2 keV enabled reasonable spatial resolution to be achieved, since the energy of the scattered photon is dependent on the scattering angle. With an object of width 30 cm and a spread in scattered energies of 176 keV the minimum resolvable distance would be about 0.35 cm.

The spectrum recorded on an MCA thus represents an approximation to the density of the object along the beam line. This information is then stored, and the object is then moved in preparation for the next line scan, leading to the formation of a two dimensional image of the electron density within a plane inside the object.

Several factors tend to reduce the accuracy of this method of scanning. The first difficulty is that the differential cross-section for Compton scattering is forward peaked, leading to a factor of two variation in scattering probability over the detector acceptance angle. This effect is somewhat mitigated by the increase in detector efficiency with decreasing photon energy. Image contrast is reduced by the Compton continuum within the detector, this drawback could be reduced by mounting
sodium iodide detectors in an anti-coincidence arrangement, although this method would be cumbersome and expensive. A larger, coaxial Ge(Li) was employed in later work [Farmer and Collins 1972, 1974], in order to increase the full energy peak to Compton ratio of the detector.

A second factor which adversely effects scanner performance is the absorption of both the primary and scattered radiation within the body. The absorption of the primary beam leads to a reduction in the intensity of the beam within the patient. The results of primary beam absorption may be minimised by an empirical compensation technique, which reduces the MCA count by an amount determined by the depth within the object. Ideally this process would be achieved by the use of a minicomputer to process the results in real time. The effect of absorption of the scattered radiation was more pronounced due to the lower energy of the scattered photons. The authors suggested the use of two detectors on opposite sides of the object to reduce the effect of absorption by summing the results from the two detectors.

A third source of error encountered with this scanning technique was that of multiple scattering. The effects of multiple scattering were minimised by improvements in the design of the detector collimator, but the number of multiply scattered photons was still considered to be high enough to cause significant image degradation.

Tests of this mode of scanning were performed on phantoms made to simulate the human body. An average spatial resolution of 5mm was obtained for a scanning time of 20 minutes per section and a radiation dose of 20 mGy. No in vivo tests were reported using the scanner, although improvements to the design were suggested, such as large detector arrays. Such improvements would bring about large increases in cost, and the authors were somewhat pessimistic about the clinical prospects for the device.
5.4.4 Plane by Plane Imaging.

An alternative to point by point or line by line scanning in non-reconstructive Compton tomography is the use of a gamma camera (fig. 5.4 iv)). The employment of such a device has the advantage of utilising available apparatus to achieve plane-by-plane scanning. The gamma camera also detects a considerably greater fraction of the scattered radiation than a single conventional detector. To date the technique has been used to obtain images of the lung, since pathology of the lung is frequently associated with large changes of density (30-100%) [Pistolesi 1977, Giuntini 1978, Guzzardi 1978].

Some of the latest work in the field has been performed by Guzzardi and May [1981]. Their scanning technique employed a collimated fan beam of monochromatic gamma rays emitted from a 0.8m linear source of $^{203}$Hg (279 keV, $\tau_m = 47$ days) to irradiate the coronal chest section. A SELO large field of view (LFOV) gamma camera was used, being positioned over the patient’s chest with the detector plane parallel to the irradiation beam. Gamma rays Compton scattered through 90° are thus detected at an energy of 181 keV.

For in vivo study of the lung, the proposed source activity was 37 Gbq (1 Ci), corresponding to a measuring time of 3 minutes for detecting $5 \times 10^8$ photons in the image. Estimated patient dose was 420 µGy per section. The method was tested using various phantoms to determine spatial and density resolution of the system. The spatial resolution was 3.5 millimetres, and the density resolution was reported to be about 14% when correction for attenuation had been made. The correction for primary attenuation was achieved by irradiating a wooden phantom and storing the results as a correction matrix on the microprocessor used to operate the scan. Clinical studies of lung density have also been performed using
parallel hole gamma cameras in association with $^{99m}$Tc and $^{137}$Cs sources [Okuyama 1977, 1979], successfully imaging lung tumors.

The gamma camera technique could be usefully employed in some non-medical applications, although such uses have not been reported in the literature as yet.

5.4.5 Industrial Applications of Compton Scattering Tomography.

The technique Compton scatter imaging has several features which make it well suited to a range of applications in non-destructive testing. The principal advantages of the Compton scattering technique are that access is only required to one side of the object (in contrast to CT) and that contact with the object is not necessary (in contrast to ultrasound techniques).

In most NDE applications, the object under study is constructed of dense material and as a result radiation measurements must be performed at fairly high energies (hundreds of keV) to obtain adequate penetration within the object. At such energies Compton scattering forms the dominant contribution to the attenuation coefficient, thus making the Compton scattering technique particularly applicable.

To date most industrial applications of Compton scattering tomography have utilised the point by point method of imaging, in association with high activity sources of $^{137}$Cs or $^{60}$Co. The point by point method enables the viewing of regions of interest within the object, the desired size, position and depth of the volume under study is determined by the intersection of the source and detector collimators. By using an appropriate scanning system, three dimensional images of the object may be obtained without computer assisted reconstruction [Holt 1985].
Among the first workers to employ the principle of backscatter imaging were Harding et al. [1984], who used the technique to detect shrink holes in aluminium castings. An X-ray tube operated at 200 kVp was used as a radiation source, the energy required was quite low due to the low density of the casting and the X-ray generator gave much higher flux than is attainable with an isotopic source. An additional advantage of the low energy source was that the Compton scattering process was fairly isotropic, with obvious count-rate improvement in the backscattering mode. To further test the potential of the technique an experimental scanner (COMSCAN) was produced. In this device a 1 mm primary beam was aimed at the object, and scattered photons were detected by a set of 64 collimated BGO scintillator detectors. The focused collimator slits were reported to accept 85% of the radiation from the scattering volume. In addition to castings, a human skull was imaged successfully using COMSCAN, demonstrating possible medical applications of the apparatus.

The Compton tomography technique has been used in civil engineering applications [Gautam 1983], to study internal structure of objects such as reinforcing bars within concrete. The method used has been the measurement of the backscatter count-rate from a small volume within the object as a function of depth. The volume is defined by the intersection of the source and detector collimation. A range of sources including \(^{137}\text{Cs}\), \(^{60}\text{Co}\) and \(^{192}\text{Ir}\), each of activities of several Curies, were tested in conjunction with NaI(Tl) scintillator detectors. In the tests, the concrete sample was moved with respect to the source-detector arrangement, allowing the depth and position of the volume of interaction to be moved within the sample. Steel bars 9.5mm thick were easily detectable at depths of about 100 mm within the concrete. The imaging capabilities of the system were not reported although the use of computer graphics would provide an excellent method of displaying the count-rate verses position information. The
scanning time for the system was not reported.

The use of Compton scattering tomography for imaging underwater structures has been proposed by Bridge (1985). In North Sea oil recovery, the steel risers need to be tested for structural defects, and ultrasonic techniques are made impractical by the level of marine growth which occurs on the outside of the riser. Removal of the outer layers of growth is straightforward, but the few millimetres of encrustation immediately outside the metal is expensive and time-consuming to remove. The Compton scattering method does not require contact with the metal and thus provides a solution to the NDT problem. Computerized tomography could be employed for this purpose (a mobile CT scanner for underwater use has been designed by AMETEK Inc. for studying pier pilings) but the structure of the riser makes it poorly suited for CT. The riser consisted of a 920mm diameter tube with 32mm wall thickness. The Compton scattering device would only be required to interrogate the outer wall of the riser, whereas the CT technique would have to scan the whole object.

The proposed system utilises a $3.7 \times 10^{11}$ Bq (10 Ci) $^{60}$Co source mounted in a uranium and lead collimator/shield weighing 60-100 kg. A backsscatter ($135^\circ$) geometry was found to be best suited to the study, with the detectors incorporated into the collimator/shield. The detector system could employ CsI(Tl) scintillators linked to silicon photodiodes for compactness (Bridge 1988). At a scanning speed of 35 cm$^2$ per minute it was estimated that voids of about 125mm$^3$ could be detected.

An interesting industrial application of the Compton scattering tomography technique was performed by the IRT Corp. of San Diego USA, who developed a device to image artillery ammunition for the US Army. The shells were not particularly well suited to high contrast X-ray CT scanning due to the very high attenuation of the thick shell walls compared with the explosive filling. The machine, known as the Automated Inspection
Device for Explosive Charge Shells (AIDECS), [Costello 1980], initially used three tightly collimated 5000 Ci $^{60}$Co source; these were later replaced by an electron linear accelerator to provide an intense narrow pencil beam of very high energy X-rays, the scattered radiation was detected using 21 collimated NaI(Tl) detectors. Each detector collimator was designed to view only a small volume within the shell, thus allowing rapid, high precision measurements of the electron density to performed. Voids within the explosive filling could be detected with diameters as low as 1.6mm, in a scanning time of under one minute. Real time imaging was obtained by feeding the output pulses from the detectors into a computer. A three dimensional survey of the projectile was obtained using an appropriate scanning pattern. A computer graphics assisted display enabled the positions of voids to be determined easily by the operator.

5.4.6 Reconstructive Compton Tomography

True reconstructive Compton scattering CT has been performed by Brateman et al [1984]. This group placed an uncollimated scintillator detector at $30^\circ$ to the beam axis of a first generation CT scanner to record the scattered count. Simultaneous transmission measurements were also recorded to give rise to a transmission tomogram, data from which could be used to improve the accuracy of the scattering image; both detectors were used in the current mode, without any energy discrimination. The technique was named SCAT-CAT.

The radiation source used was a GE Maxitron 300 X-ray source operated at 140 kVp; the X-ray beam was collimated to give very low divergence (0.01 radian) across the object. As a result of the narrow beam, the scattered radiation is produced by a narrow band or 'scattering column' within the object. The object was divided into an array of pixels, each with side
length equal to the width of the incident beam. The scattering image was reconstructed using filtered back-projection. Correction algorithms were used to compensate for the effects of both primary and secondary attenuation, solid angle at the detector and detector efficiency. Solid angle corrections were determined analytically, and attenuation coefficients could be determined from CT images. The effects of beam hardening within the object were not reported, but were probably not of great significance as the phantoms studied consisted of perspex discs 100mm in diameter with holes and aluminium pins to represent internal structure. No correction was made for multiple scattering effects, but a Monte Carlo program was used to predict the degree of multiple scattering; it was estimated that 20% of the recorded scattering resulted from multiple events for the phantom under study.

The SCAT-CAT images were effective at displaying holes within the phantoms but were less effective in displaying the aluminium pins, these results were expected as the difference in electron density between perspex and aluminium is considerably less than between perspex and air. The qualitative results were not consistent with the known phantom values, this was thought to be due to deficiencies in the reconstruction algorithm. The advantages of performing SCAT-CAT seem limited, as the presence of voids and aluminium pins could presumably have been determined from the transmission CT scan.
5.5 Experimental Reconstructive Compton Tomography.

Two possible modes of reconstructive Compton tomography were considered:

(i) Using a broad beam of gamma radiation to interrogate the object, with a collimated detector placed to detect the scattered radiation at a given angle. In this mode, spatial resolution is provided by the detector collimation.

(ii) A tightly collimated beam of photons could be used to irradiate the object, with Compton scattered photons being detected by an uncollimated large area detector. The raysums would then consist of the line integrals of the scattered counts from along the beam. Spatial resolution is thus determined by the width of the beam within the object.

The first method was considered unfeasible due to the small fraction of photons that would be scattered into the field of view of the detector and the very large amount of multiple scattering which would occur, resulting in poor contrast.

Method (ii), however, looked more promising, and some imaging attempts could be made using available equipment. In order to obtain adequate count-rate from the scattered radiation, an uncollimated coaxial n-type HPGe detector was employed. This large volume detector, described in more detail in chapter 5, offered high efficiency combined with good energy resolution which would enable the count-rate resulting from multiple scattering to be reduced by using appropriate energy discrimination.

As spatial resolution was determined by the beam width within the object, it was necessary to use a collimator with length similar to the diameter of the object under study.

The first source used was 200mCi of $^{241}$Am mounted in the holder designed by Folkard (1983), using a 50mm long tubular steel collimator of
3mm bore. A further collimator, of 10mm length and 2mm bore was mounted on the end of the steel tube. The beam width from this collimator arrangement increases from 2 to about 4mm across a 50mm diameter object.

The detector was mounted above the scanner bed, with its long axis immediately above the radiation beam. Scattered radiation originating from the beam path through the object would be detected, and each measurement thus constitutes a line integral of scattering centres. Each set of line integrals at a given angle forms a projection which could be reconstructed into an image using the filtered back-projection technique described in chapter 3.

The first object to be studied was a water-filled container placed within a hollow perspex cylinder, 60mm in diameter and 5mm thick. The SCA window was set around the 90° scatter peak for 60 keV radiation (52-56 keV). Sixty 1mm raysums and 60 projections were taken in an overall scanning time of 30hrs. The image obtained is shown in fig. 5.5a. The count-rate in air was 13 per second, rising to 225 per second through the central region.

The second test object to be scanned was a 60mm diameter beaker of water (fig 5.5b), this uniform object provided a test of the effect of beam absorption on the image, considerable attenuation of the ingoing beam would lead to 'cupping', that is a reduction in the measured density towards the centre of the object. Examination of a line sections through the image (fig 5.5c) shows a drop in image value of about 20% towards the centre, it also displays the poor image statistics obtained during the scan. The attenuation of 60 keV gamma rays in 60mm water was calculated to be 71%.

Having demonstrated the feasibility of this mode of scanning, a more interesting object was scanned, this time a human os calcis (heelbone). The os calcis contains a very large proportion of spongy or trabecular bone and as a result is particularly sensitive to the changes produced by
Figure 5.5a: Compton scan of water-filled container in a perspex surround.
Figure 5.5b: Compton scan of water-filled container.

Figure 5.5c: Line section through above image.
osteoporosis [Parfitt 1983]. The os calcis was placed within a thin-walled vessel on the scanner table and was scanned using the same parameters as used for the previous scan. The resulting image, 5.5d, appeared to show density variation within the bone. For practical imaging, however, the bone would be covered by other tissues. As a crude simulation of soft tissue, the os calcis was placed in a water filled container and the scan was repeated. The resulting image, fig 5.5e, was disappointing, all detail of internal structure having been lost. The probable causes of this were multiple scattering, the effect of which is to produce a 'background' capable of masking small density changes, and the effects of increased attenuation reducing the number of photons interrogating the bone volume and thereby increasing statistical noise.

To indicate the spatial resolution of the system, a 60mm diameter cylindrical perspex phantom was scanned. A series of holes had been drilled into the phantom, ranging in size from 0.1-10mm. The object was scanned using 42 linear steps and 60 projections at 6° intervals, counting time was 10 seconds per raysum giving a total scanning time of 7 hours. The count-rate with the primary beam passing through air was 33 per second, rising to about 1000 per second with the beam passing through the diameter of the phantom. The results of this scan are shown in fig 5.5f.

A 'Marathon' bar was scanned using similar parameters, except with a 25 second counting interval. A maximum count-rate of 140 per second was obtained, the resulting image is shown in fig 5.5g.

The range of objects that could be scanned using the 241Am source was limited by the relatively low penetration of the beam within dense, high Z materials. For objects of moderate density, the effect of attenuation is to make quantitative imaging impossible without some form of attenuation correction technique. In objects whose sizes are comparable with the path
Figure 5.5d: Compton scan of os calcis.

Figure 5.5e: Compton scan of os calcis in water.
Figure 5.5f: Compton scan of perspex phantom.

Figure 5.5g: Compton scan of 'Marathon' bar.
length of 60 keV radiation, the effect of attenuation on both the primary and secondary radiation is severe, leading to gross image degradation.

A $^{152}$Gd source, provided by Amersham International plc was used to perform Compton scattering tomography. The source emits 97 and 103 keV gamma rays, in addition a high intensity of Eu X-rays (40-48 keV) are produced. The gamma rays of energy ~100 keV are attenuated less severely than the 60 keV gamma rays for elements with an atomic number less than 69. The $^{152}$Gd source was filtered using barium nitrate to remove the Eu X-rays and the holder used was that designed for the $^{60}$Co source described in chapter 4. A 130mm long copper collimator with 2mm exit aperture was employed to provide a pencil beam, and the perspex phantom was scanned using 42 2mm steps and 60 projections at 6° intervals. The count-rate with the beam passing through air was measured to be 10 per second, with a maximum count-rate of about 75 per second. The image obtained using 60 seconds per raysum is shown in fig 5.5h. The image is very similar to that obtained using the $^{241}$Am source, the smallest resolvable void is 3mm in diameter. The central upper part of the image shows the hint of a void, indicating slightly superior contrast and resolution to the scan at 60 keV. This is probably due to lower attenuation within the object and a more parallel primary beam produced by the much longer collimator, resulting in reduced multiple scattering and better spatial resolution at large distances from the source.

High energy sources such as $^{137}$Cs or $^{60}$Co would be considerably more effective due to the considerably lower attenuation of the higher energy radiation. Unfortunately no such sources of sufficient activity were available for use at the time the Compton scattering experiments were performed.

It should be noted that the samples scanned were well suited to reconstructive scanning as they were reasonably flat such that the effects
Figure 5.5h: Compton scan of perspex phantom using $^{152}$Gd
of attenuation of the scattered beam were not considerable, the technique could be applied by using a detector positioned to the side of the object. Modifications would have to be made to the mode of scanning to correct for the asymmetric effects in attenuation resulting from different absorption thicknesses depending on the amount of material between the beam position and detector.

5.6 Coherent Scattering Tomography.

Although not of great importance as an attenuation process, coherent (Rayleigh) scattering has proved of considerable research importance as it is the coherent scattering of X-rays that produces the Bragg diffraction patterns indispensable to crystallography. In the past few years, the possibility of utilising the Rayleigh scattering process in tomographic imaging has been considered, and may become important in a range of applications in the future.

At small angles and low energies (less than 10° for 100 keV radiation), coherent scattering is the dominant scattering process [Johns 1983]. It has been noted [Hukins 1981] that if a narrow beam of X-rays is passed through biological materials then interesting diffraction patterns may be observed. The variation in these patterns between different substances is considerable, indicating that the Rayleigh scattering process is dependent upon subtle structural effects at a molecular level.

Harding et al [1985] have attempted to form tomographic images by reconstructing measurements of the Rayleigh scattered intensity. The basis of the technique was to employ an array of BGO scintillation detectors to record the intensity of X-rays scattered from a sample; the central detector measured the transmitted intensity and the other detectors
measured the radiation scattered over a 6° half-angle. Each scattering detector output is normalised using the transmitted intensity recorded by the central detector. If the distance between the object and the detector is large in comparison with the object dimensions, then all the radiation incident upon a given detector at an angle θ to the transmitted beam will be a line integral related to the differential scattering cross-section at the angle θ. Translating and rotating the object within the beam enables an image to be reconstructed.

A disadvantage with this technique results from the continuous nature of the X-ray spectrum, which causes blurring of the diffraction effects since the differential scattering cross-section is a function of photon energy. Kosanetzky et al [1986] have attempted to reduce the energy blurring by using energy-resolved scanning using a HPGe detector in association with a multi-channel analyser.

Although the count-rate obtained would be considerably lower, the use of a radioisotope source or highly filtered X-ray beam would remove the energy blurring. It would not, however, remove the blurring effects caused by the finite angular widths of the object and detector.

The coherent scatter tomography technique is of principal interest because the resulting images reveal information of molecular structure, in contrast to most other forms of radiation-based imaging which indicate only atomic species and density. If the technique of coherent or 'diffraction' tomography can be developed into a practical system then it may have a valuable role to play the non-destructive examination of objects for the detection of desirable or undesirable substances. The technique is particularly useful as it is applicable over the whole range of atomic number.
5.7 Compton Scatter Imaging: Conclusions.

The principle of imaging an object using the Compton scattered radiation from an incident beam of gamma radiation has been demonstrated by a number of workers in the fields of diagnostic medicine and non-destructive testing. A range of different scanning techniques have been performed in order to form 2 or 3-dimensional images of the object under investigation using the technique of Compton scatter imaging.

Table 5.7 compares the Compton scattering technique with transmission CT.

From table 5.7 it can be seen that the non-reconstructive Compton scattering technique may offer some useful advantages over CT in some circumstances, for example when:

(i) Variations in density rather than elemental composition are important, such as imaging of the lung for medical use.

(ii) Access to only side of an object is possible/desirable, for example in looking at the surface of an extended object.

(iii) It is necessary to view a limited volume within an object, for example the explosive filling of munitions.

The principle disadvantages of this simple Compton scattering technique are:

(i) Solid angle effects: only a very small fraction of the scattered radiation from the volume of interest will reach the detector in most circumstances. Solid angle considerations also limit the degree of collimation that is practicable.

(ii) The effect of multiple scattering which can produce considerable errors in the measured value of $\rho_e$ [Battista et al 1977].

(iii) The attenuation of the secondary (scattered) radiation, which may have an energy considerably below that of the primary beam (for example
Table 5.7 Comparison of Compton Scattering and CT Imaging.

<table>
<thead>
<tr>
<th>Parameter Imaged</th>
<th>Compton Scatter Imaging</th>
<th>Transmission CT</th>
</tr>
</thead>
<tbody>
<tr>
<td>Access from both sides necessary</td>
<td>No</td>
<td>Yes</td>
</tr>
<tr>
<td>Computer reconstruction necessary</td>
<td>No</td>
<td>Yes</td>
</tr>
<tr>
<td>Region of Interest scanning possible</td>
<td>Yes</td>
<td>No, beam must pass through whole object¹</td>
</tr>
<tr>
<td>Multiple scattering difficulties</td>
<td>Yes</td>
<td>In-scattering may reduce measured $\mu^2$</td>
</tr>
</tbody>
</table>

¹ It is possible to reduce scanning time by concentrating raysum measurements across a region of interest (ROI). An example is when it is required to scan a series of samples placed within an identical container, the container may be accurately prescanned and the appropriate raysums stored. The region of interest within the container may then be scanned for each sample and these raysums combined with the previously determined raysum values for the container to reconstruct the image.

² In-scattering may be reduced by using tight source and detector collimation, although this will limit the closeness of detector packing in a multi-detector array.
the attenuation coefficient for backscattered gamma rays from $^{60}$Co in iron is approximately 2.5 times that for the primary radiation).

(iv) The low value of the Compton differential cross-sections at large angles at high energies reduces the count-rate for backscatter geometry.

The Compton scattering technique will tend to produce a much lower counting rate than CT for a given source and spatial resolution. The flux $I$ at a detector in a CT device will be given by:

$$I \propto I_0 \exp\left(-\int \mu(x)dx\right) \quad (5.71)$$

where $r$ is the source-detector distance and $x$ is the thickness of the material.

In the case of Compton scattering tomography, however, the flux at the detector is given by:

$$I \propto I_0 \exp\left(-\int \mu(x)dx\right) \cdot N_{\text{com}}(\theta) \cdot \exp\left(-\int \mu'(x')dx'\right) \quad (5.72)$$

where $r$ is the distance of the volume of interest from the source, $R$ is the distance from the volume of interest to the detector.

It is interesting to compare the technique of Compton scatter imaging with transmission CT for crack determination in the concrete sample considered in Chapter 4. Assuming that 2mm spatial resolution is required, an estimate can be made of the counting rate using a 370 GBq ($^{103}$Cl) $^{60}$Co source. If a 150mm long collimator is used to define a 4mm² area within the centre of the sample, then by solid angle considerations the photon flux through the area would be approximately $1.3 \times 10^6$ per second. Attenuation of the ingoing beam would reduce the flux at the volume of interest by a factor of eight and within the 2mm deep element approximately 2.5% of photons would interact via a Compton scattering process.
Attenuation of the outgoing scattered beam at an angle of 135° would reduce the scattered flux to about 2% of its initial intensity. Assuming scattering to be isotropic (an overestimate) then the counting rate at a 15cm by 15cm NaI(Tl) detector within a collimator focused at the volume would be only 20 per minute if 25% transmission through the collimator is assumed! For comparison, the transmission technique in chapter 4 achieved an average counting rate of 250 per second using a 7.4 GBq source. To scan over one slice through the concrete cylinder requires 70,000 measurements and so clearly the technique is much too slow for this application.

The technique of Compton scattering tomography is of value in some applications in which it shows marked superiority to transmission CT. Two such an applications are:

(i) Flaw inspection of a steel pipes where the ability of the Compton technique to interrogate a volume of interest would prove particularly valuable allowing all the measurements to be performed on the steel structure whilst ignoring the pipe contents and any surface material [Bridge 1988].

(ii) Investigation of the near-surface volume of a large object, in such circumstances the requirement of single side access is fulfilled by the Compton technique [Harding 1984].

In assembly-line applications correction for attenuation could be made by calibrating the instrument using unflawed samples. In non-destructive examination of materials, high activity sources and focused detector collimation would be employed to provide adequate counting rates.

The technique of reconstructive Compton scattering using a large uncollimated detector was shown to enable reasonable counting rates to be achieved with low energy sources of moderately high activity being used to study low density objects with dimensions of a few centimetres. The quantitative value of the technique was shown to be limited by the effects
of attenuation and multiple scattering. Correction techniques similar to
those at present applied in emission tomography could be utilised to
increase the accuracy of the technique. Despite this the usefulness of the
technique is limited as the method possesses many of the disadvantages of
the transmission CT technique without the benefits of high counting rate.

In medical applications the dose limitations means that the Compton
technique will remain less popular than transmission CT.

Although not yet fully explored, the technique of coherent scattering
tomography has been shown to produce interesting results, and may have a
number of important applications in areas where different compounds of
similar density and atomic number need to be distinguished by a non-
destructive method.
6.0 Introduction.

The technique of X-ray fluorescence has proved a valuable method of non-destructive evaluation, finding applications in medical physics (Ahlgren 1977), archaeology (Tite 1972) and the minerals industry (Balaes 1987). The main use of X-ray fluorescence (XRF) analysis is the non-invasive evaluation of the concentration of a specific element within a sample. In many applications, it would be of value to determine spatial distribution of a given element in addition to average concentration. At present spatially resolved XRF measurements are performed on the micron-scale in scanning electron microscopy; the use of tomographic methods on a macroscopic scale is a natural extension of the XRF technique.

6.1 Brief Theory.

The process of X-ray fluorescence is a result of the rearrangement of atomic electrons following the inner shell ionization of an atom (Dyson 1973). The ionization may be produced by bombardment with charged particles such as electrons or protons, or by photoelectric interactions using high energy photons. For various practical reasons it is the latter case with which we will be chiefly concerned in this work.

The basic physics of the photoelectric effect, in which a photon is absorbed by an atom with the resulting ejection of an electron, is well
understood, the process having been an early indicator of the quantum nature of radiation (see chapter 2). To cause the emission of an electron from a given bound state, the photon energy must exceed the binding energy of the electron. In the case of the inner shells of heavy elements many tens of keV are required. Following inner shell ionization, the electrons rapidly rearrange themselves with the vacancy left by the ejected electron being filled from an outer shell. The energy released by this process may result in the emission of an X-ray photon or in the ejection of an Auger electron (the Auger effect also gives rise to the emission of X-rays though with a rather complex spectrum). The X-rays emitted are of an energy characteristic of the atomic species, thus enabling XRF to serve as a means of analysing the elemental composition of a sample.

On account of their higher energies, the X-rays generally used for imaging of macroscopic objects are the so-called K X-rays. The K series can frequently be resolved into four components, conventionally denoted by $K_{\alpha 1}$, $K_{\alpha 2}$, $K_{\beta 1}$, and $K_{\beta 2}$. The $K_{\alpha}$ lines result from transitions from the 2p to the 1s energy level, the $K_{\beta 1}$ and $K_{\beta 2}$ lines result from transitions to the 1s level from the 3p and 4p levels respectively. Fig 6.1a shows the X-ray emission notation for cadmium.

In order to excite the K X-rays, the value of $E_\gamma$ must exceed the value of the K absorption edge of the target element. The energy of the K edge increases strongly with the atomic number ($\sim Z^2$), as shown in fig. 6.1b. The probability of a photon undergoing a photoelectric interaction with an atom is determined by the photoelectric cross-section $\sigma_{ph}$, which is a complex function of $E_\gamma$ and $Z$. A full description of the theory of the photoelectric cross section may be found in Jackson [1981], and tabulated values of $\sigma_{ph}$ can be found in Storm [1970]. Table 6.1 shows values of K X-ray energies and $K_{\alpha}$ and $K_{\beta}$ intensity ratios for various elements.

For a given element the value of $\sigma_{ph}$ falls rapidly with increasing
**Figure 6.1a: X-ray emission lines from cadmium**

*(White 1934)*
Figure 6.1b K-edge as a function of atomic number.

Figure 6.1c: Photoelectric cross-section of Tm.
Table 6.1

K Shell X-ray Data for Selected Elements

<table>
<thead>
<tr>
<th>Element</th>
<th>Z</th>
<th>$K_{\alpha_1}$/keV</th>
<th>$K_{\alpha_2}$/keV</th>
<th>$K_{\beta_1}$/keV</th>
<th>$K_{\beta_2}$/keV</th>
<th>Yield Ratio $K_{\beta}/K_{\alpha}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Si</td>
<td>14</td>
<td>1.838</td>
<td>1.832</td>
<td>1.740</td>
<td>1.739</td>
<td>0.027</td>
</tr>
<tr>
<td>Ca</td>
<td>20</td>
<td>4.038</td>
<td>4.012</td>
<td>3.591</td>
<td>3.686</td>
<td>0.125</td>
</tr>
<tr>
<td>Fe</td>
<td>26</td>
<td>7.111</td>
<td>7.057</td>
<td>6.403</td>
<td>6.390</td>
<td>0.135</td>
</tr>
<tr>
<td>Cu</td>
<td>29</td>
<td>8.980</td>
<td>8.976</td>
<td>8.904</td>
<td>8.047</td>
<td>0.142</td>
</tr>
<tr>
<td>Ge</td>
<td>32</td>
<td>11.103</td>
<td>11.100</td>
<td>10.981</td>
<td>9.885</td>
<td>0.147</td>
</tr>
<tr>
<td>Zr</td>
<td>40</td>
<td>17.998</td>
<td>17.969</td>
<td>17.666</td>
<td>15.774</td>
<td>0.190</td>
</tr>
<tr>
<td>Ag</td>
<td>47</td>
<td>25.517</td>
<td>25.454</td>
<td>24.942</td>
<td>22.162</td>
<td>0.212</td>
</tr>
<tr>
<td>I</td>
<td>53</td>
<td>33.164</td>
<td>33.016</td>
<td>32.292</td>
<td>28.610</td>
<td>0.228</td>
</tr>
<tr>
<td>Xe</td>
<td>54</td>
<td>34.571</td>
<td>34.446</td>
<td>33.644</td>
<td>29.802</td>
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</tr>
<tr>
<td>Cs</td>
<td>55</td>
<td>35.959</td>
<td>35.819</td>
<td>34.984</td>
<td>30.970</td>
<td>0.234</td>
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<tr>
<td>Ba</td>
<td>56</td>
<td>37.410</td>
<td>37.225</td>
<td>36.376</td>
<td>32.191</td>
<td>0.237</td>
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<tr>
<td>Pm</td>
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<td>45.207</td>
<td>44.955</td>
<td>43.945</td>
<td>38.649</td>
<td>0.248</td>
</tr>
<tr>
<td>Eu</td>
<td>63</td>
<td>48.515</td>
<td>48.241</td>
<td>47.027</td>
<td>41.529</td>
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<tr>
<td>Gd</td>
<td>64</td>
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<td>49.961</td>
<td>48.718</td>
<td>42.983</td>
<td>0.255</td>
</tr>
<tr>
<td>Tm</td>
<td>69</td>
<td>59.935</td>
<td>58.969</td>
<td>57.576</td>
<td>50.730</td>
<td>0.262</td>
</tr>
<tr>
<td>W</td>
<td>74</td>
<td>69.508</td>
<td>69.090</td>
<td>67.233</td>
<td>59.310</td>
<td>0.269</td>
</tr>
<tr>
<td>Pt</td>
<td>78</td>
<td>78.379</td>
<td>77.866</td>
<td>75.736</td>
<td>66.820</td>
<td>0.275</td>
</tr>
<tr>
<td>Hg</td>
<td>80</td>
<td>83.106</td>
<td>82.526</td>
<td>80.258</td>
<td>70.821</td>
<td>0.278</td>
</tr>
<tr>
<td>Pb</td>
<td>82</td>
<td>88.001</td>
<td>87.343</td>
<td>84.992</td>
<td>74.957</td>
<td>0.280</td>
</tr>
<tr>
<td>Rn</td>
<td>86</td>
<td>98.418</td>
<td>97.616</td>
<td>94.877</td>
<td>83.800</td>
<td>0.286</td>
</tr>
<tr>
<td>U</td>
<td>92</td>
<td>115.591</td>
<td>114.549</td>
<td>111.289</td>
<td>98.428</td>
<td>0.289</td>
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<tr>
<td>Am</td>
<td>95</td>
<td>124.876</td>
<td>123.706</td>
<td>120.153</td>
<td>106.351</td>
<td>0.292</td>
</tr>
</tbody>
</table>

[Data from Dyson (1973) and Weast (1987)].

138
photon energy until one of the absorption edges is reached, at which point \( \sigma_{\text{ph}} \) undergoes a sharp increase, see fig. 6.1c which shows the value of \( \sigma_{\text{ph}} \) versus \( E_\gamma \) for thulium \( (Z=69) \). The probability of a photoelectric interaction increases very strongly with \( Z (~Z^a) \); a graph of \( \sigma_{\text{ph}} \) for \( E_\gamma \) of 60 keV is shown in fig. 6.1d.

The relative number of K X-rays emitted per photoelectric interaction is determined by the K shell fluorescent yield \( \omega_K \) defined as:

\[
\omega_K = \frac{X_K}{X_K + A_K} \tag{6.1.1}
\]

where \( X_K \) is the number of K X-ray photons emitted and \( A_K \) is the number of Auger electrons. Calculation of the fluorescent yield thus involves determining probabilities for the two competing processes. The probability of non-radiative transitions is largely independent of atomic number whereas the probability of a radiative transition varies as \( Z^a \) [Dyson 1973]. It is possible therefore to write the K fluorescence yield as:

\[
\omega_K = \frac{Z^a}{a + Z^a} \tag{6.1.2}
\]

where \( a \) for the K shell is \( 1.12 \times 10^5 \) [Burhop 1952]. It is observed that the value of \( \omega_K \) increases rapidly as \( Z \) increases for light elements; a graph of \( \omega_K \) as a function of \( Z \) can be seen in fig. 6.1c.

In general, therefore, it should be noticed that, for a given intensity of photons with energy exceeding the K-edge of an element, target atoms with high atomic numbers fluoresce with greater intensity and emit radiation of a higher energy. Both these factors tend to make XRF analysis easier to perform on high Z materials.
Figure 6.1d: Photoelectric cross-sections at 60 keV

Figure 6.1e: K fluorescence yield as a function of Z.
6.2 Imaging Using X-Ray Fluorescence.

As in the case of Compton scatter tomography, there are a number of methods of forming images using the XRF radiation emitted from an object under irradiation. All the techniques shown in fig 5.4 may be employed for imaging purposes.

The requirements of a XRF Tomography system are:

1. A source of radiation, emitting photons with energy exceeding the K edge of the target element.
3. A detector system with energy resolution capable of resolving the fluorescent radiation produced by the various target elements from each other and from any incident background radiation.

The scanner system must be designed to maximise the counting rate and thus minimise scanning time for the required sensitivity. Careful attention must, therefore, be paid to the selection of the source energy, detector type and collimator system.

6.2.1 Source Selection.

As outlined in section 6.1, the photoelectric cross-section for a given element is a strong function of the energy of the incident radiation, rising to a maximum when the incident photon energy just exceeds the absorption edge of the relevant electron shell. It is thus desirable for a radiation source to be employed with a gamma energy slightly above the K-edge of the target element. It should, however, be noted that the primary beam must also be capable of penetrating the sample in order to excite fluorescence throughout the desired volume.

It is, in principle, possible to use an X-ray machine in place of a
radioisotope as a primary radiation source. As discussed in chapter 4, this enables a considerably greater flux of radiation to be produced, thus increasing the XRF emission rate and allowing faster scanning. However, the continuous energy spectrum from the X-ray tube would introduce considerable difficulties as the large numbers of Compton scattered photons would inevitably create a sizeable number of spurious counts in the energy acceptance window.

6.2.2. Detector Selection.

The two most important parameters on which selection of the detector system for XRF scanning depends are energy resolution and efficiency.

The detector must be able to resolve the fluorescent radiation from the target element from:

(i) Primary radiation from the excitation source.

(ii) Compton scattered radiation produced within the sample.

(iii) Fluorescent X-rays produced by any other element present within the sample.

(iv) 'Background radiation' from other radiation sources.

Clearly the energy resolution required depends upon the energy of these various sources. Of particular importance is energy of Compton scattered radiation from the sample, which may constitute a considerable source of 'noise' if the scattered photon energy given by equation 2.41 is not clearly separated from that of the XRF radiation.

One of the main advantages of the XRF technique is that it is element specific if the target element fluorescent radiation can be resolved from those produced by any other elements within the test object. As a result the presence within the object of elements with atomic numbers close to
those of the target and with absorption edges below the source radiation will require good energy resolution.

A high full energy peak detector efficiency at the target energy is required to minimise scanning time. Count rate is also heavily influenced by the design of detector collimation, and, as with other modes of scanning a compromise has to be made between counting rate and spatial resolution.

6.3 Review of XRF Imaging.

One of the major limitations on XRF scanning is the effect of attenuation of both ingoing and fluorescent radiation within the object under study. At low photon energies, the attenuation is a strong function of the atomic number of the material. XRF scanning is thus ideally suited to applications where a high Z element is situated within a low Z material.

Biological materials consist mainly of low atomic elements such as carbon, hydrogen and oxygen, and as a result are reasonably transparent to the characteristic X-rays of heavy elements. As a result the XRF imaging technique is a potentially valuable method in a range of diagnostic techniques involving high Z materials.

X-ray fluorescence imaging in medicine was first proposed by Hoffer [1968] as a method of studying the thyroid gland. The thyroid gland is situated at the front of the neck and accumulates comparatively large amounts of iodine (10 - 15 mgm) which is used in hormone production. For many years study of thyroid function has been achieved by introducing radioactive isotopes of iodine into the body and then detecting the radiation emitted using detectors sited at the neck. Imaging of the thyroid based upon these emission techniques is extensively used in the diagnosis of thyroid tumours [Sandler 1987].
In the past the most frequently used isotope was $^{131}$I, a beta/gamma emitter with an eight day half life. Due to the high radiation dose which occurs using this source (10-100 mSv to the thyroid with an administration of a few microcuries) it has recently become more common to employ $^{123}$I produced in a cyclotron ($E_\gamma = 159$ keV, $t_{1/2} = 13.6$ hrs) or $^{132}$I ($E_\gamma = 66-76$ keV, $t_{1/2} = 2.3$ hrs) 'milked' from a $^{123}$Te generator. Although not 'organified' within the thyroid, $^{99m}$Tc has also been used for thyroid imaging [Sandler 1987]. Imaging using these techniques is facilitated by the use of raster scanning a collimated scintillation detector across the region of interest, spatial resolution being determined by the collimator aperture; gamma cameras and multi-wire detectors have also been used.

Despite the lower radiation doses resulting from administration of short-lived isotopes, Hoffer [1968] proposed that XRF scanning could be employed for thyroid studies, using this technique no radioisotopes would need to be administered to the body and, with appropriate collimation, the radiation dose could be limited to the thyroid and nearby tissues. The basis of the technique involved a collimated source of 60 keV gamma rays, employed in association with a Si(Li) detector which was used to measure the emission of fluorescent I X-rays (average energy 26.5 keV).

The work of Hoffer has been improved upon by Patton et al [1975, 1978a, 1980]. This group employed a single Si(Li) detector constructed to view the common point of irradiation of an annular $^{241}$Am, source consisting of sixteen 37 MBq discs mounted concentrically with the detector. Trial scans were performed using phantom studies which demonstrated the effectiveness of the scanning technique; a HPGe detector was selected to replace the Si(Li) device for subsequent scanning. Useful results were obtained for in vivo studies. The group reported that better differentiation between malignant and non-malignant tissue could be obtained using simultaneous $^{99m}$Tc emission and XRF imaging rather than with
XRF imaging alone, due to the different uptake and storage properties of the two elements in the thyroid. The imaging technique involved raster scanning the thyroid area of the patient in a time of about 15 minutes, with integrated counting rates of several thousand per second.

To reduce scanning time, a detector array consisting of nine 35mm diameter HPGe crystals cooled from a single liquid nitrogen vessel and connected to a single HT supply has been devised [Patton 1978b]. This system is designed to be used in association with either a focused or frontal plane collimator arrangement.

The technique of XRF is, in principle, suitable for a range of biomedical applications where high atomic number materials accumulate within the body. Heavy metals such as lead and platinum may accumulate in body organs and lead to harmful effects. In general the concentrations of these elements in tissue is small (<20 ppm) [Somervaille 1985] and the relevant organs are covered by substantial thicknesses of soft tissue, reducing count rate to a level which is probably too low for imaging within the dose and time limits available for medical scanning.

6.4 Initial Experiments.

Two possible methods of performing reconstructive tomography using XRF were considered. The first method consists of uniformly irradiating the object with gamma radiation of the appropriate energy. Projection data is then built up by translating and rotating the object across the field of view of a collimated detector. Spatial resolution is thus defined by the detector field of view.

The second mode, to be described in detail later, utilises a narrow
beam of primary radiation to define spatial resolution, with the detector
being left uncollimated.

The isotope selected for use was $^{241}\text{Am}$, which was readily available in
sources of 7.4 GBq activity. For the initial experiments a sample of 5
grammes of thulium chloride ($\text{TmCl}_3\cdot7\text{H}_2\text{O}$) was obtained. Thulium has a K-edge
of 59.34 keV, and thus has a very high photoelectric cross-section for the
59.54 keV gamma ray from $^{241}\text{Am}$.

Some relevant data on thulium is shown in Table 6.4

<table>
<thead>
<tr>
<th>Table 6.4: Thulium Data</th>
</tr>
</thead>
<tbody>
<tr>
<td>Mean Atomic Mass</td>
</tr>
<tr>
<td>Photoelectric cross-section (60 keV)</td>
</tr>
<tr>
<td>Compton cross section (60 keV)</td>
</tr>
<tr>
<td>Differential Compton cross-section at 90°</td>
</tr>
<tr>
<td>Mass attenuation coefficient (60 keV)</td>
</tr>
<tr>
<td>K Fluorescent Yield</td>
</tr>
<tr>
<td>Ratio of $K_α$/$K_β$ intensities</td>
</tr>
<tr>
<td>Energy of $K_{α_2}$ emission</td>
</tr>
<tr>
<td>Energy of $K_{α_1}$ emission</td>
</tr>
<tr>
<td>Energy of $K_{γ_2}$ emission</td>
</tr>
<tr>
<td>Energy of $K_{γ_1}$ emission</td>
</tr>
</tbody>
</table>

[Huebell 1975, Weast 1987]

The only high resolution detector available at the outset of this work
was a hyper pure germanium detector (active thickness 5mm, area 200mm$^2$)
manufactured by EG&G Ortec. The detector was mounted in a vertical geometry
with the active face directed upwards, the end window was constructed of
beryllium and thus required careful treatment. The energy resolution (FWHM)
was measured to be 207 eV at 5.9 keV and 393 eV at 59.54 keV, which enabled the four K lines of thulium to be resolved.

As a result of the upward facing detector geometry, the preliminary experiments were performed using the thulium chloride sample placed vertically above the detector face. The aim of the initial experimental work was to determine the counting rate and signal to noise ratio achievable with a range of collimators.

The apparatus was set up as shown in fig 6.4a. The thulium chloride crystals were contained in a thin-walled plastic vial, which was placed upon a 5mm thick perspex table 50mm above the detector face. Output from the detector was amplified and displayed using a Canberra series 35 MCA. Data could be transferred via BBC microcomputer to the University PRIME computers where hard copies of the spectra could obtained using the 'Calcomp' plotter.

The ^241^Am source was mounted in a lead-lined brass barrel holder described in chapter 4. The holder shielded the source and was fitted to take cylindrical lead collimators with a range of exit apertures from 0.5 to 3 mm in diameter. For the initial investigations the aim was to irradiate the thulium chloride crystals uniformly. As a result the collimator was removed from the source holder, giving an effective exit aperture of 12 mm. The source was placed 50 mm from the sample, the volume of which was less than 1 cubic centimetre. The uniformity of irradiation was tested by placing a radiographic film in the sample position for one hour; the film showed uniform blackening across a circular area 20mm in diameter.
Source
Collimator.

Detector Collimator.

Lead Shield

HPGe Detector

Preamp.

Multichannel Analyser.

BBC Micro

H.T. Supply.

PRIME

Figure 6.4a: Apparatus for XRF studies of thulium.
6.4.1. Collimator Selection.

An unfortunate result of the 90° geometry was the resulting energy of the Compton scattered photons. For 59.54 keV gamma rays scattered through 90°, the scattered photon energy is 53.7 keV, which lies between the Kα and Kβ characteristic X-ray lines of Tm. The width of the Compton scattering 'peak' is determined by the range of angles through which photons may be scattered from the sample into the field of view of the detector and is consequently rather broad. As the Compton scattered photons constitute a major source of image 'noise' it is important to design the collimation so as to maximise the ratio of XRF to Compton scattered events.

Although a high signal to noise ratio is required for accurate XRF imaging, it is also important minimise the scanning time; thus the collimator system must designed to obtain the optimum compromise between noise reduction and scanning time. The noise for any single measurement consists of random statistical variations plus the systematic effect of scattered radiation. However, for all set-ups giving counting rates adequate for scanning with 2mm resolution in times less than 24 hrs the Compton in-scattered component was found to be several times larger than the statistical noise.

A series of experiments were carried out with a range of collimators placed immediately above the detector. Each counting run lasted 1000 seconds; the results are shown in table 6.4.1a, the signal is the sum of the background subtracted Kα and Kβ intensities. The background 'noise' values were obtained by performing a simple trapezium fit to the 'background' under the X-ray peaks; more complicated forms of fitting were rejected as inappropriate for these preliminary studies.
### Table 6.4.1a:

<table>
<thead>
<tr>
<th>Detector Collimation</th>
<th>Figure</th>
<th>Total Counts/second</th>
<th>Signal/Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td>5mm diam 10mm long</td>
<td>6.4b</td>
<td>240</td>
<td>4.0</td>
</tr>
<tr>
<td>3mm diam 25mm long</td>
<td>6.4c</td>
<td>80</td>
<td>&lt;&lt;1</td>
</tr>
<tr>
<td>2mm diam 20mm long</td>
<td>6.4d</td>
<td>60</td>
<td>0.6</td>
</tr>
<tr>
<td>2 x 7mm 5mm long</td>
<td>6.4e</td>
<td>200</td>
<td>1.6</td>
</tr>
</tbody>
</table>

The results displayed subtle effects involving the transmission of photons through the collimators and the poor utilisation of emitted photons. In addition, the dose absorbed by the object was quite considerable and yet only the volume of the object in the field of view of the detector collimator contributed to the XRF signal. Photons incident outside the field of view of the collimators contributed to the noise via multiple scattering processes.

A second series of experiments was performed with a 2 mm exit aperture collimator mounted in the holder. The results are shown in table 6.4.1b.

### Table 6.4.1b:

<table>
<thead>
<tr>
<th>Detector Collimation</th>
<th>Figure</th>
<th>Total Counts/second</th>
<th>Signal/Noise</th>
</tr>
</thead>
<tbody>
<tr>
<td>Nil</td>
<td>6.4f</td>
<td>130</td>
<td>1.6</td>
</tr>
<tr>
<td>5mm diam 10mm long</td>
<td>6.4g</td>
<td>17</td>
<td>15</td>
</tr>
</tbody>
</table>

The use of source collimation caused a severe reduction in count-rate, but when used in conjunction with a collimated detector, the gain in signal to noise ratio was very large. The extremely low count-rate in the last result would preclude scanning in time periods of less than 24 hours with...
Figure 6.4b.

Channel number

Figure 6.4c

Counts per channel

Compton scatter peak

Compton Scatter Peak

Tm Kα peak

Tm Kβ peak
Figure 6.4d.

Thulium Kα peak.

Compton Scatter Peak

Thulium Kβ peak

Channel number

Counts per channel

Figure 6.4e.

Tm Kα peak

Compton scatter peak

Tm Kβ peak

Channel number

Counts per channel
Figure 6.4f.

Thulium Kα peak
Compton scatter peak
Tm Kβ peak

Counts per channel
Channel number

Figure 6.4g.

Tm Kα peak
Compton scatter peak
Tm Kβ peak

Counts per channel
Channel number

148c
reasonable image statistics when using the 7.4GBq $^{241}$Am sources.

A further difficulty encountered using the scanning system based upon the uniform irradiation of the sample was that it is necessary to move the source with the object during scanning. This problem could have been solved using a suitable rig. However, the availability of a large area detector enabled a second mode of XRF tomography to be attempted.

6.5 Using the Coaxial HPGe Detector.

In order to perform reconstructive XRF tomography using a line by line technique, it was necessary to obtain a detector possessing the following properties:

(i) Good energy resolution (<2 keV at 60 keV) to separate XRF radiation from the target element from that of other elements and from primary/scattered radiation.

(ii) A large surface area to provide reasonably uniform detection efficiency over an object a few centimetres in diameter placed vertically below the detector.

(iii) A high efficiency at the XRF energies of interest (20-100 keV).

An n-type coaxial HPGe detector was kindly loaned by Dr. P M Walker to perform the XRF scanning. The detector (EG&G Ortec GMX25210) was quite old, and, according to its previous operators at SERC Daresbury, had sustained neutron damage. Despite its history the detector possessed adequate resolution (1.4 keV FWHM at 60 keV) and high efficiency over the range of photon energies used in XRF scanning (~20-100 keV).

Fig 6.5a-c show respectively XRF spectra from Tm, Eu and a combined sample of Tm and Eu, demonstrating the ability of the detector to resolve...
Figure 6.5a Spectrum from Tm.  

Tm Ka peak 50.2 keV  
Tm Kβ peak 58.25 keV  

Figure 6.5b Spectrum from Eu.  

Eu Ka peak 41.2 keV  
Eu Kβ peak 47.5 keV
Counts per channel.

Figure 6.5c: Spectrum from combined Tm/Eu sample.
the Eu $K_a$ line in the presence of Tm. The Eu $K_a$ and Tm $K_a$ are not, however, fully resolved.

The intrinsic full energy peak efficiency of the detector was measured by placing 10 μCi reference sources of $^{152}$Eu and $^{241}$Am (300±5) mm away from the front face. The results of these measurements are shown in fig 6.5d.

To evaluate the crystal dimensions, the detector face was scanned using a collimated beam of 59.5 keV gamma rays from a 7.4 GBq $^{241}$Am source. The beam exit diameter was 1 mm, and the results are shown in fig 6.5e. The scan was performed by mounting the source on the scanner turntable and then taking a single projection across the fixed detector.

A second scan was then performed, in this case the scan was performed along the detector axis, again using 59.5 keV gamma rays. The results are shown in fig 6.5f, which displays a low efficiency region within the detector centre, possibly due to a mechanical structure within the cryostat.

During XRF tomography scanning it was intended to place the test object on the scanner turntable, positioned vertically below the detector. It was thus important to ascertain the uniformity of the detector efficiency for sources of radiation placed upon the turntable. A 60 mm by 60 mm square grid was marked out on a 10 mm deep cylinder of expanded polystyrene placed 85 mm below the detector axis. A 0.37 MBq $^{241}$Am point source was placed on the grid and measurements were taken at 36 positions. Two sets of measurements were taken, at 22 keV and 59.5 keV in order to test uniformity over the range of energies used in XRF scanning.

The result of the 59.5 keV scan is shown in fig 6.5g, the 22 keV scan is shown in fig 6.5h. In each case the plot shows that the recorded count rate does not vary widely over the central region but drops by 25% in the case of 59.5 keV and over 30% for 22 keV radiation for sources positioned
Figure 6.5d: Intrinsinc Photopeak efficiency of HPGe Detector.
Figure 6.5e: Scan across HPGe detector face.

Figure 6.5f: Scan along HPGe detector axis.
Figure 6.5g: HPGe detector uniformity at 60 keV.

Figure 6.5h: HPGe uniformity at 22 keV.
about 30 mm from a position centrally below the detector. In addition the efficiency is generally greater towards the detector front face, falling away at increasing distances along the detector axis. As a result of these features, large diameter objects would suffer a degree of image degradation, however samples with diameters of 30 mm or less placed on the centre of the turntable should not suffer from large inaccuracies produced by detector non-uniformity. The effects of detector non-uniformity were further reduced by using 360° rotation during scanning.

A large area planar detector placed so as to view the sample from vertically above would have provided optimum scanning uniformity, however no detector operating in such a convenient geometry was available.

6.6 Reconstructive Scanning.

To perform reconstructive tomograms using XRF, the same source collimator was used as in reconstructive Compton scattering tomography. The 7.4 GBq $^{241}$Am source was placed in a lead source holder with a primary collimator 3 mm in diameter and 50 mm in length. At the end of the collimator any of a range of secondary collimators in the width range 0.1 - 5 mm could be mounted, to produce either ribbon or pencil beams, depending on the cross-section of the end collimator aperture.

Using a 1 mm diameter end collimator, the beam width was measured as 2.5 mm at 50 mm from the exit orifice using a radiographic film. The beam width and therefore spatial resolution, would, of course, vary across the object, giving the lowest spatial resolution towards the centre. A spatial resolution of approximately 3 mm for a 60 mm diameter object was obtained using this system in the Compton scattering mode (see chapter 5). The
source was mounted in the nylon sleeve holder, positioned to direct the gamma beam parallel to the detector axis and 85 mm below its centre. In this arrangement the beam passes approximately 20 mm below the detector cryostat. The turntable was mounted with its centre of rotation 30 mm behind the beryllium window of the detector. A photograph of the apparatus is shown in fig. 6.6a.

In a preliminary scan the thulium chloride crystals were placed on the scanner turntable and a spectrum was collected using a Canberra series 20 MCA. The SCA window was set to accept photons in the range 45.4 to 63.3 keV to bracket the K X-rays, this enabled a counting rate of over 250 per second to be achieved, although the signal to noise ratio was comparatively poor. The time required for this scan was fifteen hours. The resulting image indicated the basic shape of the sample, but showed a strong edge effect, pixel values being much higher at the perimeter of the sample than at the centre. The reason for this was suspected to be the short mean free paths of 60 keV photons in thulium chloride ($<0.1$ mm), which resulted in the almost total absorption of the irradiating beam within the outer layer of the sample.

A series of aqueous solutions of thulium chloride were prepared and spectra from deionized water and from 2% and 10% solutions are shown in fig 6.6b, c and d. The position of the scattering peak still proved inconvenient, and images obtained continued to display considerably lower pixel values at the image centre compared to at the edges.

To compare the Compton scattering and XRF tomographic techniques, a CsI(Tl) scintillation crystal mounted on a silicon photodiode (designed by Goode [1987]) was scanned employing an initial energy window set around the 90° scattering peak for 59.5 keV gamma rays (47-58 keV) using 60 projections and 33 1 mm raysums. A counting time of 40 seconds per raysum was used, a maximum count-rate of 1800 counts per raysum was obtained.
Figure 6.6a: Photograph of Scanner for XRF Tomography.
Figure 6.6b: Spectrum from deionized water.

Compton scatter peak 53 keV

Figure 6.6c: Spectrum from 2% Tm solution.

Thulium Kα peak.
Counts per channel

Figure 6.6d: Spectrum from 10% Tm solution.
A second scan was performed with the SCA window set around the K X-ray peaks of both I and Cs (26-34 keV). The two images and associated pixel histograms are shown in fig 6.6 e. The 'scattergram' on the right shows an outer ring resulting from scattering in the magnesium oxide packing material placed around the crystal, but no internal structure. The XRF 'fluorogram' (left), however, clearly displays the CsI crystal, but the rapid beam attenuation within the CsI leads to a severe reduction in pixel value at the centre. These images display the elemental sensitivity of the XRF technique, in addition to the problems caused by attenuation of the ingoing beam. The number of fluorescent photons was reduced in this case due to the absorption of the Cs Kβ radiation by the iodine atoms.

6.7 Further Experimental Studies.

A sample of gadolinium chloride was obtained and solutions were prepared in 15mm diameter vials. The 90° Compton scattering peak energy is conveniently above the Kα X-ray energy of Gd, as can be seen from figs 6.7a and b. Fig 6.7a shows a spectrum taken from a 60mm diameter water filled vessel, while fig 6.7b is the spectrum obtained when a 10mm vial of 10% Gd solution was placed at the centre of the water-filled vessel. A discriminator window of 38.7-45.2 keV was used for tomographic scanning, which enabled most of the Compton scattered events to be rejected.

The 10% Gd vial placed at the centre of the 60mm diameter vessel filled with deionized water was scanned over a sixteen hour period, the result is shown in fig.6.7c. As can be seen the water vessel is difficult to discern, indicating the small contribution from scattering. As with Tm scans, the centre of the image exhibits lower pixel values than the outer region. The line section taken through the image centre shown above the image clearly demonstrates a reduction in pixel value at the centre, caused
Figure 6.6e. Scans of CRI detector.
Figure 6.7a: Spectrum from 60mm water container.

Figure 6.7b: Spectrum from container with central 10% Gd solution.
Figure 6.7c: XRF scan of gadolinium solution in water-filled container.
by lack of penetration into the Gd sample. The mean free path of 60 keV gamma rays into a 10\% Gd solution is approximately 1 \text{mm}.

Single projections through the sample are shown in fig. 6.7d, the lower curve shows the effect of subtracting a trapezium shaped background contribution which removes the shoulders caused by Compton scattering. The flattening of the peak due to non-penetration of the centre of the sample is apparent from this projection.

Mutually perpendicular single projections through a pair of 10\% and 5\% Gd solutions are shown in figs 6.7 e and f. The projections show a difference in raysum amplitude in rough agreement with the different concentrations, however the shape of the projections displays the lack of penetration into the samples.

A pair 25\text{mm} diameter sample bottles containing sodium iodide solutions of initially unknown concentration were then scanned using an SCA window set around the K X-ray peak of iodine (27-34 keV). The result is shown in fig. 6.7g. Forty 2\text{mm} wide raysums and 60 projections were taken at 6° intervals in a scanning time of 17 hours. The image shows a strong edge effect which is especially noticeable on the more dense, lower sample and displays artifacting between the two samples, probably due to under determination resulting from lack of beam penetration at many angles.

In addition to the lack of penetration of the primary beam, the secondary XRF radiation is also attenuated by the sample. The effect of self-absorption is reduced in a pure element because the XRF photons are always of an energy below the K-edge of the target material. Despite this, the secondary attenuation is not negligible as can be seen from fig. 6.7h which shows the same sample as above, except that the beam was aimed 5\text{mm} lower on the sample bottle. As can be seen the edge artifact is slightly more severe than in the previous case. The two scans are compared in fig.
Figure 6.7d. Single projections of Gd sample in water vessel.
Figures 6.7e and f: Orthogonal projections from 10% and 5% Gd solutions.
Figure 6.7g: XRF scan of two NaI solutions.

Figure 6.7h: Repeat of above scan with beam lower in samples.
Figure 6.7: Comparison of XRF scans of NAl solutions at different depths.
6.71 which includes line sections through the images, the upper pair correspond to the beam passing through the sample at the lower position.

6.8 XRF Studies Using a $^{152}$Gd Source:

In an effort to reduce the primary beam attenuation, a source of $^{152}$Gd was obtained from Amersham International plc. The source emits gamma rays of 97 and 103 keV and also Eu X-rays in the energy interval 40.8-48.2 keV. A spectrum from the source (approximate activity 7.4 GBq) obtained using the HPGe detector is shown in fig. 6.8a.

Since the 40 keV radiation is above the K-edge for iodine, some means of removing the Eu X-rays had to be found. The most effective means of doing this was to employ a filter of Ce, which has a K-edge at 40.4 keV, and therefore a very high attenuation coefficient for Eu X-rays. Since no Ce was available, a sample of barium nitrate powder was employed (K-edge of barium is at 37.4 keV), which has an attenuation coefficient of 41.6 cm$^{-1}$ at 40 keV and 3.89 cm$^{-1}$ at 100 keV. The filter (about 3mm thick) caused considerable attenuation of the 100 keV photons, but the count-rate obtained was still adequate for XRF scanning.

The source was mounted in a lead collimator holder originally designed for use with a $^{60}$Co source as described in chapter 4. The collimator used was a copper cylinder 130mm in length bored to 3mm internal diameter with the final 25mm bored to 1.5mm diameter. The barium nitrate filter was placed between the source face and the collimator. A second filter to remove Ba X-rays was considered, but was found to be superfluous. A spectrum from the filtered source is shown in fig 6.8b.

The two sample solutions were placed upon the scanner table and a scan was performed using the same parameters as described above. The result is shown in fig 6.8c. As can be seen, the edge effect is less marked, although
Figure 6.8a: Spectrum from Gd-153 source.

Eu X-rays.

Gamma rays (97 & 103 keV).

Channel number

Counts per channel

Figure 6.8b: Filtered Spectrum from Gd-153 source.

Gamma rays (97 & 103 keV).

Counts per channel

Channel number

155a
Figure 6.8c: XRF scan of two NaI solutions using Gd-153.
still considerable on the more concentrated upper sample. A single projection was performed manually which displayed a higher count rate from the less concentrated sample. The pixel values of the lower and higher concentration samples were: 2.5 ± 0.6 and 1.5 ± 0.35 respectively, indicating incorrect determination of relative density.

The sodium iodide solutions were subsequently scanned using transmission CT, using 84 x 1 mm steps and 90 projection at 2° intervals and 2000 counts per raysum. The scanning took a time of 12 hours. The resulting image is shown in fig 6.8d and e, in monochrome and colour scale. The images were obtained in a shorter time than the XRF scans and the solution concentrations were calculated to be 7 and 12% respectively.

To avoid the difficulties associated with high absorption of the beam some low concentration solutions of europium nitrate were then scanned using the 156Gd source. Initially the beam was used without the barium filter to provide a reasonably high count-rate, however a considerable number of scattered X-rays from the source fell into the SCA energy region of interest. This resulted in a very small degree of contrast between solutions of widely differing concentration (a 2% change in pixel value for solutions of 0.37% and 0.19% Eu by mass).

The barium filter was not employed as the primary beam was too strongly attenuated to allow a practical count rate with such dilute solutions. In place of the barium a 1 mm thick tin foil filter was used, which allowed 30% transmission of the 100 keV radiation. The counting rate remained low leading to a thirty hour run being required to obtain acceptable statistics (about 160 counts per raysum from the 0.19% solution). The pixel values for the two solutions were measured as 1.9 ± 0.25 for the 0.37% sample and 1.5 ± 0.3 for the 0.19% solution. The lack of contrast was mainly due to a high background count. Room background could have been reduced by increased shielding around the detector. Using a
Figure 6.8d: Transmission scan of two NaI solutions using 100 keV gamma rays (monochrome).

Figure 6.78e Transmission scan of two NaI solutions using 100 keV gamma rays (false colour).
brighter source would enable less noisy image in the same counting time, and assuming background and scattering remained constant, it would be possible to set up a calibration graph of pixel value versus Eu concentration.

Further experimental work was precluded by a detector malfunction.

6.9 Conclusions.

The technique of XRF analysis has been employed for many years in a range of medical and non-medical applications. However, imaging using fluorescence radiation has not been widely employed on a macroscopic scale.

XRF imaging is possible using both reconstructive and non-reconstructive tomography. With either technique there are several severe limitations if quantitative imaging is required.

Studies using reconstructive XRF tomography have been performed on elements with atomic numbers in the range 53-69, using primary beam energies of 60-100 keV. Although images were obtained successfully using a large-volume coaxial HPGe detector, a range of difficulties were revealed. These were:

(i) An expensive high resolution detector with large active area is required.

(ii) The technique suffers severely from attenuation of both the primary and secondary radiation. The primary beam attenuation can be reduced by using higher incident energies, however the fall off in the photoelectric cross-section with increasing energy would require a considerably more intense radiation beam to provide satisfactory count-rate.

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Compton scattered radiation can lead to excessive data degradation leading to low contrast and poor accuracy.

The attenuation of the relatively low energy XRF radiation is an inherent limitation to the technique and XRF scanning is thus unsuitable for large, dense objects.

The problem of self-attenuation within the sample was clearly shown in the experimental work. Uncorrected scans were unable to provide consistent concentration values for samples of high density. In general it is probable that XRF imaging will only find uses in specialised applications where the distribution of a high Z element within a low Z matrix is required.

The results of attenuation could be reduced in applications where the sample under investigation were flat, or where the scan is performed to study surface features. If a disc shaped object were to be scanned, an intense high energy primary beam could be employed to give reasonably uniform irradiation along the beam path radially through the object. The fluorescent radiation would suffer low attenuation if the disc were fairly thin or if the interrogating beam were near the surface.

In general, more accurate results could be obtained by applying an attenuation corrections to the data, using methods similar to those employed in single photon emission tomography (see chapter 3). However, the correction for attenuation is more complicated in XRF scanning for two reasons:

(i) Two energies are involved (those of the primary beam and fluorescent radiation) and the related attenuation cross-sections vary considerably in the K-edge region.

(ii) Where the fluorescent element is concentrated, much of the attenuation is a result of primary self-absorption. Correction for
this self-absorption is not possible without foreknowledge of the distribution of the element.

The result of the attenuation of the primary beam is to reduce the apparent concentration of the target element deep within the object. If the object is fairly uniform then correction of the final image could be achieved without undue difficulty by post-correcting using correction matrices obtained by performing two transmission CT scans at an energies close to that of the primary and fluorescent radiation.

The effects of Compton scattering on XRF scans could be reduced by employing some form of detector collimation. For the coaxial detector used in these studies, a vee-shaped collimator under the detector axis would limit the acceptance of radiation to the primary beam path, thus reducing the amount of multiple in-scattering. The employment of an MCA with background subtraction facility in place of the SCA would also enable many of the scattered events to be rejected.

Probably the simplest way of reducing the scattered contribution would be to employ a high primary beam energy such that no singly scattered photon could have an energy approaching that of the fluorescent energy of interest. The high energy source would also reduce the effect of primary attenuation, although a very intense beam would be required to produce an adequate count rate as a result of the low value of the photoelectric cross section at high energies.

The XRF technique in non-destructive testing is competing against other techniques, especially transmission CT. The advantage of the XRF technique is that it is element sensitive. However, in situations ideally suited to XRF imaging the transmission CT technique is also highly effective. For a useful XRF yield the photoelectric absorption cross-section must be quite large, in such situations the target materials will also have a high attenuation coefficient, and will thus be easily detectable.
using transmission CT. XRF tomography would be of greatest value in applications where two or more elements of similar atomic number were distributed within an object, in such situations the attenuation coefficient of the elements may be too close to permit spatial distribution by conventional transmission CT. The sensitivity of the transmission CT technique to the presence of a specific element could, however, be improved. The use of two correctly chosen gamma energies or appropriately filtered X-ray beams, could enable high contrast between elements with similar Z, using the large variations in attenuation coefficient immediately above and below a K-edge to distinguish between the different elements.

XRF tomography has been shown to be capable of qualitative imaging of the distribution of a specific element in matter. The method requires considerable refinement before it can be employed as a quantitative tool in non-destructive evaluation. It is the opinion of the author that the applications of the reconstructive XRF tomography technique are limited, and that in most cases a suitable transmission CT technique could be employed more easily.
Bremsstrahlung Tomography.

7.1 Introduction.

Pure beta emitters such as $^{45}$Ca, $^{32}$P and $^{90}$Sr are employed in a variety of industrial and medical applications. The direct detection of these isotopes deep within matter is made difficult by the short range of beta particles (typically a few millimetres in solids). Bremsstrahlung emission comprises one mode of energy loss for fast moving electrons in matter; the range of this radiation substantially exceeds that of the primary beta radiation, thus enabling the detection of beta emitters at depths far exceeding the beta particle range. The measurement of this radiation has been used [Bengtsson 1964] to detect beta emitters in medical applications and imaging has been performed using a gamma camera [Sullivan 1980, Balachandran 1985].

The possibility of carrying out reconstructive emission tomography using bremsstrahlung was suggested by Choudhary [1987] and a scanner originally designed for computerized transmission tomography studies has been employed to perform bremsstrahlung imaging of a $^{32}$P source placed in a water-filled container. The source was imaged successfully, indicating that the technique has some potential for determining the location of pure beta emitters distributed within large assemblies.

To make accurate determination of the absolute source activity, it is necessary to correct for the attenuation of the emitted radiation. The continuous nature of the bremsstrahlung spectrum makes correction for this somewhat more difficult than in the case of single photon emission tomography and a totally satisfactory method of achieving this has yet to be found.
7.2. The Bremsstrahlung Spectrum.

Bremsstrahlung is one of the ways in which fast moving electrons lose energy in passing through matter. The linear rate of bremsstrahlung emission increases approximately with the square of the atomic number of the absorbing medium and linearly with the electron energy. However, over the energy range commonly encountered with beta particles, bremsstrahlung only accounts for a small percentage of the total energy loss, inelastic collisions with the atomic electrons of the absorber being the dominant process below about 10 MeV.

There are two types of bremsstrahlung associated with beta decay, internal and external. Internal bremsstrahlung is emitted as a result of the changing dipole moment of the electron-nucleus system due to the creation and separation of the electron and proton (Babu 1985). The external bremsstrahlung, however, results from the interactions between the electron and the electric field around the nuclei of the medium traversed.

This second process is the more important method of bremsstrahlung production. The X-radiation emitted in the bremsstrahlung process is in the form of a continuous spectrum with a maximum energy equal to that of the electron kinetic energy. Superimposed on this spectrum are characteristic X-ray lines resulting from ionising collisions between the electron and the atoms of the absorber and from self-absorption of the bremsstrahlung via the photoelectric effect. In practice bremsstrahlung is usually of the thick-target variety in which the electrons lose a substantial amount of their energy within the source.

Calculations of bremsstrahlung spectra for a number of beta emitting isotopes have been made by a number of authors, and these have been compared with experimentally determined values (Shivamaru 1984, Gopala 1986). The general shape of the primary beta particle bremsstrahlung spectrum is independent of the absorbing medium and consists of a continuous distribution with a maximum

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energy equal to the maximum energy of the beta spectrum, $E_{\text{max}}$.

7.3. Experimental Studies.

In order to study the bremsstrahlung spectrum, a $^{32}$P liquid source was obtained from Amersham International plc. The source (code PBS.11) was of 74Mbcq (2mCi) initial activity and 2ml in volume. $^{32}$P has a half-life of 14.3 days with a maximum beta energy of 1.709 MeV. A 76mm by 76mm NaI(Tl) detector with an aluminium end-window was employed with a Canberra Series 20 multichannel analyser to record and display the spectrum. The continuous spectrum obtained from the $^{32}$P in its 23mm diameter glass vessel is shown in figure 7.3a. The ratio of detected photons to emitted betas was found to be about 0.03. Surprisingly the spectrum also shows a distinctive peak at about 33 keV. The source was subsequently placed above a vertical geometry planar HPGe detector and the spectrum displayed clear peaks at 32.1 and 36.4 keV, corresponding to the K X-rays of barium as shown in figure 7.3b. The $^{32}$P solution was then placed in a 0.8mm wall thickness plastic vial and the spectrum was recorded again using the NaI(Tl) detector, the result is shown in figure 7.3c, the barium X-ray peak is no longer present. The barium was assumed to be a component of the glass in the original container. This was verified by irradiating the empty glass container with a beam of 59.54 keV gamma rays from an 18.5 GBq $^{241}$Am source; the fluorescence spectrum was detected using the HPGe detector and plainly showed the barium K X-rays.

The spectrum in figure 7.3c. shows a broad peak at 30.7 ± 0.4 keV, with the intensity falling away smoothly at higher energies. The decrease in intensity with energy is quite rapid; approximately 80% of the detected events were of energy 150 keV or less.

The observed pulse height distribution does not correspond directly to
Figure 7.3a.

SPECTRUM OBTAINED FROM $^{32}$P IN GLASS VIAL
11-500 keV

Figure 7.3b:

SPECTRUM OBTAINED FROM $^{32}$P IN GLASS VIAL
USING AN HPGe DETECTOR

Ba KαX-Ray

Ba KβX-Ray
Figure 7.3c

SPECTRUM OBTAINED FROM $^{32}$P IN PLASTIC VIAL
11-500 keV

Figure 7.3d.

$^{32}$P SPECTRA THROUGH INCREASING THICKNESSES OF ALUMINIUM
Figure 7.3e.

BREMSSTRAHLUNG PEAK ENERGY AS A FUNCTION OF WATER THICKNESS
the emitted bremsstrahlung spectrum, partly as a result of the detector response characteristics and partly as a result of the absorbing material between the source and the detector. The primary effect of the material around the source is to attenuate the emitted radiation, and the preferential absorption of the lower energy radiation leads to spectral hardening. Figure 7.3d, shows the effect on the spectrum of placing a series of aluminium plates between the source and detector. The peak of the bremsstrahlung spectrum shifts to higher energies with increasing thicknesses of intervening material. This effect has been employed to determine the depth of a point source beta emitter within matter (Gaur 1982, Choudhary 1987) a graph of bremsstrahlung peak energy as a function of depth in water for $^{32}$P is shown in fig 7.3a.

7.4. Bremsstrahlung Tomography Scanning.

The scanning system used for bremsstrahlung tomography scanning was a modification of the transmission tomography system described in chapter 4. The sample object is placed upon the turntable and scanned across the field of view of the collimated 50mm by 50mm NaI(Tl) detector, mounted in a brass holder of 5mm thickness lined with 5mm of lead for shielding. A variety of 10 mm long lead circular or slot collimators may be attached to the front of the holder to define a narrow field of view and thus determine spatial resolution of the system. The detector output signal is fed to a linear amplifier and thence to a counter-timer via a single channel analyser (SCA). The number of detector counts (the ray-sums) are an approximate line integral along each ray path. A photograph of the apparatus is shown in fig. 7.4a and a block diagram of the system is shown in fig. 7.4b.

Prior to performing a bremsstrahlung tomography scan it was necessary to select the most appropriate energy window. The energy window was chosen so as
Bremsstrahlung Tomography.

Figure 7.42: Photograph of apparatus for
Figure 7.4b.
to maximise the count-rate with the source in front of the collimator whilst minimising the 'background' count taken with no source present. Selection of a wide energy window extending to several hundred keV gave the highest count-rate with the source in front of the collimator, however the count rate with the source outside the field of view of the collimated detector remained high. The background count rate decreased when the source was removed, indicating that a substantial number of photons were able to penetrate the detector shield. As a result of this, the energy window was narrowed to bracket the 20-90 keV interval and, in addition an extra 20mm thickness of lead was placed around the collimator to reduce the number photons leaking through the detector shield.

It should be remarked that the NaI(Tl) detector used was much bigger than necessary for the detection of the relatively low energy bremsstrahlung photons. In an optimised system a much smaller detector could be employed which would greatly increase the signal to background ratio.

The initial scan (fig 7.4c) was made with the 74 MBq/cm² ³²P source in the glass container placed centrally in a 55mm diameter container of deionized water; a 2mm wide by 7mm high collimator was employed with a counting time of 25 seconds for each of the 34 raysums. A line section through the image centre is shown in fig 7.4d, which illustrates a maximum of bremsstrahlung from the object centre with intensity dropping continuously at greater distances from the active solution. The spatial resolution was improved by mounting a 1mm by 7mm high collimator for a second scan as shown in the colour scale image in fig. 7.4e, a counting time of 40 seconds was used. In each case 60 projections were used, overall scanning times were 14.5 and 46 hours respectively.

The maximum range of ³²P beta particles is about 7mm in unit density materials, but the majority are stopped within a few millimetres of their origin; in addition most of the bremsstrahlung emission occurs during the first part of their flight path when their energy is high. Therefore the choice of collimator widths of 1-2 mm is justified in these circumstances.
Figure 7.4c: Bremsstrahlung scan of 32P source, 2mm spatial resolution.

Figure 7.4d: Line section through image above.
Figure 7.4e: Bremsstrahlung scan of $^{32}$P, 1mm spatial resolution.

Figure 7.4f: Bremsstrahlung scan of $^{32}$P in 120mm diameter water container.
The sample bottle was then placed in a 120mm diameter vessel of water and
and a further scan was performed using 60 projections and 60 2mm raysums per
projection. As a result of the hardening of the spectrum the SCA window was set
to accept pulses in the range 30-150 keV. A counting time of 38 seconds per
raysum was used, giving an overall counting time of just over 40 hours, yielding
a maximum of 600 counts per raysum with the source in the centre of the
collimator field of view, falling to 130 counts with the water vessel walls near
to the collimator opening. The result of this scan is shown in fig. 7.4f.

A final scan was performed, this time with two 32p solutions, one of 22±2
MBq in the glass vial, and one of 14.5 ± 2 MBq in the plastic vial. The two
sources were placed diametrically opposite in a 56mm radius plastic container
filled with water. The 2mm wide slot collimator was employed, 62 raysums and 60
projections were used; the total scan time was 54 hours. Because of high room
background the SCA window was set to 20-55 keV; the signal to background ratio
was 2.8:1. Figure 7.4g shows a 3-dimensional plot of the image obtained in this
scan; the relative heights of the two peaks is in broad agreement with the
activities of the two sources. The apparent high values at the edges of the
image are an artifact resulting from the plotting package, a single projection
from the sample is shown in fig. 7.4h.

7.5. Discussion of Results.

The results of the scanning reveals that the technique of reconstructive
bremsstrahlung tomography is capable of imaging pure beta emitters at depths
within matter far exceeding the range of the primary beta radiation. The maximum
depth at which bremsstrahlung imaging is effective varies with source activity,
the nature of the beta spectrum and the material constituting the absorbing
medium, and is also affected by the room background level.

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3-DIMENSIONAL PLOT FROM A BREMSSTRAHLUNG TOMAGRAM OF TWO $^{32}$P SOURCES

Figure 7.4g.
Counts per Raysum

Figure 7.4h

SINGLE PROJECTION FOR TWO 32P SAMPLES IN A WATER FILLED CONTAINER

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To accurately determine the absolute activity of a beta source using bremsstrahlung tomography, however, it is necessary to make correction for a number of factors. In common with other forms of emission tomography, corrections must be made for:

(i) Source-detector solid angle.

(ii) The attenuation of the radiation in the material between the source and the detector.

(iii) The detector efficiency at the energy of the emitted radiation

(iv) The effects of scattering.

The solid angle or geometrical factor problem encountered in bremsstrahlung tomography is similar to that in single photon emission tomography (SPECT) and can be minimised using the opposed detector pair arrangement [Arimizu 1969] and by taking the geometric mean of the two detector outputs for each raysum [Choudhary 1987]. Similar results can be obtained by combining data from opposite views with a single detector using 360 degree rotation.

The problem of attenuation is also encountered in single photon emission tomography, with the effect that intensity of the detected photons does not correspond to the number emitted by the source, leading to incorrect values for both source activity and spatial distribution. A number of methods of correction have been applied to single photon emission tomography; corrections may be applied before [Kay 1975], during [Goit 1972] or after [Chang 1979] reconstruction. In the case of bremsstrahlung tomography, however, the situation is complicated by the continuous nature of the spectrum resulting in spectral hardening through the transmitting medium. As a result, the effective attenuation coefficient within any given volume element will vary with the position of the source in relation to it. This effect can be minimised by reducing the range of accepted photon energies using a single channel analyser, but this is only
achieved with the loss of count rate and the subsequent reduction in statistical accuracy for a given scanning time. Alternatively, since the shape of the bremsstrahlung spectrum is shown to be depth dependent, it may be possible to make ray by ray corrections for self-absorption on this basis. The variation of detector efficiency with photon energy is a further source of systematic error in the bremsstrahlung tomography technique. Clearly the importance of this effect can be reduced by limiting the width of the energy acceptance window of the SCA, and by selecting a detector with reasonably uniform efficiency across this energy range.

An attempt was made to achieve an approximately monochromatic bremsstrahlung system by doping the ~2P solution with thulium chloride. The aim was to set up an energy window bracketing the K X-rays of thulium and to then scan for the fluorescent radiation induced by the beta particles. The XRF count-rate was, however, too low for scanning over reasonable time periods.

The effect of scattering is normally considered to be less important than that of attenuation (Oppenheim 1984); it tends to reduce image contrast and to increase apparent activity when attenuation corrections are applied. Just as for attenuation correction techniques for bremsstrahlung tomography, any correction for scattering is made difficult by the continuous nature of the spectrum. The effect of scattering may be reduced in importance by using tight collimation provided the reduction in absolute detection efficiency can be tolerated.
7.6 Conclusions.

Experiments performed using a simple emission tomography scanner have demonstrated the effectiveness of bremsstrahlung tomography for the imaging of pure beta emitters in matter. The scanning time for sources of about 20 Mbq placed in water filled containers of up to 150mm diameter is rather long using the present apparatus (tens of hours), although this time could be reduced by using a multi-detector arrangement. The range of application of bremsstrahlung tomography may be limited to low density, low atomic number materials since the spectrum of the emitted radiation is peaked at fairly low energy (about 30 keV) and the depth at which a beta emitter may be imaged is limited unless the source activity is very high. Further work on evaluating possible practical applications of the technique is required.

Obtaining accurate quantitative values of the absolute source activity has not yet been possible due to the difficulties of correcting for such factors as attenuation and scattering. Finding solutions to these problems appears non-trivial due to the continuous nature of the bremsstrahlung spectrum.
Recent years have seen an increasing number of imaging techniques based upon the detection of ionizing radiation. Most important among these has been transmission computerized tomography (TCT), which has proved highly successful in both the medical and industrial fields. In TCT the imaged parameter is the linear attenuation coefficient, $\mu$; the dependence of $\mu$ upon density and atomic number enables the technique to provide useful information about the internal structure and constitution of the object of interest.

Most industrial studies in transmission tomography have, to date, used X-ray based scanners designed for medical use. These devices enable rapid imaging of objects of moderate density, however, they exhibit a number of limitations. The main cause of difficulty lies with the polychromatic nature of the X-ray beam. This leads to major problems in quantitative imaging as a result of the beam hardening which occurs as the X-rays pass through the sample under evaluation. The comparatively low photon energies attainable using most X-ray tubes also effectively limits the scanners to objects devoid of dense or high atomic number material. In addition, these devices are also prohibitively expensive to purchase and maintain for wide ranging industrial application by other than very large companies.

Some experiments have been performed using a simple single-beam system employing a gamma ray source. These experiments have further demonstrated the usefulness of a small-scale CT scanner of moderate cost. Many of the studies have been performed using $^{241}$Am, whose principal gamma ray energy is similar to that of the mean energies of medical X-ray machines. Measurements have included the scanning of a system of flowing sand to determine time-averaged voidage.
The use of higher gamma ray energies has enabled scans to be performed on a reinforced concrete sample with useful results.

The major disadvantage with the gamma ray scanner is the extremely long scanning times required (usually several hours). This is mainly due to the very low photon output of available radioactive sources in comparison with X-ray tubes. A second reason is the use of a single detector, with the result that only a very small fraction of the emitted photons are detected. The scanning rate may be substantially improved by the use of a multi-detector fan beam system. The use of such a system would enable scan times of a few minutes per slice and would greatly increase the usefulness of the scanner.

Other recent developments in non-destructive evaluation have used the detection of the secondary radiations produced by the interaction of an irradiating beam within an object of interest to study its internal composition. The various interaction processes between high energy photons and matter enable techniques based upon secondary radiations to be more sensitive than those based upon transmission alone.

A natural extension of these techniques is to perform tomographic imaging using secondary radiations. These techniques are summarized in table 8.

To date, most imaging using secondary radiations has used non-reconstructive tomography, with the plane of interest defined using appropriately designed collimation applied to the source and detector. These techniques require a very long scanning time even with sources of high activity. This is a result of a very low photon utilisation since the vast majority of secondary photons are emitted outside the restricted field of view of the collimated detector.

In an attempt to reduce scanning time, reconstructive tomography has been performed employing an uncollimated high volume detector in association with a tightly collimated primary beam to provide spatial resolution. The raysums
obtained consist of line-integrals of the secondary radiation emitted along the beam path.

Table 8: Modes of Tomographic Imaging Using Primary and Secondary.

<table>
<thead>
<tr>
<th>Imaging Technique</th>
<th>Imaged Parameter</th>
<th>Comments</th>
</tr>
</thead>
<tbody>
<tr>
<td>X-ray Transmission CT.</td>
<td>(\mu(p,Z,E))</td>
<td>Very fast (seconds/scan). Difficulties due to beam hardening.</td>
</tr>
<tr>
<td>(\gamma)-ray Transmission CT.</td>
<td>(\mu(p,Z,E))</td>
<td>Multi-beam scanning quite fast (minutes/scan). Good quantitative accuracy.</td>
</tr>
<tr>
<td>Emission Tomography.</td>
<td>Spatial distribution of administered radiolotope</td>
<td>Attenuation and scattering corrections required.</td>
</tr>
<tr>
<td>Bremsstrahlung Tomography.</td>
<td>Spatial distribution of beta emitter.</td>
<td>As above, but correction methods more difficult.</td>
</tr>
<tr>
<td>Compton Scatter Tomography.</td>
<td>Electron density.</td>
<td>As above, the technique is slow compared to TCT.</td>
</tr>
<tr>
<td>Coherent Scatter Tomography.</td>
<td>Structural effects at molecular level.</td>
<td>As above, technique provides information not obtainable by other techniques.</td>
</tr>
<tr>
<td>X-ray fluorescence Tomography.</td>
<td>Spatial distribution of specific element</td>
<td>As above, technique slow but element selective.</td>
</tr>
<tr>
<td>Photo-nuclear Tomography.</td>
<td>Spatial distribution of specific isotope.</td>
<td>Not yet developed.</td>
</tr>
</tbody>
</table>
This method of scanning has been attempted for both Compton scattered and X-ray fluorescent radiation, in each case with limited success. For both techniques, a tightly collimated beam of low energy (60-100 keV) gamma radiation has been used to interrogate the object, the secondary radiations being detected using a large volume HPGe detector.

The Compton scattering technique has been used successfully to image low density objects with a spatial resolution of about 3mm. The quantitative value of the technique has been poor, and the method was not found to be effective in non-ideal geometries due to the effects of reduced intensity as a result of attenuation and reduced contrast due to the effects of unwanted scatter.

The effects of attenuation could be considerably reduced by the use of a higher primary beam energy, for example from a $^{137}$Cs or $^{60}$Co source. A high source activity would be required to provide adequate count-rate in association with the necessarily narrow beam collimation.

The technique of reconstructive XRF tomography was investigated as a potential method of determining the spatial distribution of specific elements in matter. The method relies on the detection of characteristic X-rays to enable the detection of a single element; a high resolution detector was thus required to resolve elements of similar atomic number. The basic principles of the technique were clearly demonstrated, however, the effects of attenuation and scattering severely limited the application of the technique. As with Compton scattering tomography, the attenuation of the primary beam could be reduced by the use of a higher energy irradiating source, however, the energy of the secondary radiation for a given element is fixed and hence the secondary effects can not be simply dealt with.
The XRF technique has most potential for the imaging of high atomic number materials in a low density matrix. In such applications, however, transmission CT also performs extremely well, and hence the XRF technique is of limited utility except where elements of very similar atomic number need to be distinguished. In such applications the use of multi-energy CT scanning using carefully selected sources or filtered X-ray beams to exploit the strong changes in μ adjacent to absorption edges would provide a faster alternative to XRF tomography.

The detection of pure beta emitters within objects is made difficult by the short range of beta particles. A method of determining the spatial distribution of a pure beta emitter has been investigated by employing the technique of bremsstrahlung emission tomography. This imaging technique uses the detection of the bremsstrahlung radiation produced as the beta particles are slowed down in matter to form raysums from which the beta emitter distribution may be determined. The principle of the technique has been clearly demonstrated, although, as a result of the fairly low energy of the vast majority of bremsstrahlung photons the method is limited to samples with low density and fairly small size (a few tens of centimetres). As with Compton and XRF tomography the ability of the technique to be used for accurate quantitative imaging was limited by the effect of attenuation.

Further work is thus required to devise of correction routines for dealing with the effects of attenuation and scattering in these new imaging modes. Correction is in all three cases complicated by the presence of more than one photon energy.

For the sake of completeness it would be interesting to perform tomographic imaging using detection of the annihilation photons produced by the pair-production process stimulate by an external source. This was not attempted.
due to the low pair production cross-section at available photon energies. In principle the technique would be comparatively easy to perform although scanning time would be exceedingly long.

None of the secondary photon imaging techniques are capable of giving high speed scanning. In contrast to transmission CT, a very small fraction of the emitted photons are detected and this wastage of photons is an inherent limitation on these techniques. As a result, within a given scanning time, transmission CT provides superior results in all but a small range of applications. It is therefore highly probable that none of the techniques are likely to supplant TCT as a technique in non-destructive testing. Within certain specialised fields, however, the secondary photons may be employed to provide useful complementary information.
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