It is a perpetual battle of good versus evil; that is, of signal versus noise.

G.W. Fraser, 1989.

*X-ray Detectors in Astronomy*

Ueber eine neue Art von Strahlen.
(Vorläufige Mittheilung)

W.C. Röntgen, 1895.

*Sitzungs-Berichte der Physikalisch-medizinischen Gesellschaft zu Würzburg, 9*  
132-141
An investigation of the imaging performance of new digital x-ray detectors

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Abstract

Recent advances in detector technology have made possible to develop imaging detectors with enhanced x-ray response and fast readout electronics. In this project a quantitative evaluation of the imaging performance of two of these new x-ray detectors has been made and their possible application in the field of medical imaging has been studied.

The systems investigated are based on position sensitive detectors, and produce digital output in real time. Gaseous and semiconductor detectors that use different modes of x-ray detection (counting mode and integrating mode) and different irradiation geometries (scanned projection and area projection) were considered.

An experimental procedure for the assessment of the imaging characteristics of digital x-ray detectors was set up. The evaluation process was based on widely accepted image assessment and dosimetry techniques. It included both subjective and objective measurements of image quality, and took into account the effect that various digital parameters – such as sampling aperture, sampling distance and quantization – have in the imaging characteristics of the digital radiography system.

Subjective measurements were based on phantom studies, and consisted in the determination of the threshold-contrast detail-detectability properties of the imaging systems under investigation at different exposure levels. Objective measurements included the study of spatial transfer characteristics and resolution using modulation transfer function analysis, and of system noise characteristics such as noise power spectrum and detective quantum efficiency. The results of the experimental measurements have been compared with equivalent figures of system performance obtained with conventional film-screen combinations.

Computer models were used to investigate the parameters that alter the performance of the detectors, and to study the effect that exposure conditions (e.g. filtration and kVp) and photon statistics have upon image quality.
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Introduction

Digital imaging devices were first introduced into medical applications in the late 1960s and early 1970s, following the development of the computerised gamma camera and of computed tomography (CT). However, these techniques are normally reserved for very specialised studies. To this date film-screen radiography is still the most common technique used in medical imaging.

A digital radiography system differs from a conventional x-ray imaging system in that it incorporates some means for conversion of analogue image information into digital format (Smathers and Brody 1985). Ideally, this conversion takes place in conjunction with the detection of the x-rays transmitted through the patient. A data acquisition system performs the conversion and then transfers the digitised image information to a central unit for processing, display and storage.

Advances in different areas of scientific research over the last decade have made possible to develop image transducers based on gaseous and semiconductor detectors, photostimulable phosphors and image intensifier-television camera systems (Charpak 1989, Perez-Mendez 1990, Harrison 1991), but the impact that these developments have made on diagnostic radiography is still relatively small.

The development of new x-ray imaging systems requires concern about the diagnostic value of the images that the new system produces and the dose delivered to the patient during the examination. It is therefore important to discuss briefly why there is a need for low dose imaging systems and which are the criteria normally followed to define and evaluate image quality.

The need for low-dose systems

The extensive use of x-rays in medicine constitutes the largest man-made source of public exposure to ionising radiation. In developed countries about 90% of the radiation dose to the population from all sources excluding the natural background radiation is due to medical x-rays. In the UK alone the annual number of x-ray examinations is of the order of 35 million. This amounted to a collective UK population dose of ~20000 man Sv in 1990 (Shrimpton 1994). About three quarters of
the radiation dose are due to film based radiography, and the remainder corresponds to less frequent examinations such as CT and fluoroscopy.

A report made by the National Radiological Protection Board (NRPB 1990) based on a national survey of patient doses in the UK (Shrimpton et al 1986) indicated that there are wide variations in the dose delivered to the patient for the same type of examination carried out in different hospitals. The survey demonstrated that there is an ample margin for possible dose reductions and indicated which areas of the radiological practice are most susceptible of optimisation. The different methods that could be used for reducing dose to the patient can be divided in two general groups: those that involve changes in radiological procedures and those that involve changes in the equipment used. The development of new x-ray imaging detectors tries to address the problem of dose reduction by the second on these two methods, that is, by improving the relation between the dose delivered to the patient and the diagnostic value of the images that the new system produces.

During the last two decades there have been many developments in medical x-ray imaging equipment. However, these developments have been primarily aimed at improving image quality without much concern in achieving reductions in patient dose. For this reason the NRPB report also called the attention over recent advances in digital x-ray imaging equipment, such as computed radiography and digital spot fluorography, and to the use of scanned beam radiography systems and beam equalisation techniques, which could help to redress the balance between patient dose and image quality.

The problem of image quality

The main goal of the x-ray imaging process is to reveal internal anatomical structures and to show any pathological condition within the body of a patient. The ability to achieve this goal is determined by how well the imaging system can convey the image information to the viewing physician. This is what is normally referred to as image quality. Image quality is, therefore, a complex concept, and cannot be described in a simple way by a single figure of merit.

The approach normally followed in image quality evaluation is to identify the steps in the imaging process that mostly affect the properties of the final image. It is possible in principle to measure objectively all aspects of the performance of an x-ray imaging
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system, such as system noise and spatial transfer characteristics. However, to be able
to estimate the effects of such factors in the detectability of details or on the
perceptibility of patterns it would be necessary to have a model of visual perception
that could describe precisely the role of the human observer (Hay et al 1985). For this
reason, subjective measurements of system performance are normally based on the
detectability of simple details using test objects of a given design.

On the other hand, when a new imaging technique or system is evaluated, it is very
difficult to predict its diagnostic usefulness. Therefore objective measurements of
system performance predominate. The description of image quality in this case is
made in term of concepts such as resolution, contrast and noise, and the device
showing a superior set of physical parameters is normally considered to be better
(Evans 1981). It seems therefore reasonable that a complete investigation of the
imaging performance of a new x-ray imaging detector should involve both subjective
and objective evaluation methods in order to have the best possible description of the
advantages and limitations of the system.

Aims and layout of the thesis.

The aims of the project reported in this thesis have been to evaluate the imaging
performance and to study the possible applications in the field of medical imaging of
two detectors that could be suitable for the development new x-ray imaging devices.
The first system investigated was based on a Multiwire Proportional Chamber
(MWPC), which is a gaseous detector, developed by a group working at the Budker
Institute of Nuclear Physics in Novosibirsk, Russia (Babichev et al 1989). The second
system used solid state detectors known as Charge-Coupled Devices (CCDs), and was
assembled in-house as a prototype system.

The thesis has been organised as follows:

Chapter 1 presents the basic x-ray imaging concepts that are related to the description
of image quality characteristics. The physical parameters associated with the
evaluation of imaging detectors are identified and the effect that changes in these
parameters have upon several image properties are discussed.

Chapter 2 reviews the principles of proportional counter operation and the general
properties of MWPCs. The main components and design characteristics of the digital
radiographic system based on a MWPC are also described in this chapter. The x-ray imaging properties of this system are then investigated in Chapter 3. A subjective analysis of imaging performance of the MWPC system is presented, and the results are compared with those obtained with a modern film-screen combination. The effects that various digital parameters and modes of operation of the multiwire chamber have on the detectability of simple objects are discussed. The spatial transfer and noise characteristics of the system are determined and the relevance of the fundamental properties of the detector on the imaging performance of the system are discussed.

Chapter 4 describes the main principles associated with CCD design and operation and reviews the physical processes associated with semiconductor and scintillation x-ray detection. The implementation and calibration of an x-ray imaging system prototype based on charge-coupled devices are described in Chapter 5. The basic characteristics of the individual components of the system are analysed, and the main operational parameters are determined. Chapter 6 presents the results of the evaluation of the imaging performance of the CCD-based system. The evaluation is mainly based on the determination of the spatial resolution and noise characteristics of the system. A simple subjective evaluation of imaging performance is also presented. Chapter 7 summarises the most important conclusions of the thesis, and presents some suggestions on how this work could be extended in the future.
CHAPTER 1

Basic x-ray imaging concepts

This chapter presents most of the basic x-ray imaging concepts that will be used throughout this thesis. The main technical limitations of film-screen radiography are briefly reviewed and the advantages of digital radiography over traditional methods are discussed. The physical parameters associated with image quality characteristics are identified and the effects that changes in these parameters have upon several image properties are discussed. The theoretical foundations on which different image quality evaluation methods are based are also presented.

1.1. Limitations of film-screen radiography

Film-screen radiography is the most common technique used in medical imaging. This technique has been developed over a long period of time and has reached an advanced level of sophistication and performance. The final aim of any new digital radiographic device would be to eventually replace film-screen (f-s) radiography. It is therefore very important to establish the main limitations and disadvantages of this imaging modality and how digital radiography might help to solve these problems.

1.1.1. Contrast and latitude

Perhaps the most well known limitation of film-screen systems is the non-linearity of response. The shape of the characteristic curve of film means that there should be a compromise between display contrast and system latitude. It is normally assumed (e.g. Nishikawa et al 1987) that the eye requires a difference of 0.01 in optical density for an object to be visible and that a threshold signal to noise ratio of 5 is needed for detectability (Rose 1948). The combined effect of these two requirements is to restrict the detectability of low contrast objects. Using a film with a characteristic curve with a larger gradient (the film gamma) can improve low contrast detectability but it also results in a reduced film latitude.
When a single radiograph is taken it is normally exposed in such a way that the contrast of the anatomical structure of interest falls in the region of maximum gamma. This implies that other structures may well be found in regions of decreased sensitivity and, therefore, would need a second exposure using different mAs and/or kVp if the clinician needs to examine them, with the result of additional dose to the patient.

1.1.2. Film and screen noise

In an ideal situation the noise in a radiographic system should be dominated by the quantum fluctuations of the incoming photon flux. However, both the film and the screen introduce additional fluctuations during the photon detection process. These extra sources of noise (which are normally referred to as screen mottle and film granularity) can significantly reduce the signal-to-noise ratio of the image and degrade the ability of the system to correctly reproduce fine image detail.

A good example of this problem is the effect of noise in the detection of microcalcifications in mammography. In some mammographic film-screen systems film granularity is larger than quantum mottle at low (~1 mm$^{-1}$) spatial frequencies, and practically all systems are dominated by granularity above 5 mm$^{-1}$ (Nishikawa and Yaffe 1985). The presence of film granularity can reduce the signal-to-noise ratio by as much as 50% at high spatial frequencies, thus severely limiting the low contrast performance of the system.

1.1.3. Scattered radiation

Scattered radiation is another factor that can greatly reduce the image contrast and signal to noise ratio. Moreover, in order to ensure optimal film density for a given film-screen combination the scattered radiation can only be reduced at the cost of increased dose to the patient, because of the loss of primary photons that would otherwise contribute to image formation.

The methods most commonly used to reduce scatter in traditional radiographic systems are antiscatter grids and air gaps. These antiscatter techniques often require larger exposure times which in turn needs an increase in dose due to film reciprocity law failure. Furthermore, in certain x-ray procedures (e.g. low kVp) low-angle coherent scattering can be relatively important, and therefore efficient scatter
reduction requires the use of high ratio grids or large air gaps (Muntz et al 1983).

1.1.4. Image assessment

The radiological procedure involves the participation of a human observer, and therefore the process of viewing and interpretation of the image has also to be taken into account. Despite its relatively sophisticated performance, the human eye has the limitation of logarithmic response and is also deficient in terms of contrast sensitivity and grey scale differentiation (Chesters 1982). Only a limited range of contrasts can be accommodated by the human visual system, and even if some information has been captured by the imaging system (i.e. it has been “detected”) it might still not be visible to the human observer.

1.2. Digital radiography

An ideal solution to all the problems described in the previous section would require an increase in the dynamic range and system sensitivity without sacrificing the excellent spatial resolution characteristics of film-based systems. In most cases digital devices have a linear response, which leads in a natural way to improvements in dynamic range. The images they produce are also suitable for employing imaging processing techniques, which may overcome the problem of low contrast detectability. Unfortunately there are disadvantages, particularly in terms of resolution. Nevertheless this kind of system can still prove very useful for certain applications in which high resolution is not critical and low dose is of very important.

Unlike conventional film-screen radiography, digital radiographic systems have the intrinsic feature that image acquisition is completely decoupled from processing and display. Perhaps the most important consequence of this fact is that performance of the imaging system can be separated into those aspects related to the signal to noise ratio (SNR) properties of the image data, and those related to the subjective assessment of the image by a human observer. While the latter can be optimised by digital image processing techniques and an adequate display mode, objective image quality can be determined purely in terms of the SNR of the recorded data.

The origins of digital imaging in medicine can be traced back to the late 1960s, when it was first introduced in radionuclide imaging. However, it was not until the development of computed tomography (Hounsfield 1973) that the potential of systems
that used digital techniques for data acquisition, generation and display of the image was fully realised. A steady expansion of digitally-based imaging methods has followed those initial developments, and nowadays several imaging modalities including Single Photon Computed Tomography (SPECT) and Nuclear Magnetic Resonance Imaging (MRI) are completely digital. Extensive reviews of the most important concepts, image transducers and associated image-acquisition techniques in digital radiography have been made in the past (Smathers and Brody 1985, Harrison 1988). Only a brief description of the major characteristics that distinguish digital radiography systems from traditional film-screen systems will be given here.

1.2.1. Analogue-to-digital conversion

The main characteristic that distinguishes a digital x-ray system is that it incorporates some means for converting the analogue image information into digital format. The analogue-to-digital conversion process involves the discrete sampling of both the spatial distribution (spatial sampling) and of the intensity (quantization) of the incident x-ray photon flux. The spatial sampling is characterised by the sampling aperture of the detector, and determines the number of pixels in the final image. The quantization of the intensity levels depends on the precision of the analogue-to-digital converter (ADC) and determines the number of discrete grey levels that will be present in the digital image. The total number of pixels, the pixel size and the number of quantization levels depend strongly on the type of examination and are normally chosen in such a way that they match certain clinical requirements, such as resolution. Typical values are 1024×1024 for the pixel matrix size, 0.1-0.5 mm for the physical pixel size and at least 8-bit (256 grey levels) for the ADC resolution (Dance 1988).

1.2.2. Dose reduction

Dose in conventional radiography is determined by the sensitivity of the film-screen system and by the film latitude. Most digital radiography systems have a linear response and therefore dose savings can be made by restricting the dose to give only a required noise level. The x-ray spectrum can also be matched to the response of the image receptor in such a way that the lowest dose is used for a given SNR. In this case there are no restrictions in terms of contrast loss, since windowing techniques can be used to accommodate the pixel values in the range of grey scales appropriate for the human visual system.
1.2.3. Image processing

The possibility of using digital image processing methods before or during display is another important feature of digital radiography. Post-processing techniques can also be used to enhance the detectability of certain structures of interest or to extract functional or physiological information.

Temporal subtraction is used to remove time-independent features in a sequence of images and retain only the time dependent information. The best known example of this technique is Digital Subtraction Angiography (DSA). Systematic errors (non-stochastic noise) and artefacts can also be removed using subtraction methods. Dual energy techniques make use of images obtained with two different x-ray spectra. Decomposition methods can be used to find a set of coefficients that describe the photoelectric and Compton contributions in terms of a given set of basis functions. These coefficients can then be used to calculate a weighted sum, so that a given material can be cancelled from the image (Lehman et al 1981).

The methods described above require the use of multiple images and are highly dependent on the data acquisition process. Single images can also be processed using spatial filters, normally applied in Fourier space, in order to enhance their SNR characteristics. For example, edge enhancement or noise reduction can be achieved by applying high-pass or low-pass filters, respectively (Gonzales and Wintz 1987).

During the last few years there has been an increasing interest in the application of image processing methods as a clinical tool for computer aided diagnosis and for quantitative analysis of radiographic images (Doi et al 1993). Methods are being developed for the application of pattern recognition algorithms to the automatic detection of image features and the analysis of digital radiographs.

1.2.4. Exposure techniques

Unlike film-screen systems, digital radiography systems are suitable for using different collimation methods. Some digital radiography systems use an area-exposure technique, as in conventional radiography. This technique has the obvious advantage of short exposure times but it suffers from the same drawbacks of film-screen radiography in terms of the scatter contribution. A completely different approach has been the use of point-scanning techniques, where a finely collimated beam of x-rays
irradiates each pixel in turn. In this case excellent scatter rejection can be achieved, but the benefits are practically offset by the extremely long acquisition times required. The scanning-slit technique, which consists in scanning the patient with a narrow fan beam of radiation, is a compromise between the two previous methods. In this case it is possible to obtain good scatter rejection and at the same time have reasonable scanning times. However, this method does not completely overcome tube loading problems.

Formation of the image can be done in real time by some systems, while others require an intermediate storage and readout step between the acquisition of the image data and the actual display and analysis of the image. Present real-time detectors are mostly based on image intensifier tubes with an internal CsI layer as the image converter. The main disadvantage of this kind of systems is that they have limited spatial resolution. Storage phosphor plates have some interesting properties like better resolution and a large dynamic range, but they do not produce images in real time.

1.3. Subjective assessment of imaging performance

The two main mechanisms that affect the performance of human observers are the presence of visible noise and unsharpness (which is equivalent to a loss of high spatial frequencies) in an image. The combined outcome of these two defects is to impair perception. The effect that they can have on the detection process can be quantified in terms of the threshold contrast \( C_{th} \). \( C_{th} \) is defined as the minimum contrast necessary for the detection of a given image signal, and it is considered to be a measure of the detectability of that signal (Chesters 1992).

Although, in principle, any signal can be made detectable given enough contrast, the minimum contrast required for this to happen varies widely. The threshold contrasts of a wide range of signals can provide a good overall indication of the performance of a given system as an imaging device. Most often details that have a simple shape such as a disc or a square are employed in perception studies (e.g. Webb and Johnson 1990). In a digital system the contrast for such signals is normally defined as

\[
C = \frac{|N_b - N_o|}{N_b} \tag{1.1}
\]

where \( N_b \) and \( N_o \) are the mean number of counts averaged over the background and the
Visible noise has the effect of increasing the threshold contrast, whereas the
unsharpness only affects $C_{th}$ if the size of the object is comparable to or smaller than
the extent of the unsharpness, as it will be seen later in §1.4. There are, of course,
quantitative measures for both noise and unsharpness, namely the Noise Power
Spectrum (NPS) and the Point Spread Function (PSF), respectively, but the
measurement of $C_{th}$ allows direct quantification of the effects that these two
mechanisms have on the detectability of a signal taking into account the role of a
human observer. It also has the added advantage that it is relatively simple to measure
and therefore $C_{th}$ studies can be readily made in the clinical environment, for example,
whereas NPS and PSF (or the corresponding Modulation Transfer Function, MTF)
can only be measured under carefully controlled laboratory conditions.

In order to characterise the performance of an imaging system it is necessary to
measure $C_{th}$ for a wide range of signals and doses. The normal procedure is to use test
objects that have been designed for this purpose. These methods are known as
Threshold-Contrast Detail-Detectability (TCDD) studies, and have been used
extensively in all kinds of systems, from conventional radiography to CT.

Most test objects generate disc-shaped signals in a range of diameters and contrasts.
The traditional method of use is to ask a group of observers to specify the lowest
contrast at which a signal of a given diameter becomes detectable. If only a small
number of observers is used then repeated readings can be done to determine the mean
threshold contrast for each signal.

1.4. Perception models

1.4.1. The Rose-De Vries model

The first models of human visual perception were developed in the early 1940s in an
attempt to explain the existence of a threshold for contrast detectability. Rose (1942,
1948) and De Vries (1943) were the first to introduce the idea that random
fluctuations in the light quanta absorbed in the retina confuse the visual system and
that a minimum (threshold) signal-to-noise ratio is necessary for detection. The signal-
to-noise ratio used by Rose and De Vries was the ratio of an incremental signal
integrated over the area of the detail (see figure 1.1) to the standard deviation of a
comparable area of the background.

Using these concepts it can be shown (Hay 1982) that at the threshold of detectability the contrast is given by

\[
C_{th} = \frac{2k_{th}}{\sqrt{\pi} \sqrt{N_b}} \frac{1}{D}
\]

where \(D\) is the detail diameter and \(N_b\) is the mean number of quanta per unit area in the background. The constant \(k_{th}\) is known as the threshold signal-to-noise ratio, and determines the minimum number of standard deviations by which the image signal must exceed the noise in order to be detected. The value of \(k_{th}\) has not been determined precisely, but values between 2 and 5 are normally used.

Two important conclusions can be reached from equation 1.2. First, it indicates that at a given level of exposure (i.e. \(N_b\) constant) the threshold contrast is expected to change as the inverse of the detail diameter, and second, that \(C_{th}\) would be proportional to the inverse square root of the mean number of background counts for objects of the same diameter.

1.4.2. Geometrical and noise limitations

If there were no geometrical limitations then detectability would be only governed by noise, as indicated by the Rose-De Vries model. However, in a digital system the image contrast is reduced if the diameter of the contrasting detail becomes smaller.
than the pixel sidelength, from pure geometrical reasons.

Figure 1.2 shows a simplified representation of a Contrast-Detail diagram. In this diagram the noise limit is indicated by a straight line with slope minus one, in accordance with equation 1.2, and a given pixel size has been indicated by a vertical line that cuts the "diameter" axis at some point $p$. As the detail diameter is decreased low contrast objects reach the noise limit for relatively large sizes, and they become invisible long before reaching the geometrical limit imposed by the pixel size. Therefore, it can be said that low contrast detectability is noise limited. On the other hand, high contrast objects begin to decrease in contrast when they get smaller than the picture element, and become invisible when they reach the noise limit. In this case geometric limitations selectively affect high contrast details in the image.

![Figure 1.2. Schematic representation of geometrical and noise limitations in a Contrast-Detail diagram (from Evans 1981). The dashed line indicates the threshold contrast as predicted by the Rose-De Vries model.](image)

1.4.3. Interdependence of contrast and resolution

The Rose-De Vries model indicates that the minimum diameter that can be perceived is inversely proportional to the contrast. It has been found, however, that this model is in good agreement with experimental results only for a limited range of object diameters. The experimental measurements yield higher $C_m$ values than the theoretical predictions for both very large and very small diameters.

It has been observed that $C_m$ tends asymptotically to a minimum value for large
diameters. There is still no satisfactory explanation for this behaviour, but it has been suggested that the eye of the observer cannot cope with the whole image at once, and that the effective integration area "seen" by the visual system is smaller than the actual detail area (Barrett and Swindell 1981).

Hay and Chesters (1976) proposed a modified version of the Rose-De Vries model that attempts to explain the discrepancies between the model and experimental results for small details. They proposed that equation 1.2 is only valid if the picture element adequately represents the signal diameter. From the point of view of visual perception the minimum dimension of the picture element is determined not only by the physical characteristics of the imaging system (e.g. pixel size) but also by the unsharpness arising from the refractive components of the eye. The effect of the unsharpness is to increase the effective area over which the eye integrates the background fluctuations. If $a_u$ represents the magnitude of the unsharpness then $C_{ih}$ is given by

$$C_{ih} = \frac{2k_{th}}{\sqrt{\pi}} \frac{1}{\sqrt{N_b}} \left( \frac{D + a_u}{D^2} \right)$$

(1.3)

As before $C_{ih} \propto 1/\sqrt{N_b}$, but now if $D >> a_u$ then $C_{ih} \propto D^{-1}$, as in the Rose-De Vries model, and if $D << a_u$ then $C_{ih} \propto D^{-2}$. That is, if the detail is smaller than the pixel size we would expect the threshold contrast to vary as the inverse square of the detail diameter. Also, when $D << a_u$ we have that $C_{ih} \propto a_u$, which could explain why (at least for small details) a system that produces sharp images is better than a system that produces blurred images.

If the unsharpness is only associated with the imaging device then it can be described by the point spread function of the system. In the more general case unsharpness can arise from the character and structure of the patient, the exposure conditions used ($kVp$, focal spot size, etc.) and viewing conditions.

Equations 1.2 and 1.3 have been derived under the assumptions that the dominant source of noise are the quantum fluctuations of the input photon flux. While this could be a good approximation for a quantum limited system, in practical terms there are always some other noise contributions, and the total noise measured in the image does not necessarily follow Poisson statistics. Harrison and Kotre (1986) derived an expression for $C_{ih}$ based on the model of Hay and Chesters, but they included explicitly the effect of several sources of unsharpness. If the system has linear
response then the threshold contrast is given by

\[ C_{th} = \frac{2k_{eh} p \sigma_z}{\sqrt{\pi} S} \sqrt{D^2 + \sum \frac{a_{ui}^2}{D^2}} \]  

(1.4)

In this expression the total unsharpness is calculated as the sum in quadrature of different factors that can contribute to image blurring. At least three main sources of unsharpness that influence the sampling aperture can be taken into account: the imaging system PSF \( (a_u) \), the blurring effect of the eye \( (a_e) \) and the effect of digitisation \( (a_i) \). It also should be noted that Poisson statistics are not necessarily assumed in this equation, \( \bar{S} \) and \( \sigma_z \) are the measured signal and standard deviation in the image, and \( p \) is the pixel size.

A comparison of the Contrast-Detail curves predicted by the models described previously is shown on figure 1.3. A pixel size \( p = 1 \) has been assumed, and both the detail diameter and threshold contrast are given in arbitrary units. The dimension of the unsharpness has been assumed to be of the order of the pixel size, that is \( a_u \approx p \).

![Figure 1.3](image)

\textbf{Figure 1.3.} Comparison of Contrast-Detail curves as predicted by the models described in this section. The total unsharpness is considered to be of the order of the pixel size, that is \( a_u = p \).

It can be observed that, although the underlying assumptions used to derive equations 1.3 and 1.4 are basically the same, the fact that the total unsharpness is added in
quadrature rather than linearly has the effect of producing a more gradual change between the $1/D$ behaviour for large diameters, and the $1/D^2$ behaviour for small diameters.

1.5. Transfer function analysis

When an object is imaged by a radiographic system its spatial distribution is degraded by the resolution properties of the system, and various sources of noise are superimposed on the image. In addition, because of the discrete sampling inherent to digital systems the signal and noise may be undersampled, thus producing aliasing effects in the image (Giger and Doi 1984, Fujita et al 1985).

Modulation Transfer Function (MTF) analysis has been successfully used in the past to characterise the resolution properties of both conventional and digital imaging devices. Strictly speaking transfer function analysis can only be applied to linear, spatially invariant systems (Barret and Swindell 1981), but while these conditions are not always satisfied they can usually be approximated. Linearity is normally satisfied in most digital systems. Spatial invariance, on the other hand, can never be met because of the finite pixel size and focal spot size.

The different methods that are normally used in the practical determination of the MTF can be broadly classified in two groups: wave methods and spread function methods. The first one is based on the use of spatial sine waves or square waves, while the second method involves the Fourier transform of the measured line spread function.

1.5.1. Wave methods

The theoretical framework for the use of sine wave methods in the determination of the MTF is based on considering a sinusoidal variation of x-ray fluence as the input signal for the radiographic system. A photon fluence distribution of this kind can be described by

$$\Phi(x, y) = \Phi_{av} + \Phi_o \cos(2\pi f x)$$  \hspace{1cm} (1.5)

where $f$ is the spatial frequency (i.e. the inverse of the spatial wavelength associated with the input signal), and is normally given in units of mm$^{-1}$. $\Phi_{av}$ and $\Phi_o$ are the mean and the amplitude of the incident photon fluence. Figure 1.4 shows a computer-
generated image which represents a noiseless sinusoidal distribution of x-ray fluence as described by equation 1.5, with spatial frequency 0.5 mm\(^{-1}\), mean intensity \(\Phi_{av} = 100\) and amplitude \(\Phi_o = 20\).

![Image of simulated sinusoidal variation of x-ray fluence](image)

**Figure 1.4.** Simulated sinusoidal variation of x-ray fluence with spatial frequency of 0.5 mm\(^{-1}\), mean intensity 100 and amplitude 20. The graph corresponds to the horizontal profile at the position indicated in the image.

The information carried by the input signal at a given frequency is described by the modulation of the wave (Johns and Cunningham 1983), which is defined by

\[
M_{in} = \frac{\Phi_o}{\Phi_{av}} = \frac{\Phi_{max} - \Phi_{min}}{\Phi_{max} + \Phi_{min}}
\]  

(1.6)

It can be demonstrated that the output signal of a linear imaging system at unit magnification can also be described by a sinusoidal function with the same spatial frequency as the input, but with a different amplitude

\[
S(x, y) = S_{av} + S_o \cos(2\pi fx)
\]  

(1.7)

The output modulation would be therefore

\[
M_{out} = \frac{S_o}{S_{av}} = \frac{S_{max} - S_{min}}{S_{max} + S_{min}}
\]  

(1.8)

The modulation transfer function of the system is defined as the ratio of the output modulation to the input modulation, normalised at zero spatial frequency.
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\[
MTF(f) = \frac{M_{\text{out}}(f)}{M_{\text{in}}(f)} \cdot \frac{M_{\text{in}}(0)}{M_{\text{out}}(0)} \tag{1.9}
\]

In practical measurements of the MTF the input signal is normally generated for different frequencies with the same modulation, i.e. \(M_{\text{in}}(f) = M_{\text{in}}(0)\) for all \(f\), then

\[
MTF(f) = \frac{M_{\text{out}}(f)}{M_{\text{out}}(0)} \tag{1.10}
\]

which means that the MTF can be determined simply by measuring the modulation of the output signal.

However, it is not easy to generate in practice a sinusoidal variation of x-ray fluence. A way around this problem is to measure the modulation of the system using a square wave (bar) pattern, which is much easier to fabricate, and from these data an estimation of the MTF can be calculated using a correction function based on the Fourier transform of a square wave. If \(T_j(f)\) represents the measured square wave response function (SWRF) then the MTF would be given by

\[
MTF(f) = \frac{\pi}{4} \sum_k B_k \frac{T_j(kf)}{k} \tag{1.11}
\]

(Coltman 1954), with \(k = 1, 3, 5, \ldots\), and \(B_k = 1, 0\) or \(-1\) according to

\[
B_k = \begin{cases} 
(-1)^m \cdot (-1)^{(a-1)/2} & \text{if } r = m \\
0 & \text{if } r < m 
\end{cases} \tag{1.12}
\]

Here \(m\) is the total number of primes into which \(k\) can be factored, and \(r\) is the number of different prime factors in \(k\).

The square wave response method has the advantage that it is very easy to implement in practice, and is also less sensitive to the effect of image noise when compared to spread function methods. This method has been previously applied to the determination of the MTF of digital radiography systems by Drost and Fenster (1984), Hillen et al (1987) and Papin and Huang (1987).
1.5.2. Spread function methods

Spread function methods are based on the fact that there exists a simple relationship between the Line Spread Function (LSF) and the MTF. The LSF of an imaging system is defined as the radiation intensity distribution in the image plane of an infinitely narrow and infinitely long slit (line source) of unit intensity (Rossman 1969). The MTF of a linear, shift-invariant system is given by the one-dimensional Fourier transform ($\mathcal{F}_1$) of the LSF

$$MTF(f) = |\mathcal{F}_1[LSF(x)]| = \left|\int_\mathbb{R} LSF(x) e^{-2\pi i ft} \, dx\right|$$  \hspace{1cm} (1.13)

The LSF is a system transfer characteristic for the special case of a one-dimensional input. However, imaging is in general a two-dimensional process and, therefore, in the general case the MTF is a two-dimensional function. In a rotationally invariant system a single measurement would be enough to characterise the full two-dimensional MTF, but if the system is not isotropic then the two-dimensional character of the imaging system cannot be ignored, and all possible orientations of the LSF should in principle be considered.

Three different methods are commonly used to measure the LSF of medical x-ray imaging systems: the slit, wire and edge techniques. In the slit method the LSF is determined from an image of a very narrow slit. In the wire method an inverted LSF is obtained from the profile of an image of a very thin wire. Finally, in the edge method the LSF is obtained by differentiating the response of the system to a sharp edge (the Edge Spread Function, ESF), that is

$$LSF(x) = \frac{d}{dx} ESF(x)$$  \hspace{1cm} (1.14)

There are three main problems associated with the use of Fourier methods for the determination of the MTF in digital devices: the presence of noise in the radiographic image, the finite size of the sampling aperture and the anisotropy of the sampling process. They are described in the following sections.

1.5.2.1. Effect of noise in the determination of the MTF

Statistical fluctuations will occur in the determination of the LSF as a consequence of
both quantum and detector noise. This in turn results in stochastic fluctuations in the calculated MTF values. Previous work has found that noise can be one of the main problems when Fourier methods are applied to traditional (analogue) imaging systems (Reichenbach 1991).

A theoretical analysis of the influence of noise in MTF measurements using the slit, wire and edge techniques has been made recently by Cunningham and Reid (1992). They have shown that for both quantum and detector noise limited systems the edge method produces the best results at low spatial frequencies, while the slit method is superior at high spatial frequencies.

Differentiation of a noisy scan obtained from the edge data amplifies the high-frequency components of the scans, where the SNR is typically lower. In the simple case of a measured edge spread function $ESF(x)$ that can be written as the sum of an average density distribution $\langle ESF(x) \rangle$ and a noise component $n(x)$ independent of the signal level

$$ESF(x) = \langle ESF(x) \rangle + n(x)$$  \hspace{1cm} (1.15)

it can be demonstrated, using the differentiation theorem for the Fourier transform (Bracewell 1986), that the MTF is given by

$$MTF(f) = \langle MTF(f) \rangle + 2\pi f |S_1(n(x))|$$  \hspace{1cm} (1.16)

The factor $f$ in the second term of equation 1.16 indicates that the contribution of noise to the measured MTF increases proportionally with frequency. Some techniques have been developed to deal with this problem, such as assuming a parametric form for the LSF at higher frequencies, but they can have a significant effect on the estimated MTF values.

Numerical differentiation of the edge spread function data introduces a second problem. A correction factor has to be calculated to compensate for the frequency passband of the finite-element differentiation required to obtain the LSF from the ESF (Cunningham and Fenster 1987). If $MTF_*(f)$ represents the estimated modulation transfer function, then the corrected MTF is given by

$$MTF(f) = \frac{MTF_*(f)}{\text{sinc}[f/(2f_N)]}$$  \hspace{1cm} (1.17)
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where \( f_N \) is the Nyquist frequency associated to the sampled data. Figure 1.5 shows a plot of the correction factor given by equation 1.17 as a function of the normalised spatial frequency \( (f/f_N) \). It can be seen that the greatest relative error, which is 57%, occurs at the Nyquist frequency \( (f = f_N) \) and becomes insignificant below \( f_N/4 \).

![Figure 1.5. Correction factor for finite element differentiation in the determination of the MTF of digital systems as a function of the normalised spatial frequency.](image)

1.5.2.2. Undersampling and anisotropic sampling

Digital imaging devices have the intrinsic feature of possessing a finite pixel size. In practice this means that the image scene data can be undersampled. When traditional spread function methods are applied to digital imaging devices aliasing can cause significant errors in the MTF estimate. The effect of undersampling is to overestimate the values of the MTF at higher frequencies.

The edge technique is useful to calculate a one-dimensional estimate through the centre of the two-dimensional MTF. In analogue imaging devices the edge can be rotated and the scan can always be done perpendicular to the edge, thus providing a full description of the MTF. In digital imaging systems the sampling grid is fixed and cannot be rotated relative to the image, that is, the sampling grid is rotationally anisotropic since the distance between sampling points along lines perpendicular to
the edge varies with the angle of rotation. For this reason it is customary to determine the MTF only in the horizontal and vertical directions.

1.5.3. The extended-edge technique

The problems caused by undersampling were identified by Judy (1976) in early studies of the LSF and MTF of a computed tomography scanner. He proposed an over-sampling technique that can be applied to any digital imaging system in order to eliminate the effects of undersampling in the determination of the MTF.

Spatial oversampling of the pixel matrix can be achieved using an edge placed slightly off the perpendicular to the direction of interest. For example, the determination of the MTF in the vertical direction requires the edge to be placed close to the horizontal direction. In this case an edge response function that has a sampling interval smaller than the pixel size can be obtained from a horizontal image profile that crosses the edge at some point (see figure 1.6).

![Figure 1.6. Schematic diagram of the oversampling technique used in the determination of the edge response function.](image)

The sampling distance $\Delta y'$ in the oversampled scan is related to the pixel spacing $\Delta x$ by

$$\Delta y' = \Delta x \cdot \sin \theta$$

(1.18)

where $\theta$ is the angle between the edge and the perpendicular to the direction of interest. Equation 1.18 indicates that even for a relatively large angle of $\theta = 5^\circ$ a tenfold reduction in the sampling distance can be obtained. If $n_p$ is the number of sample points in the profile then the spectral resolution of the MTF is given by
In the last few years the extended edge technique (Reichenbach et al. 1991, Fujita et al. 1992) has become a widely used method to determine the MTF of digital imaging systems. This method is similar to the oversampling technique proposed by Judy, but uses a procedure called scan-line averaging to reduce noise effects without affecting the MTF estimate. Scan-line averaging consists in registering many edge scans using estimates of the edge location. Noise in the resulting "extended" (oversampled) scan can be reduced by applying a running average to the ESF.

1.5.4. Noise equivalent aperture

The noise equivalent aperture (NEA) is defined as the inverse of the total volume under the surface formed by the square of the two dimensional MTF (Sandrik and Wagner 1982)

\[ a_e = \left[ \int_{-\infty}^{\infty} \int_{-\infty}^{\infty} MTF^2(u,v) \, du \, dv \right]^{-1} \quad (1.20) \]

This quantity expresses the size of the effective area over which an object point is averaged when the image is formed, and is also referred to as effective sampling aperture or average blur size. If the system has circular symmetry then \( a_e \) can be expressed as

\[ a_e = \left[ 2\pi \cdot \int_0^\infty MTF^2(f) \, df \right]^{-1} \quad (1.21) \]

The noise equivalent aperture has been used previously as a figure of merit that folds the MTF into a single number, since it has been found to correlate well with the visual impression of sharpness (Evans 1981).

1.6. Noise analysis

The radiographic imaging process is inherently noisy because of the quantum nature of radiation. As discussed in §1.4, noise sets a fundamental limit to the quality of radiographic images. In addition, the transfer of information through the imaging
system can be degraded by additional sources of noise related to the detection mechanism. The detectability of low contrast signals is dependent upon the imaging system's signal-to-noise ratio, and also the aesthetic acceptability of the images can be strongly influenced by the magnitude and character of the noise superimposed to them (Cowen and Workman 1992).

The methods to analyse the noise in radiographic images are very much the same as those used to analyse other types of random processes. Image noise analysis is normally done using two dimensional noise samples obtained by exposing the detector to a uniform radiation field. A brief description of the most important statistical concepts associated with the determination of the noise properties of an imaging system is given below.

### 1.6.1. Stochastic and deterministic noise

The different noise sources contributing to the final image can be classified as stochastic or deterministic. An ensemble of noise samples \( n(x,y;1), n(x,y;2), ..., n(x,y;i) \) obtained under the same exposure conditions defines a random process, and each member of the ensemble is called a *realisation* or an *instance* of the process. Stochastic noise processes are those in which the noise at a particular point \( (x,y) \) is a random variable over different realisations of the process. It is only the probability characteristics, and not the actual values of the noise at that point, which are determined by the generating process. Quantum noise is usually the most significant contribution to the stochastic noise. This source of noise is due to the random distribution of the incoming x-ray quanta over the area of the image. Random fluctuations related to the detection mechanism can be another important source of stochastic noise. In the particular case of digital radiography the electronic noise associated to the readout system and the digitisation process can also be considered as sources of stochastic noise.

Deterministic noise is related to those processes such that the noise signal \( n(x,y) \) at a particular point in the image is the same for each replication of the observation. The noise contribution in this case is completely specified by the process which generates it. Typical examples of deterministic noise are systematic errors introduced by the instrumentation or signal processing, and variations in the detected signal due to features of the image transducer.
1.6.2. Statistics of stationary ergodic processes

The statistical analysis of a random process is very much simplified if certain conditions are satisfied. A stochastic process is said to be ergodic if the statistics taken over all points of a single realisation are the same as the statistics of the ensemble taken for a particular set of points. For an ergodic process it is thus possible to use space averages over a single image instead of averages over the ensemble. However, in some cases (like Wiener spectrum determination) both space and ensemble averages are needed.

The random process is said to be stationary if the statistics are the same over all areas and are not affected by a shift of origin. This can be a limiting factor in experimental measurements since it could be very difficult to produce large image areas that satisfy this condition. For practical purposes, however, radiographic image noise is normally considered to be both ergodic and stationary.

1.6.3. First order statistics

The first order statistics of a stationary ergodic process are completely specified by its probability density function (pdf), which can be interpreted as the probability that any value of the process \( n(x,y) \) selected at random lies in the range \( n_i \) to \( n_i + dn_i \) (Barret and Swindell 1981)

\[
p(n_i) \, dn_i = \Pr(n_i < n \leq n_i + dn_i)
\]

Quantitative comparison of two different random processes can be simplified if it is made in terms of the moments of the distribution. The mean value \( \bar{n} \) and the variance \( \sigma_n^2 \) are equal to the first and second central moments of the probability density function, and are given respectively by

\[
\bar{n} = \lim_{X,Y \to \infty} \frac{1}{2X} \frac{1}{2Y} \int_{-X}^{X} \int_{-Y}^{Y} n(x,y) \, dx \, dy
\]

\[
\sigma_n^2 = \lim_{X,Y \to \infty} \frac{1}{2X} \frac{1}{2Y} \int_{-X}^{X} \int_{-Y}^{Y} [n(x,y) - \bar{n}]^2 \, dx \, dy
\]

In practice it is customary to work with the noise fluctuations about the mean, which are given by \( \Delta n(x,y) = n(x,y) - \bar{n} \).
1.6.3.1. The Gaussian probability distribution

Virtually all statistical processes in radiological imaging involve just two types of random variables: Poisson distributed random variables and Gaussian random variables. The Gaussian (or normal) distribution applies to continuous random variables which can be described by the probability density function

\[ p(r) = \frac{1}{\sqrt{2\pi\sigma^2}} \exp \left[ -\frac{(r-\bar{r})^2}{2\sigma^2} \right] \]  

(1.25)

where \( \bar{r} \) represents the mean value of the distribution and \( \sigma^2 \) the variance.

This probability distribution is very useful because it has a very important mathematical property. If the observed random variable is the sum of a large number \( N \) of other random variables then the central limit theorem (Papoulis 1965) states that the pdf of the observed random variate approaches a Gaussian distribution as \( N \to \infty \), regardless of the probability density functions of the individual random variables.

1.6.3.2. X-ray source – Poisson statistics

The Poisson distribution is very important in radiographic imaging because it is closely related to the description of the quantum nature of radiation. The probability that a given number \( q \) of quanta are produced over a time \( t \) is given by

\[ p(q, t) = (kt)^q e^{-kt} \frac{e^{+kt}}{q!} \]  

(1.26)

where \( \bar{q} = kt \) is the mean number of quanta and \( k \) is a proportionality constant. The most important property of the Poisson distribution is that the variance is equal to the mean, that is \( \sigma_q^2 = \bar{q} \).

1.6.3.3. X-ray detector – Binomial probability

The interaction of incident x-ray quanta with a detector can be described using a binomial probability distribution. Photon detection can be considered to be a binary process: either the quantum interacts with the detector or it does not. If the probability of interaction is represented by \( \eta \), then the probability of non-interaction is \( 1-\eta \). If \( N_o \)
represents the number of incident quanta then the probability of having exactly \( N_{\text{int}} \) interactions on the detector is given by

\[
p(N_{\text{int}}) = \frac{N_{\alpha}!}{(N_{\alpha} - N_{\text{int}})! N_{\text{int}}!} \eta^{N_{\text{int}}} (1 - \eta)^{(N_{\alpha} - N_{\text{int}})}
\]  

(1.27)

where \( p(N_{\text{int}}) = 0 \) for \( N_{\text{int}} > N_{\alpha} \). It can be shown that the mean and the variance of the number of interacting quanta are

\[
\bar{N}_{\text{int}} = \eta N_{\alpha} \\
\sigma_{N_{\text{int}}}^2 = \eta N_{\alpha} (1 - \eta)
\]  

(1.28)

1.6.3.4. Cascaded probabilities

From the statistical point of view the most simple case of the radiation detection problem corresponds to the situation in which a Poisson-distributed source of x-rays is incident on a detector for which the probability of interaction is binomial. The probability of having interactions is given by the product of the probability of having incident quanta and the conditional probability that if quanta are present, \( N_{\text{int}} \) will interact

\[
p(N_{\text{int}}) = \sum_{N_{\alpha} = 0}^{\infty} \Pr(N_{\text{int}} | N_{\alpha}) \Pr(N_{\alpha})
\]  

(1.29)

If the source is Poisson with mean \( \bar{N}_{\alpha} \) then it can be shown (Barrett and Swindell 1981) that the pdf of the interacting quanta is also Poisson but with mean \( \eta \bar{N}_{\alpha} \). This result is very useful for calculating the variance of cascaded imaging processes.

1.6.4. Second order statistics

1.6.4.1. The auto-correlation function

The probability density function and its associated moments describe the magnitude of the noise, but do not provide any information about its spatial or temporal characteristics. In the general case the noise at any two points in an image can be statistically related, that is, the noise can have certain spatial structure. Second order statistics describe the joint probability that any two points in the image take some
particular values. For a Gaussian process the second order probability function is completely specified by the first joint moment of the noise fluctuations, which is called in this case the auto-correlation function (acf)

\[ C(\Delta x, \Delta y) = \lim_{x,y \to \infty} \frac{1}{2X} \frac{1}{2Y} \int_{-X}^{X} \int_{-Y}^{Y} \Delta n(x, y) \Delta n(x + \Delta x, y + \Delta y) \, dx \, dy \]  

(1.30)

The auto-correlation function describes the spatial periodicities of the image data by multiplying the image by a shifted version of itself and calculating the mean value of that product. The distance by which the image is shifted is called the lag. The result of the calculation will show peaks for those values of the lag for which periodicities exist.

It can be seen from equations 1.24 and 1.30 that the scale value of the acf is given by \( \sigma_n^2 = C(0,0) \). Therefore, the acf includes a description of the total image variance.

1.6.4.2. The Wiener spectrum

The Wiener Spectrum (WS) is essentially an analysis of noise variance in terms of its spatial frequency components, that is, it can be used to describe the noise statistics in the spatial frequency domain. It is also called Noise Power Spectrum (NPS), and for a stationary ergodic process it is defined by

\[ W(u, v) = \lim_{x,y \to \infty} \left( \frac{1}{2X} \frac{1}{2Y} \int_{-X}^{X} \int_{-Y}^{Y} \Delta n(x, y) \exp[-2\pi i(ux + vy)] \, dx \, dy \right)^2 \]  

(1.31)

where the symbol \( \langle \rangle \) denotes an ensemble average. The Wiener-Khintchin theorem states that the Wiener spectrum and the acf are Fourier transform pairs

\[ W(u, v) = \mathcal{S}_2\{C(\Delta x, \Delta y)\} \]  

(1.32)

and therefore the WS and the acf are equivalent measures of image noise. For a Gaussian process either function gives a complete statistical description of the random process. Using equations 1.30 and 1.32 it can be demonstrated that

\[ \sigma_n^2 = C(0,0) = \int \int W(u, v) \, du \, dv \]  

(1.33)

that is, the variance of the random process is equal to the total volume under the 2-dimensional Wiener spectrum. This is an important result from the practical point of
view since it provides an easy way to find the scale value of the measured NPS.

If the image noise is isotropic as well as stationary then the WS can be expressed as a function of the radial distance. Once again, while isotropy can be assumed for film-screen radiographic systems, this is not the case with DR systems, which have to be analysed using a different approach.

1.6.4.3. Wiener spectrum measurements in digital systems

The most common method for the determination of the noise Wiener spectrum of digital systems is the scanning-slit technique (Dainty and Shaw 1974, Wagner 1977). In this method the analysis of a two-dimensional noise sample is reduced to a one-dimensional case by scanning the data using a long, narrow slit. If the Wiener spectrum is measured using a slit scanning in the horizontal direction, the measured spectrum \( \hat{W}_s(u) \) is related to the true spectrum by

\[
\hat{W}_s(u) = \int_{-\infty}^{\infty} W_s(u, v) |T(u, v)|^2 dv
\]

(1.34)

where \( T(u, v) \) is the MTF of the measuring slit. If the scanning slit is infinitesimally narrow and infinitely long then the MTF of the measuring slit is given by a Dirac delta function, \( T(u, v) = \delta(v) \), and \( \hat{W}_s(u) = W_s(u, v) \). In practice the slit has a length \( L \) and a width \( a \), which means that \( T(u, v) = \text{sinc}(au) \cdot \text{sinc}(Lv) \), and

\[
\hat{W}_s(u) = \frac{1}{L} \text{sinc}^2 (au) W_s(u, 0)
\]

(1.35)

Therefore, a one-dimensional central section of the two-dimensional spectrum can be obtained from the measured spectrum by multiplying \( \hat{W}_s(u) \) by the appropriate correction factors. Similarly, the effect of the finite pixel size in a digital device is to multiply the incident analogue Wiener spectrum by the square of the spatial-frequency response of the sampling aperture (Giger et al. 1984). The product is normally referred to as the analogue-presampling Wiener spectrum.

If \( \Delta n_{pk} \) represents a discrete, zero mean, two-dimensional noise sample, a series of synthetic slits of length \( L \) can be obtained by averaging the pixel data in a direction perpendicular to the scanning slit. For a vertical slit of length \( L \) pixels scanned horizontally, the synthetic slits are given by
\[ \Delta n_k = \frac{1}{L} \sum_{j=0}^{L-1} \Delta n_{jk} \]  

The Wiener spectrum estimate can be obtained as the ensemble average of the discrete Fourier transform of \( \Delta n_k \) (Finkelstein and Norton-Wayne 1976), that is

\[ \hat{W}\xi(\mu,0) = \frac{2}{N \Delta x} \left( \sum_{k=0}^{N-1} \Delta n_k \exp\left[-2\pi i u k/N\right] \Delta x \right)^2 \]  

(1.37)

1.6.4.4. Spectral components

The noise power spectrum represents the distribution of the total image variance in the spatial frequency domain. This means that the total Wiener spectrum can be expressed as the sum of the noise power spectra arising from individual noise components. Assuming that two image frames taken under the same exposure conditions have identical noise properties, with statistically independent stochastic noise components, the effect of subtraction on the digital Wiener spectrum is to increase the random noise by a factor of 2. The Wiener spectrum of the subtracted frame (Giger et al 1984) would be therefore given by

\[ W_{s_{\text{sub}}} = 2 \left[ W_{s_{\text{ng}}} - W_{s_{\text{FPN}}} \right] = 2 W_{s_Q} \]  

(1.38)

where \( W_{s_{\text{ng}}} \) represents the Wiener spectrum of a single-frame image, \( W_{s_{\text{FPN}}} \) is the Wiener spectrum of static, frame-independent noise, and \( W_{s_Q} \) is the Wiener spectrum of the stochastic noise components. Similarly, the Wiener spectrum of an image obtained by averaging \( N_f \) frames is given by

\[ W_{s_{\text{avg}}} = \frac{1}{N_f} W_{s_{\text{ng}}} + \frac{(N_f - 1)}{N_f} W_{s_{\text{FPN}}} \]  

(1.39)

This equation indicates that if quantum noise is dominant then frame averaging should reduce noise by a factor proportional to the number of frames averaged. Equations 1.38 and 1.39 can be used to separate the contributions due to stochastic noise using subtraction followed by averaging (Giger et al 1986, Cowen and Workman 1992). The method requires to acquire four single-frame images under the same exposure conditions.
conditions. Two pairs of images are subtracted, and the subtracted images are then averaged in order to compensate for two-fold increase of stochastic noise.

1.7. Signal to noise ratio

A review of the medical imaging literature reveals that different definitions of signal-to-noise ratio (SNR) are used. If a simple detection task is assumed, for example if an observer is required to say whether or not a specified object is present at a certain location in a noisy image, then the signal to noise ratio is defined as the difference of the mean responses over the object and the background divided by their common standard deviation. This is the definition normally adopted in threshold detection studies, as it has been shown to be closely related to human detection performance (Wagner and Brown 1985). Some authors refer to this definition as detail SNR, or Lesion SNR. Throughout this work the notation $dSNR$ will be used to refer to this kind of signal to noise ratio. The acronym SNR will be used in the classical sense, that is we will call a signal to the result of a measurement or series of measurements of some specific quantity such as voltage, number of quanta, number of electrons, etc., and noise to the root mean square deviation of that quantity.

1.7.1. Monoenergetic $dSNR$

The number of x-ray photons per area of the spatial resolution element that are detected at the image plane by an imaging system is known as the carrier signal (Motz and Danos 1978). For a monoenergetic x-ray beam the carrier signal is given by

$$N(E) = A_{pix} \eta(E) \Phi_x$$

(1.40)

where $A_{pix}$ is the area of the spatial resolution element of the imaging system, $\eta(E)$ is the quantum efficiency of the detector at energy $E$, and $\Phi_x$ is the mean incident x-ray fluence at the entrance plane of the detector. If $N_o$ represents the carrier signal averaged over the detail area and $N_b$ is the carrier signal corresponding to an equivalent area of background close to it, then the image signal is defined as the difference

$$\Delta N = |N_b - N_o|$$

(1.41)

In the ideal case (i.e. when the detector does not introduce any additional noise
fluctuations) the statistical fluctuations associated with the carrier signals can be described by Poisson statistics, and the noise associated with the detection of the image signal would be $\sqrt{N_b + N_o}$. In this case the $dSNR$ would be given by

$$dSNR = \frac{|N_b - N_o|}{\sqrt{N_b + N_o}}$$

(1.42)

Using these definitions Motz and Danos (1978) derived an expression of $dSNR$ for a monoenergetic beam in terms of the basic quantities illustrated in figure 1.7. Assuming that there is no scatter contribution, $dSNR$ can be expressed as

$$dSNR = \frac{\sqrt{A_{\mu a} \eta(E) c_E x e^{-\mu_b(E) L} (1 - e^{-\delta})}}{\sqrt{1 + e^{-\delta}}}$$

(1.43)

where $\delta = L_a |\mu_a(E) - \mu_b(E)|$, $c_E$ is the exposure to photon fluence conversion factor at energy $E$, and $X$ is the entrance exposure to the patient. This equation indicates that in a quantum limited system the $dSNR$ should be proportional to the square root of the entrance exposure. Also, for fixed exposure conditions the only way to improve $dSNR$ would be to increase the quantum efficiency of the detector.

Figure 1.7. Schematic representation of a simple radiographic imaging task. A contrasting detail of thickness $L_a$ and linear attenuation coefficient $\mu_a$ is embedded in a medium of thickness $L$ and attenuation coefficient $\mu_b$. 

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1.7.2. Broad spectrum $dSNR$

A detailed description of $dSNR$ in a realistic x-ray study requires that not only the energy response of the detector, but also the difference between modes of x-ray detection (counting or integrating detectors), the scatter contribution and the effects of a broad-spectrum x-ray energy distribution are taken into account. When polyenergetic x-ray spectra are used to image a contrasting detail the energy dependence of the detector response contributes to the detected contrast due to the difference in the energy modulation of the incident x-ray beam depending whether it passes through the detail or beside it (Sandborg and Alm Carlsson 1992).

In a practical detection experiment the number of detected quanta is large. The carrier signal is the result of the sum of many independent quantum contributions and, according to the central limit theorem, can be described by a random variable $S$ which follows a Gaussian distribution. When the signals $S$ arise from a photon fluence distribution $\Phi_s(E)$ and the detector responds to each detected quantum with a response $\psi(E)$ then the mean carrier signal and the signal variance (Tapiovaara and Wagner 1985) can be expressed as

$$
\bar{S} = \int_E N(E) \psi(E) dE
$$

$$
\sigma_s^2 = \int_E N(E) \psi^2(E) dE
$$

Since the distribution of the carrier signals corresponding to the contrasting detail ($S_o$) and the background ($S_b$) are the result of mutually independent quantum contributions the distribution of the image signal $\Delta S = |S_b - S_o|$ is also a Gaussian. According with the previous definition, the $dSNR$ would be given by

$$
dSNR = \frac{|S_b - S_o|}{\sqrt{\sigma_b^2 + \sigma_o^2}}
$$

Substituting the corresponding expressions for the carrier signals in 1.41 we obtain

$$
dSNR^2 = A_{pix} \frac{\left[\int_E [\Phi_b(E) - \Phi_s(E)] \eta(E) \psi(E) dE\right]^2}{\int_E \left[\Phi_b(E) + \Phi_s(E)\right] \eta(E) \psi^2(E) dE}
$$

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In a photon counting detector the quantum response $\psi(E)$ is a constant, and therefore cancels out from equation 1.42. In energy integrating detectors the quantum response is a random variable $m(E)$. It can be demonstrated (Mandel 1959) that in this case

$$d\text{SNR}^2 = A_{\text{pix}} \left[ \int E \left[ \Phi_b(E) - \Phi_o(E) \right] \eta(E) \langle m(E) \rangle dE \right]^2 \right] \int E \left[ \Phi_b(E) + \Phi_o(E) \right] \eta(E) \langle m^2(E) \rangle dE$$

(1.47)

where $\langle m(E) \rangle$ and $\langle m^2(E) \rangle$ are the mean and the mean square response for a detected quantum of energy $E$. A well known example of this type of detector is a phosphor screen, in which $m(E)$ represents the number of emitted light quanta that follow the absorption of one x-ray photon. Equation 1.43 indicates that $d\text{SNR}$ depends in a complex way on the composition, density and thickness of the contrasting detail, on the absorption properties of the imaging detector, and on the statistical properties of the detector response.

### 1.8. Detective quantum efficiency

While quantum efficiency can be used to describe the input-output relationships of an imaging system in terms of quantity, it does not provide information about the transfer of noise through the system. It has already been pointed out in the previous section that the information content of an image is primarily determined by the signal to noise ratio, therefore a complete description of the performance of an imaging detector should include the effects of the transfer of both signal and noise through the system.

The concept of Detective Quantum Efficiency ($DQE$) was developed in the late 1940s in an attempt to find a system performance parameter that could be used to compare different imaging modalities (Jones 1949). Since the noise associated with the input quantum fluctuations (the input noise level) provides an absolute scale with respect to which the output noise can be compared, the classical definition of $DQE$ is given simply as the square of the ratio of the output SNR to the input SNR

$$DQE = \frac{SNR_{\text{out}}^2}{SNR_{\text{in}}^2}$$

(1.48)

As such, $DQE$ represents the fraction of the incoming photons that would have to be
detected by an "ideal" detector (i.e. a detector that does not introduce any additional noise fluctuations) in order to produce the same signal to noise ratio as the one actually observed in the real detector. For this reason the square of the output signal to noise ratio is normally referred to as the number of noise equivalent quanta (NEQ).

### 1.8.1. Energy modulation

The noise in the output image produced by an x-ray imaging system that uses an energy integrating detector is primarily determined by the quantum fluctuations corresponding to the detection of individual x-ray photons. Swank (1973b, 1974) was the first to demonstrate that the signal to noise ratio in the output image is given by

\[
SNR_{\text{out}} = \sqrt{I_x \eta_0 \Phi_x}
\]

if it is assumed an incident monoenergetic beam and that the number of x-rays detected in a given time interval follow a Poisson distribution. In this equation \(I_x\) is an statistical factor associated with the fluctuations in energy deposition per absorbed x-ray, and is known as the Swank factor. The product \(A_n = I_x \eta_0\) represents the effective quantum utilisation of the detector and is sometimes called the noise equivalent absorption.

For a monoenergetic incident beam of energy \(E\), the normalised absorbed x-ray energy distribution \(p(E, E_0)\) represents the probability that an incoming photon of energy \(E_i\) will deposit an energy \(E\). It can be shown (Swank 1973b, Dick and Motz 1981) that all the information transfer properties of the detector can be derived from the moments of this absorbed energy distribution (AED). The \(j\)-th moment of the AED is given by

\[
M_j(E_0) = \int_0^\infty p(E, E_0) E^j dE
\]

The zero-th moment of the distribution is equivalent to the quantum efficiency of the detector, that is \(\eta_0 = M_0\). The first moment is related to the average energy absorbed per x-ray interaction, and is a measure of the energy signal efficiency of the detector. The energy absorption efficiency is therefore given by

\[
A_E(E_i) = M_1(E_i)/E_i
\]

Another useful parameter is the fractional energy absorption efficiency, which is defined as the ratio of the energy absorption efficiency and the quantum efficiency.
Basic x-ray imaging concepts

\[ A_x(E_i) = \frac{A_x(E_i)}{\eta_Q} \] \hspace{1cm} (1.52)

If the energy of the incident x-ray quanta is above the K-edge of the detector then some energy can be lost through x-ray fluorescence, thus below the K-edge \( A_x(E_i) \approx 1 \) and above the K-edge \( A_x(E_i) < 1 \). Finally, the noise equivalent absorption is given in terms of the first and second moments by

\[ A_n = \frac{M^2}{M_2} \] \hspace{1cm} (1.53)

For a polyenergetic x-ray spectrum \( \Phi_x(E_i) \) the average moments of the absorbed x-ray energy distribution \( \overline{M}_j \) can be calculated as a convolution of the corresponding moments of each monoenergetic component \( E_i \) with the incident x-ray spectrum

\[
\overline{M}_j = \int_0^E E^j \int_0^E \Phi_x(E_i) \ p(E, E_i) \ dE_i \ dE \\
= \int_0^E \Phi_x(E_i) \ M_j(E_i) \ dE_i \] \hspace{1cm} (1.54)

The average statistical factor \( \overline{I}_x \) and the noise equivalent absorption for the x-ray spectrum can then be calculated with equation 1.53 using the average moments of the distribution.

1.8.2. Frequency modulation

The description of DQE in the spatial frequency domain requires that the imaging system is considered as consisting of a series of statistically independent channels, each one with its own signal and noise transfer characteristics (Fu and Roehrig 1984). Signal transfer modulation is described by the system MTF, while noise transfer is characterised by the noise Wiener spectrum. For a linear system (Hillen et al 1987, Cunningham and Reid 1992) the frequency-dependent number of noise equivalent quanta can be expressed as

\[
NEQ(f) = \frac{\langle S \rangle^2 \ MTF^2(f)}{W_S(f)} = \frac{MTF^2(f)}{W_{ASIS}(f)} \] \hspace{1cm} (1.55)

Here \( \langle S \rangle \) is the measured signal amplitude at zero spatial frequency (large area average) for a given exposure level, and \( W_{ASIS}(f) \) refers to the Wiener spectrum of the relative output noise fluctuations. If the input noise follows a Poisson distribution then
SNR^2 is given by the incident x-ray fluence, \( \Phi_x \). In this case the frequency dependent DQE is given by

\[
\text{DQE}(f) = \frac{\text{MTF}^2(f)}{\Phi_x W_{SSS}(f)}
\]  

(1.56)

This equation is important from the experimental point of view since it indicates that DQE can in principle be determined from experimentally measured quantities.

1.8.3. Multi-stage imaging systems

If the digital radiographic system involves several steps in the process of image formation then the signal and noise transfer characteristics through the system can be analysed in terms of elementary amplification and scattering processes.

Rabbani et al (1987) have developed a model that uses multivariate moment generating functions to express the input-output relationships of noise power spectra through a series of cascaded steps. The model considers the information carriers (e.g. x-ray or light photons) as a series of random impulses. Only three elementary processes are used to describe the propagation of signal and noise through the system: binary selection, amplification and scattering. If \( \Phi_{r-1} \) and \( W_{S_{r-1}}(f) \) represent the mean quantum fluence and the noise Wiener spectrum at the input of the \( i\)-th stage of the system, and \( \Phi_i \) and \( W_{S_i}(f) \) represent the corresponding output, then the propagation of signal and noise for each elementary process are as follows:

**Binary selection.** A binary selection process is characterised by a probability \( (1 - \eta) \) that an input quantum will appear at the output, \( \Phi_i = (1 - \eta) \Phi_{r-1} \), where \( \eta \) represents the probability of interaction. The Wiener spectrum is given by

\[
W_{S_i}(f) = (1 - \eta)^2 W_{S_{r-1}}(f) + \eta(1 - \eta) \Phi_{r-1}
\]  

(1.57)

**Scattering.** The scattering process is characterised by the modulation transfer function of the stage, \( \text{MTF}_i(f) \). The photon fluence is preserved, that is \( \Phi_i = \Phi_{r-1} \) and the Wiener spectrum is given by

\[
W_{S_i}(f) = [W_{S_{r-1}}(f) - \Phi_{r-1}] \text{MTF}^2(f) + \Phi_{r-1}
\]  

(1.58)

**Amplification.** An amplification process is described by the mean gain \( g_i \) and the gain
variance $\sigma^2_{\delta_i}$. The output photon flux increases by a factor $\bar{g}_i$, $\Phi_i = \Phi_{i-1}$, and the Wiener spectrum is given by

$$ W_{S_i}(f) = \bar{g}_i^2 W_{S_{i-1}}(f) + \sigma^2_{\delta_i} \Phi_{i-1} $$  \hspace{1cm} (1.59)

Using these expressions Cunningham et al (1994) obtained a generalised expression for the spatial-frequency dependent $DQE$ of a $N$-stage system with a Poisson distributed input, which is given by

$$ DQE_{i,N}(f) = \left[ 1 + \sum_{i=1}^{N} \left( \frac{1 + \varepsilon_{\delta_i} |MTF_i(f)|^2}{P_i(f)} \right) \right]^{-1} $$  \hspace{1cm} (1.60)

where

$$ P_i(f) = \prod_{j=1}^{i} \bar{g}_j |MTF_j(f)|^2 $$  \hspace{1cm} (1.61)

and $\varepsilon_{\delta_i}$ is the Poisson excess factor, which is defined as

$$ \varepsilon_{\delta_i} = \frac{\sigma^2_{\delta_i}}{\bar{g}_i} - 1 $$  \hspace{1cm} (1.62)

Equations 1.51 and 1.52 can be used to obtain an expression for the $DQE$ of a system with an arbitrary sequence of stages that can be described in terms of amplification or scattering mechanisms, given only the average gain $\bar{g}_i$, the Poisson excess term $\varepsilon_{\delta_i}$ and the modulation transfer function $MTF_i(f)$ of each stage. The particular values that these parameters take according to the type of process are given in table 1.1. Notice that in this model binary processes are considered a particular case of amplification.

<table>
<thead>
<tr>
<th>Process</th>
<th>Gain</th>
<th>MTF</th>
<th>Type</th>
<th>$\sigma^2_{\delta_i}$</th>
<th>$\varepsilon_{\delta_i}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Deterministic</td>
<td>0</td>
<td>1</td>
<td>Poisson</td>
<td>$\bar{g}_i$</td>
<td>0</td>
</tr>
<tr>
<td>Amplification</td>
<td>$g_i$</td>
<td>1</td>
<td>Poisson</td>
<td>$g_i$</td>
<td>0</td>
</tr>
<tr>
<td>Scattering</td>
<td>1</td>
<td>$MTF_i(f)$</td>
<td>Binary</td>
<td>$g_i(1 - g_i)$</td>
<td>$-g_i$</td>
</tr>
</tbody>
</table>

Table 1.1. Values of the gain variance ($\sigma^2_{\delta_i}$) and Poisson excess factor ($\varepsilon_{\delta_i}$) for amplification and scattering processes in a multi-stage system.
1.9. Monte Carlo methods

The development of new x-ray imaging detectors requires the optimise the parameters that are related to image quality while maintaining low radiation dose to the patient. Computer simulation has proved to be a very useful tool for examining the merits of various design strategies and for finding the optimal choices of detector parameters, such as composition and thickness (Bencivelli et al 1991, Boone 1992).

The Monte Carlo method is probably the most common technique employed for the study of radiation transport in medical radiation physics. An extensive review of the Monte Carlo techniques as applied to medical radiation physics problems has been recently made by Andreo (1991).

In this thesis all the simulations that involve radiation transport were developed using the Electron-Gamma Shower v.4 (EGS4) code system (Nelson et al 1985). EGS4 is a general purpose package for the Monte Carlo simulation of the coupled transport of electrons and photons in an arbitrary geometry. EGS4 is not a stand-alone Monte Carlo code, but a set of subroutines that have to be linked to a series of user-written routines, which are called the user code. The user code contains a description of the geometry of the problem and the scoring algorithms needed to extract the relevant quantities from the transport subroutines. In a simulation each particle is followed until it reaches a pre-determined energy limit (cut-off) or a geometrical boundary defined by the user.

All the results reported in this thesis are based on very simple slab or cylindrical geometries. Accurate determination of the x-ray energy deposition required the introduction of two important modifications into the Monte Carlo code. First, correct energy scoring was ensured by including the parameter reduced electron-step transport algorithm (PRESTA) developed by Bielajew and Rogers (1987) into the EGS4 code. This algorithm was developed to facilitate the selection of the appropriate electron-step sizes required to avoid energy scoring artefacts when working at low energies.

The second modification was the introduction of a method for the correct K-edge sampling in compounds, according to the algorithm developed by Del Guerra et al (1991). In the standard version of the EGS4 system the equivalent K-edge of a compound is assumed to be the weighted average of the K-edge of each element in the
compound. This means that the energies of the fluorescent photon and the photoelectron are not sampled correctly, which in turn leads to inaccurate energy scoring in the compound material when the energy of the incident photons is close to any one of the K-edge values.

The pseudo-random generator used by EGS4 is based on an algorithm developed by Marsaglia et al (1990). The same algorithm was used for the sampling routines in the user codes and in some other simulations not involving radiation transport. The cross section data used by EGS4 are taken from Storm and Israel (1970) and the K-edge fluorescent yields are based on those reported by Lederer et al (1968).
Chapter 2

A MWPC for digital radiography

2.1. Introduction

The physics of gas-filled particle and x-ray detectors was very intensively studied in the first half of this century, before first scintillator detectors in the 1950s and then semiconductor detectors in the 1960s replaced them in many areas of nuclear physics research.

A new impetus to the use of proportional counters was given by the particle physicists in the late 1960s after the invention of the multiwire proportional chamber (MWPC) by Georges Charpak and co-workers (1968). The MWPC is a position-sensitive variant of the well known single-wire proportional counter.

The pioneering work on x-ray imaging with multiwire chambers was done by Victor Pérez-Méndez in the early 1970s when he began to use MWPCs for radio-isotope imaging (Kaplan et al. 1973). Since then MWPCs have been used for biomedical applications as diverse as x-ray diffraction, positron-emission tomography, nuclear medicine and β-autoradiography (Bateman et al. 1987, Charpak 1989).

Over the last 10 years a group from the Budker Institute of Nuclear Physics in Novosibirsk has been developing a new digital radiographic system based on a MWPC (Babichev et al. 1989, 1991a), the Siberian Digital Radiographic Device (SDRD). The system is based on a linear MWPC and uses a scanned-projection irradiation geometry to form 2-dimensional images.

In this chapter the basic principles of proportional counter operation and the general characteristics of multiwire proportional chambers are briefly reviewed. The main components and design characteristics of the SDRD are described, and the most important properties of the linear multiwire chamber on which the system is based are discussed.
2.2. Basic principles of proportional counter operation

X-ray proportional counters consist of an arrangement of conductive electrodes enclosed in a pressurised gas container. The operation of a proportional counter is based on a series of well defined physical processes. First, the incoming x-ray photons are absorbed and converted into charged particles. These charged particles move through the host gas towards the electrodes, creating further ionisations on their way. Finally, the negative ions are amplified in a region of enhanced electric field. Signals induced in the electrodes can provide information about the energy and interaction position of the absorbed x-ray photons.

![Diagram of different regions of operation in gas-filled detectors](image)

**Figure 2.1.** Different regions of operation in gas-filled detectors. The pulse amplitude is plotted as a function of the applied voltage for two events that deposit different amounts of energy in the detector (from Knoll 1989).

Figure 2.1 shows a schematic diagram of the four possible regions of detector operation for a gas counter operated in pulse mode in terms of the applied potential. The lines represent the pulse amplitude for two events that deposit different amounts of energy in the detector. In the region of ion saturation charge recombination is prevented, but the ions are not given sufficient energy to produce further ionisation. The applied voltage must exceed a certain threshold ($V_T$) in order to start the process of avalanche formation. In the proportional region the collected charge (i.e. the pulse amplitude) is proportional to the original number of electron-ion pairs created by the primary ionization event. At higher voltages pulse amplitude still increases with energy (limited proportionality region) but linearity is lost.
2.2.1. X-ray conversion

At energies of clinical interest (between 10 and 100 keV) x-rays interact with the gas molecules mainly via the photoelectric effect, with the immediate release of a primary photoelectron followed by a cascade of Auger electrons and/or fluorescent photons. The product of most primary ionisation events is a localised cloud of low-energy secondary electrons in the region immediately after the inlet window, which is called the conversion or drift space. Ignoring fluorescence, the number of electrons in the initial cloud produced by the absorption of an x-ray of energy $E$ can be written as

$$N_e = \frac{E}{w_i}$$

(2.1)

where $w_i$ is the average energy needed to create a secondary ion pair, which depends on the gas composition (e.g. $w_i$ = 26.2 eV for Ar and 21.5 eV for Xe). The creation of secondary ion pairs cannot be considered a series of independent events and therefore the fluctuations in $N_e$ are actually smaller than what would be expected from Poisson statistics. The variance in $N_e$ is given by

$$\sigma^2_{N_e} = FN_e$$

(2.2)

where $F$ is an empirical factor (known as the Fano factor) which is introduced to match the experimentally measured variance. The Fano factor has values ranging from 0.17 (Ar, Xe) to 0.32 (CO$_2$) in commonly used proportional counting filling gases.

2.2.2. Gas multiplication

Once created, the electron cloud begins to drift towards the nearest anode under the influence of the applied electric field. As the electrons get closer to the high field region they gain sufficient energy between successive collisions to excite or ionise the molecules of the filling gas. This process of ionisation by collision forms the basis of avalanche multiplication in proportional counters, since the electrons released during the ionisation process produce further electron-ion pairs as the electron avalanche propagates towards the anode.

The positive ions created during the avalanche process move more slowly than the electron cloud and gradually diffuse, drifting towards the cathode. The space charge represented by these positive charges can distort the electric field near the anode wire.
The net effect of the positive-ion sheath is to reduce the electric field at small radii, and, consequently, the magnitude of the output pulse.

The number of electrons collected at the anode per electron in the x-ray induced charge cloud is called the gas gain $G_s$. In the general case of a non-uniform electric field the gas gain for a drift from $x_1$ to $x_2$ (Knoll 1989) is given by

$$G_s = \exp\left[\int_{x_1}^{x_2} \alpha_T(x) \, dx\right]$$  \hspace{1cm} (2.3)

where $\alpha_T(x)$, which is called the first Townsend coefficient, is the inverse of the mean free path for ionisation, and represents the number of ion pairs produced per unit length of drift.

If the dependence of the Townsend coefficient on the electric field distribution is known then it is possible in principle to compute the multiplication factor for any field geometry. An approximate expression for $G_s$ in a coaxial cylindrical chamber (Sauli 1977) is given by

$$G_s = ke^{CV_e}$$  \hspace{1cm} (2.4)

where $k$ is a constant that depends on the gas mixture, $C$ is the capacitance per unit length of the system and $V_e$ is the applied voltage. This expression is obtained by assuming that $\alpha_T(x)$ depends linearly on the energy of the electrons and that $V_e \gg V_T$, where $V_T$, which is known as the threshold voltage, is the voltage at which avalanche multiplication starts.

The pulse magnitude $P = G_s N_e$ is proportional to the original x-ray energy provided that $G_s$ does not become too large. The reason why there could be loss of proportionality is that the multiplication factor cannot be increased indefinitely. As the applied anode voltage is increased first strict proportionality is lost due to space charge effects near the anode wire, and finally $P$ becomes independent of $E$ in what is called the Geiger-Müller regime.

### 2.2.3. Gas filling

The choice of gas filling in a proportional counter is mainly determined by the specific experimental conditions in which the detector will be used. Sauli (1977) gives a
detailed description of the most important requirements that can conflict when choosing the gas for a particular application.

Noble gases are normally used as the main component of the gas mixture in proportional counters since gas multiplication occurs at much lower electric field strengths than in complex molecules. However, excited and ionised atoms of the gas are produced during the avalanche process. In particular, noble gas atoms emit ultra violet (UV) light very efficiently when excited by electron collisions. Furthermore, this process starts at lower field strengths than charge multiplication.

The UV emission sets a low limit to the attainable gain in a gas counter due to the efficiency with which it can cause photo-emission from the cathode surface and thus initiate a continuous discharge. A small proportion of a polyatomic gas such as CO$_2$ or CH$_4$ is normally added to the mixture in order to absorb most of the energetic UV photons and permit stable operation at higher gains. For this reason these gases are known as quenchers.

2.2.4. Count rate capability

Space charge effects due to the slow motion of the positive ions limit the count rate capability of the chamber. It has been shown (Hendricks 1969) that in a cylindrical chamber operating at a count rate of $R$ counts per unit length of wire there is an effective reduction in the field around the anode wire which is equivalent to a change in the applied voltage given by

$$
\delta V = -\frac{QRT_s}{4\pi\varepsilon_o}
$$

where $Q$ is the total charge in one pulse, $T_s$ is the total positive ion transit time and $\varepsilon_o$ is the permittivity of free space. Equations 2.4 and 2.5 indicate that the gain decreases exponentially if the applied rate and/or the pulse amplitude are increased. This effect can be compensated for by spreading the events over a greater length of wire, which is the approach followed when using planes of anode wires in a multiwire proportional chamber.
2.3. The multiwire proportional chamber

2.3.1. Chamber geometry

A typical MWPC consists of 3 parallel planes of wires. The mid plane is formed by a set of very thin (~20 μm diameter) parallel and equally spaced wires held at a positive potential, and is called the anode plane. Two cathode planes (which can be made of wires or foils) are placed at the same distance on either side of the anode plane. The wire structure is supported by a set of reinforced plastic frames, which also provide support for the electrical contacts and the printed circuit boards. This structure is in turn enclosed by a gas-tight container which has a suitable inlet window. Figure 2.2 shows a diagram of a cross-sectional view of the wire arrangement. The two most important parameters that define the chamber geometry are the anode wire spacing (s) and the distance between the anode and cathode planes (l).

![Figure 2.2. Schematic diagram of the geometry of a typical MWPC. l is the anode-cathode distance, and s is the distance between anode wires.](image)

The electrodes are arranged in such a way that the electric field is uniform over most of the volume of the chamber and only increases at very short distances from the anode wires, as indicated by the equipotentials and field lines shown in figure 2.3. This means that the electrons liberated by an ionising event in the gas will drift following the field lines until they approach the high field region near the anode wires, where the process of avalanche multiplication occurs.

It is possible to demonstrate that the analytic expression for the electric field near the anode wire reduces to that of a cylindrical proportional counter, that is
\[ E(r) = \frac{CV_0}{2\pi\varepsilon_0} \frac{1}{r}, \quad r << s \]  

(2.6)

where \( r = \sqrt{x^2 + y^2} \). Therefore, the same expressions that describe the operational characteristics of a cylindrical chamber can be used for the MWPC provided that the correct value for the capacitance per unit length, which is given by

\[ C = 2\pi\varepsilon_0 \left[ \left( \frac{\pi l}{s} \right) - \ln \left( \frac{\pi d_w}{s} \right) \right]^{-1} \]  

(2.7)

is used. In this equation \( d_w \) is the anode wire diameter. Computed values of the capacitance using equation 2.7 for typical geometrical parameters \( (d_w = 10-20 \ \mu m, \ l = 2-4 \ mm, \ s = 1-3 \ mm) \) indicate that, in general, the capacitance decreases rapidly with wire spacing and it does not depend very strongly on the wire diameter.

Figure 2.3. Schematic diagram of the electric field distribution near the anode wires in a MWPC (from Sauli 1977).

### 2.3.2. Position sensitivity

Imaging applications of the MWPC require the determination of the interaction point in two dimensions. Several readout methods designed to attain this position information were developed during the 1970s (Reading 1977). In a two coordinate MWPC the information is read out using the induced signals from the cathode planes. The coordinates of the event are normally determined using either the delay line readout method or by charge centroiding (Baru 1993).
The accuracy of localisation is determined by the anode wire spacing and by the range of the secondary products that follow the ionisation process. It is clear that a smaller anode wire spacing will produce better position resolution. However, spacings of less than 2 mm are very difficult to operate (Sauli 1977) because of the complex inter-relationship that exists between different operational parameters such as the applied voltage and capacitance, and because of electrostatic and mechanical limitations that depend, for example, on the anode wire diameter. Typical values for the anode wire distance in imaging chambers reported in the literature are of the order of 1 mm (Bateman 1987).

Secondary products affect the resolution properties of a multiwire chamber because they can deposit energy away from the primary interaction site. The range of the secondary products depends on the nature of the gas filling, the pressure at which the chamber is operated and the type and energy of the secondary products (electrons or fluorescent photons). The range of Auger electrons or photoelectrons can be determined (Sauli 1977) using the semiempirical formula

\[ r_e = 0.71 E^{1.72} \]  

(2.8)

where \( E \) is the electron energy in MeV and \( r_e \) is given in units of g/cm\(^2\). The low density of gases leads to large ranges at atmospheric pressure. For example a 30 keV electron has a range of \(-3\) mm in xenon \( (\rho_{xe} = 5.485 \times 10^{-3} \text{ g/cm}^3) \).

### 2.3.3. The choice of gas

As mentioned in section 2.2.3, the choice of gas filling depends on the intended application. Table 2.1 lists some of the most important properties of commonly used proportional counting gases. \( Z \) and \( A \) are the atomic number and mass, respectively, \( \rho \) is the density at atmospheric pressure and \( w_i \) is the ionization energy in eV per electron-ion pair. When high efficiency is required Xe with small proportion of a quencher gas \( (\text{CO}_2, \text{CH}_4 \text{ or isobutane}) \) is normally used.

<table>
<thead>
<tr>
<th>Gas</th>
<th>( Z )</th>
<th>( A )</th>
<th>( \rho [\text{mg/cm}^3] )</th>
<th>( w_i )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ar</td>
<td>18</td>
<td>39.9</td>
<td>1.66</td>
<td>26</td>
</tr>
<tr>
<td>Kr</td>
<td>36</td>
<td>83.8</td>
<td>3.49</td>
<td>24</td>
</tr>
<tr>
<td>Xe</td>
<td>54</td>
<td>131.3</td>
<td>5.49</td>
<td>22</td>
</tr>
<tr>
<td>( \text{CO}_2 )</td>
<td>22</td>
<td>44.0</td>
<td>1.86</td>
<td>33</td>
</tr>
</tbody>
</table>

Table 2.1. Physical characteristics of commonly used proportional counter gases.
Figure 2.4 shows the mass attenuation coefficients for Xe between 10 and 100 keV. It can be observed that the photoelectric effect totally dominates over all of the other interaction processes. With a Xe filling a chamber has an optimum operating x-ray energy just above the Xe K-shell absorption edge (34.6 keV). This energy is well suited for x-ray work, since it is approximately equal to the mean energy produced by a typical x-ray tube operating in the range 70-80 kVp.

2.4. Scanned-projection digital radiography systems

Scanned Projection Radiography (SPR) is a technique in which the patient is scanned with a fan beam of radiation and the transmitted x-ray photons are detected with a linear array of detectors. This technique received considerable attention in the early 1980s, primarily because of its potential for low-scatter imaging and suitability for digital radiography applications.

Perhaps the most familiar implementation of scanning fan beam geometry is the projection view mode in Computed Tomography (CT) scanners. CT projection views were used in the early days of digital radiography as an aid for patient localisation. It was soon realised that the good scatter rejection provided by the fan-beam geometry coupled to the possibility of digital manipulation of the image offered enhanced low-contrast detectability when compared to traditional film-screen radiography.
A number of dedicated SPR systems were developed in the early 1980s trying to exploit this idea. Drost and Fenster (1982) build a digital SPR prototype system based on a septaless, high-pressure xenon ionisation detector. The main aim was to improve spatial resolution and quantum efficiency in comparison with the detectors used in CT scanners. In their design they used the idea of focusing the collector electrodes on the x-ray tube focal spot (Drost and Fenster 1984, Drost et al 1984) in order to increase the depth of the working volume of gas and at the same time avoid parallax errors.

Mattson et al (1981) used a linear self-scanning photodiode array optically coupled to a phosphor screen as the x-ray detector in a SPR prototype. A system based on this design was used for clinical trials of digital radiography of the chest (Fraser et al 1983). The development of the SDRD started at around the same time. The first prototype was completed in 1984 (Baru et al 1985) and after several improvements in the design of the chamber a second prototype started clinical trials in 1986.

2.5. The Siberian Digital Radiographic Device

2.5.1. Description of the system

The SDRD consists of a standard tungsten target x-ray tube, a pair of slit collimators (one before and one after the patient), a mechanical scanning system, a linear MWPC, and the associated control and readout electronics. Figure 2.5 shows a layout diagram of the main components of the system. Two different versions of the device have been built that differ mainly in their spatial resolution capabilities. In this section a description of the standard resolution (ST) version of the device is given. The high resolution (HR) system will be treated in section 2.6.

A tantalum slit collimator between tube and patient (fore-slit) forms a 1 mm thick, 40 cm wide fan-shaped beam. A second collimator between patient and detector (aft-slit) suppresses scattered photons. The tube, collimators and detector are mounted on a rigid gantry. The system scans the patient at constant speed, producing a scan-line every 30 ms. Clean wire-hits are selected in a highly parallel electronic system and separately read out at a rate which is limited to a maximum of 600 kHz in each parallel channel by space-charge effects in the gas. Pixel counts stored in 16-bit scalers are read into CAMAC memory for each scan-line, taking approximately 8 seconds to build a two-dimensional image in RAM, available on-line for immediate
processing and analysis. All stages of the process, scan, readout and graphics display, are controlled from a PC which also stores images on hard disc. The image size is variable, depending on the length of the scan and the number of active wires, up to a maximum of 320×256 pixels. The pixel size at the plane of the patient is 1×1 mm², and is defined by the step size between scan-lines, the fore-slit aperture and the pitch of the anode wires.

![Figure 2.5. The Digital Radiographic Device. (1) x-ray tube, (2) slit collimator, (3) linear MWPC](image)

### 2.5.2. The MWPC with fan-anode plane

The active element of the system is a linear multiwire proportional chamber, which was specially designed for the SDRD (Baru et al. 1989a, 1989b). The anode plane is formed by 320 sense-wires, each of which points directly towards the focal spot of the x-ray tube (see figure 2.6). The distance from the focal spot to the centre of the chamber is 1350 mm. The fan-anode plane geometry solves the problem of keeping constant spatial resolution along the working depth of the chamber without affecting count rate capability and efficiency. The distance between the anode and cathode planes is approximately 2 mm. The cathode planes are inclined with respect to the anode plane to compensate for changes of gain in the gas along the anode wires.

X-ray photons are converted in the region between the upper cathode and the drift electrode and the electrons drift through the cathode before being multiplied near the anode wires. The working gas mixture is Xe+20% CO₂ at 3 atm pressure. The total
depth of the chamber is divided into a 2.5 cm inactive layer gas, followed by a 5 cm layer of active gas. The whole system is enclosed in a duraluminium box.

![Diagram of the linear MWPC with fan anode plane.](image)

Each anode wire is connected to an amplifier-discriminator unit that produces a logical output pulse if the energy deposited is bigger than ~2 keV. A selection circuit between every pair of neighbouring anode wires is used to reject coincidence events that could result from the absorption of x-ray photons in the region between wires.

### 2.5.2.7. Efficiency

The quantum efficiency of the chamber is given by the fraction of the incident photons that interact within the gas volume of a single detector element (Rutt et al 1983). For monoenergetic photons it is given by

\[
\eta_0(E) = \left( e^{-\mu_w(E)L_w} - e^{-\mu_i(E)L_i} \right) \left(1 - e^{-\mu_i(E)L_a} \right)
\]

(2.9)

where \(\mu_w\) and \(\mu_i\) are the attenuation coefficients of the inlet window material and the pressurised gas mixture, respectively, \(L_w\) is the thickness of the inlet window, \(L_i\) is the depth of the inactive layer of gas, and \(L_a\) is the working depth of the chamber. The first factor in equation 2.9 describes the fractional transmission of primary photons.
through the front window and the inactive part of the chamber, while the second factor indicates the fractional attenuation of primary photons in the working depth of gas.

![Graph](image)

**Figure 2.7.** Quantum efficiency as a function of incident x-ray energy for the chamber described in this section (Xe+20%CO₂ at 3 atm, \(L_\text{c}=0.5 \text{ mm}, L_\text{g}=2.5 \text{ cm}, L_\text{w}=5 \text{ cm}\). The dotted line represents the fractional absorption on the active depth of gas. The solid line in figure 2.7 shows the quantum efficiency as a function of x-ray energy for a chamber with a 0.5 mm thick inlet duraluminium window, 2.5 cm inactive layer of gas, and 5 cm working depth, as calculated with equation 2.9. This geometry corresponds to the chamber that has been evaluated in this thesis. The dotted line in the same figure represents the fractional attenuation that corresponds to the working depth of the chamber. It can be seen that attenuation in the front window causes a severe drop in quantum efficiency at low energies, while primary transmission accounts for a more gradual decrease at higher energies. The quantum efficiency of 32% at 60 keV obtained from this calculation is in good agreement with the measured value of 30% reported by Baru et al (1989b).

In general terms, the rejection of coincidence events means that the total efficiency of the detector depends not only on the quantum efficiency of the chamber, but also on the coincidence rate. The number of coincidence counts depends on several factors such as chamber voltage and x-ray energy. At higher x-ray energies the range of the photo-electrons increases, and the number of coincidences grows. An increase in the chamber voltage \((V_\text{c})\) causes an increase in gas amplification (equation 2.4), which
also translates into a higher coincidence rate. Finally, increasing the counting rate results in a reduction of gas gain (equations 2.4 and 2.5) and, therefore, a decrease in the fraction of coincidences with respect to the total number of counts.

2.5.2.2. Count rate capability

The counting rate capability of the chamber depends on the space-charge accumulated on the working gas volume at high rates, and on the dead time of the electronics. The reduction in count rate can be described by the reduction in the effective chamber voltage (Mathieson 1986), which is given by

\[
\delta V = \frac{Q R l^2}{2 \mu_i V_a C}
\]  

(2.10)

where \( \mu_i \) is the positive-ion drift mobility. Equation 2.10 indicates that the reduction in the chamber voltage \( V_a \) is proportional to the square of the anode-cathode gap \( l \). In the SDRD the rate dependence on space-charge effects is minimised by maintaining the anode-cathode distance as small as possible. Present chamber designs use an anode-cathode gap of 2 mm.

The errors in the electronics depend on the dead time of the amplifier-discriminator units and on the dead time of the selection circuits used to reject coincident events. The dead time in the discriminators leads to a reduction in the detection of single hit events with increasing counting rate, while the dead time in the selection circuits results in a fraction of coincident events being counted as single hits, thereby partially compensating the errors related to the discriminator units.

The chamber is calibrated in such a way that the reduction in the total efficiency of single-hit counts is of the order of 20% at a count rate of 600 kHz per channel when irradiated with a 70 kVp beam using an added filtration of 0.3 mm Cu.

2.5.3. Spatial resolution

The resolution properties of a scanning digital radiographic system depend on different factors for the longitudinal (along the slit) and the transverse (along the scan) directions. The theoretical resolution limit in the longitudinal direction for a septaless detector like the SDRD is given by the Nyquist limit associated with the distance between sampling points \( \Delta x \), that is \( f_{\text{Nyquist}} = 1/(2 \cdot \Delta x) \). The transverse resolution is
limited either by the transverse sampling frequency or by the cutoff frequency associated with the aft-slit aperture \((A_y)\), \(f_c = 1/A_y\). The transverse resolution limit is obtained with the smallest number of samples when the transverse sampling Nyquist frequency \(f_{Ny} = f_{Ny} = f_{Ny}\) is equal to the aft-slit aperture cutoff frequency (Tesic \textit{et al} 1983). If equal resolution is to be obtained in both directions (i.e. \(f_x = f_y\)) then the sampling distances should be equal and the slit width should be twice the inter-detector spacing.

2.5.3.1. \textit{Longitudinal resolution}

In the SDRD the working length of the anode wires is several centimetres in order to reduce space charge effects and increase detection efficiency. If the wires were parallel this would introduce parallax errors. For this reason the anode wires are stretched pointing towards the focal spot of the x-ray tube. On the other hand, the absorption of an x-ray quantum by a Xe atom leads to the emission of photo-electrons, Auger-electrons and fluorescent photons. The ranges of these particles influence the spatial resolution of the chamber since it can result in the operation of several channels simultaneously. Baru \textit{et al} (1989a) have shown using computer simulations that in the SDRD the probability of simultaneous counting is at least 10 times higher for neighbouring wires than for wires that are further apart. In the standard resolution version of the SDRD a set of selection circuits are used to exclude coincidence events. In this case the spatial resolution would be expected to be close to the geometrical limit imposed by the distance between anode wires.

2.5.3.2. \textit{Transverse resolution}

Although in principle better resolutions can be obtained by decreasing the inter-detector spacing and, correspondingly, the beam width, in practical terms there are limitations not only because of the chamber design, but also by restrictions imposed in terms of tube loading and finite focal spot size.

If the width of the beam is decreased then there is loss of primary radiation due to the unsharpness of the fore-slit at the plane of the aft-slit. This problem is illustrated in figure 2.8. From simple geometrical considerations it can be seen that the loss of primary radiation depends on the focal spot size \((F)\), beam width (fore-slit aperture, \(A_f\)) and fore-slit magnification. Tesic \textit{et al} (1983) have analysed the dependence of primary loss on these parameters and concluded that the slit geometry degrades the
primary transmission very rapidly for focal spot sizes greater than 0.8 mm, fore-slit magnifications greater than a factor 2 and beam widths narrower than 1 mm.

![Diagram of focal spot and apertures](image)

**Figure 2.8.** Geometrical projection of the focal spot (F) at the aft-slit plane for a fore-slit aperture $A_f$. $I$ represents the primary transmission and $P$ penumbra.

Similar considerations were taken into account in the design of the SDRD (Babichev et al 1989). In the ST version of the SDRD the aperture and the magnification of the fore-slit are $A_f = 1$ mm and $M = 2.21$, respectively. The aft-slit collimator, however, was chosen to be interchangeable between 1 mm or 3 mm in order to achieve either increased resolution or higher counting rates, respectively.

### 2.5.4. Contrast sensitivity and dynamic range

The contrast sensitivity of the system is determined by the non-uniformities of the MWPC, by the differences of registration thresholds between channels and by the statistical fluctuations on the number of detected x-ray quanta. The contributions due to defects on the chamber and electronics are stable over time and can be corrected by measuring the distribution of pixel counts on an image taken while the chamber is irradiated with a uniform x-ray flux.

The statistical fluctuations depend only on the number of registered x-ray quanta. The maximum number of counts per pixel is of the order of 18,000 working at 600 KHz per channel (i.e. an exposure time of 30 ms per scan-line), and this corresponds to a mean square deviation of 0.7%. Taking into account the influence of defects on the chamber and in the registration electronics the contrast sensitivity of the system has been estimated to be of the order of 1% (Babichev et al 1991b).
The fact that the chamber has linear response, practically zero intrinsic background and high counting rate capability means that it is possible to have a very wide dynamic range. The upper limit is determined by the maximum number of counts per pixel \((-2 \times 10^4)\), and the lower limit is determined only by the statistical fluctuations on the number of counts.

2.6. The high resolution system

If an x-ray quantum is absorbed near the centre between two neighbouring anode wires, the primary ionisation in the majority of cases is divided into two parts, and two channels of the detector count simultaneously. In the standard resolution version of the SDRD such events are rejected by a special selection circuit.

In the high resolution (HR) system (Babichev et al 1992) an extra set of scalers is introduced between every pair of anode wires. Coincidences (within 75 ns) between neighbouring wires are recorded only in these extra scalers at a rate which is comparable to the single-hit rate. The information for each line is stored in 640 16-bit scalers, corresponding to a horizontal pixel size at the patient of approximately 0.5 mm. The vertical pixel size is matched to this by reducing the time for each line-scan to 15 ms and adjusting the aperture of the collimators to 0.5 mm fore-slit and 1 mm aft-slit.

2.6.1. Counting efficiency

The main problem related to the use of coincidence counts to improve the spatial resolution of the chamber is that the counting efficiency of single-hit and coincidence channels is very different. The primary ionization in coincidence events is normally divided in an arbitrary way between two neighbouring wires. This process leads to a different pulse height distribution for single and double hits and therefore different dependencies of the efficiency on the energy of the incoming quanta and on the counting rate at which the chamber is operated (Shekhtman 1993).

Figure 2.9 shows a plot of the normalised efficiency of the chamber for anti-coincidence and coincidence channels, measured at 70 kVp, with 0.3 mm Cu added filtration (Ponomarev and Shekhtman 1993). It can be observed that the counting efficiency of single hits is higher than that of double hits at all counting rates. For
example, at a count rate of 100 kHz per channel the efficiency of counting for single hits is approximately 20% higher than for double hits.

It is interesting to note that the chamber efficiency for anti-coincidence channels first increases with counting rate, reaches a maximum at about 200 kHz and then decreases down to ~80% at 600 kHz per channel. The efficiency of coincidence channels, on the other hand, decreases monotonically with counting rate. The decrease in efficiency of double hits is very steep, from about 80% at 100 kHz down to only 40% at 600 kHz. This non-uniformity of response due to the two interleaved types of readout channel is compensated for in the final image using a normalising algorithm.

![Graph showing normalized efficiency vs. counting rate](image)

**Figure 2.9.** Normalised chamber efficiency as a function of counting rate for anti-coincidence and coincidence channels.

### 2.6.2. Detector aperture response

The shape of the aperture response function is dependent upon the channel counting mode. Figures 2.10a and 2.10b show the measured aperture response functions for anti-coincidence and coincidence channels, respectively. The curves were obtained by irradiating the chamber with a narrow (100 μm) beam of radiation, and translating the slit across the detector array while measuring the signal from a single detector element.
The shape of the aperture response function of the anti-coincidence channel is similar to the square aperture response that would be expected from a perfect detector, whereas the aperture of the coincidence channel is very narrow, with long extending tails that overlap considerably with adjacent channels. The corresponding full width at half maximum (FWHM) and full-width at tenth maximum (FWTM) are 0.88 and 1.32 mm for the anti-coincidence channel, and 0.52 and 1.81 mm for the coincidence channel.

![Figure 2.10](image)

**Figure 2.10.** Measured detector aperture response in the high-resolution chamber. (a) Anti-coincidence channel. (b) Coincidence channel.

The effect that the difference on counting efficiency and detector aperture response between anti-coincidence and coincidence channels have on the imaging characteristics of the HR chamber will be discussed in Chapter 3.
This chapter presents the results of an evaluation of the imaging performance of the SDRD. Sections 3.1 and 3.2 describe the contrast-detail performance of the device as compared to a modern film-screen combination used in routine clinical practice in the UK. The effect that different geometrical parameters and modes of operation of the multiwire chamber have on the detectability of simple objects is also studied in section 3.2. The second part of this chapter is devoted to a comparison of the spatial transfer and noise characteristics of the two different modalities in which the SDRD can be operated (standard and high resolution). Finally, some typical clinical images produced by the SDRD are shown and the associated patient doses for these images are compared with those currently achieved in England for the same examinations.

3.1. Contrast-detail evaluation

3.1.1. The test object

The Leeds test object TO.10 (Hay et al 1985) was used as the basis of the contrast-detail performance evaluation. TO.10 is a well known test tool which was originally designed for the assessment of the overall image quality of fluoroscopic x-ray systems by means of the threshold-contrast detail-detectability (TCDD) test. The phantom (see figure 3.1) is made of a 1 cm thick perspex plate that contains twelve sets of thin metal discs covering the range from 11.1 mm down to 0.25 mm in diameter. The sets are labelled with capital letters, starting with A for the largest details and up to M for the smallest ones. Each set contains nine different contrasts which are numbered from 1 to 9, starting with the highest contrast object. The test details produce calibrated input contrasts to the imaging system under investigation if the phantom is irradiated with an x-ray beam of known quality. The calibrated contrast values of TO.10 specified by the manufacturer of the phantom correspond to exposure conditions of 70 kVp with a 1.0 mm Cu filter added to override any inherent tube filtration.
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The TCDD test consists of exposing the TO.10 phantom using a clinically realistic dose. The images are evaluated under subdued background lighting in order to simulate the viewing conditions used by radiologists in a clinical situation. The recommended viewing distance is approximately four times the displayed diameter of the phantom. The threshold contrast for each diameter is determined by counting the number of objects of that diameter that can be seen in the image and using the tables of calibrated contrasts provided by the manufacturer of the phantom.

![Diagram of the Leeds test object TO.10](image)

**Figure 3.1.** Layout diagram of the Leeds test object TO.10. Detail sizes range from 11.1 mm diameter (row A) down to 0.25 mm (row M).

During this investigation several beam qualities were used in order to test the efficiency of the MWPC under different exposure conditions. For this reason the object contrast values produced by TO.10 under each new set of exposure conditions had to be re-evaluated. A simple mathematical phantom with x-ray attenuation characteristics equivalent to those of TO.10 was used to calculate tables of calibrated contrasts for an arbitrary set of exposure factors (filtration and kVp). The object contrasts were calculated in terms of photon fluence using a pencil beam model and taking into account the difference in the energy distribution of the x-ray beam passing through the detail and the background.

3.1.2. Exposure techniques and dosimetry

Normal exposure conditions on the SDRD include the use of 0.1-0.3 mm added Cu filtration (depending on the type of examination) to optimise the x-ray spectrum for
the energy response of the detector and at the same time reduce the dose to the patient. In this study two different irradiation conditions were used with the TO.10 phantom in order to simulate different clinical imaging situations. A 0.1 mm Cu added filtration plus 6 cm Perspex was used to represent chest x-ray conditions and a 1.0 mm Cu filter was used to represent abdominal imaging. In what follows these two filtrations will be referred to as THX and ABDO, respectively.

Entrance air kerma values for each image of the phantom were measured using either an air ionisation chamber (Keithley Instruments, model 35050A) or an air-equivalent plastic scintillator probe. The choice of a particular dosimeter was governed by the sensitivity and availability of each system. The ion chamber was used for the film-screen exposures made in the UK, and the scintillator and/or the ion chamber were used for the SDRD exposures in Russia. Both dosimetry systems were cross calibrated.

![Figure 3.2](image)

**Figure 3.2.** Schematic diagram of the exposure conditions for the TO.10 images corresponding to THX filtration. For ABDO filtration a single 1 mm Cu filter was used instead of the 0.1 mm Cu filter and the Perspex block.

The position of dosimeters was dependent upon the exposure conditions used, but all values of air kerma have been corrected to represent entrance absorbed doses and hence permit meaningful comparisons between phantom images. The position of the dosimeter was important in this study since the test object (TO.10) was used in two composite phantoms. For the THX exposure conditions the phantom consisted of 6 cm of Perspex + TO.10, and for the ABDO conditions the phantom consisted of 0.9 mm of Cu + TO.10. In the case of THX conditions the dosimeters were placed on the entrance surface of the phantom, while in the case of ABDO conditions the dosimeters were placed on the surface of TO.10, that is, air kerma was measured after 1 mm Cu filtration. The results were corrected to take this difference into account. All values of
entrance air kerma given in this section correspond to a distance between the focal spot of the x-ray tube and the dosimeter of 112 cm.

3.1.3. Data collection

3.1.3.1. SDRD images

**ST-SDRD.** The optimum count rate capability of the ST-SDRD corresponds to $10^4$ counts per pixel. Starting at this point a series of images was taken at progressively lower exposures, decreasing tube current down to levels of the order of 500 counts per pixel. Table 3.1 shows the exposure conditions for the ST-SDRD under two different filtrations at 70 kVp, using a 3 mm aft-slit collimator. The mean number of counts ($N_b$) measured over a small area of background in the phantom images is also shown in the same table. Table 3.2 shows the exposure conditions for images acquired with the ST-SDRD with a 1 mm aft-slit collimator under THX filtration.

<table>
<thead>
<tr>
<th>Table 3.1. Exposure conditions for TO.10 on the ST-SDRD, 3 mm aft-slit collimator, 70kVp.</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Image</strong></td>
</tr>
<tr>
<td>--------------</td>
</tr>
<tr>
<td>ST01</td>
</tr>
<tr>
<td>ST02</td>
</tr>
<tr>
<td>ST03</td>
</tr>
<tr>
<td>ST04</td>
</tr>
<tr>
<td>ST05</td>
</tr>
<tr>
<td>ST06</td>
</tr>
<tr>
<td>ST07</td>
</tr>
<tr>
<td>ST08</td>
</tr>
<tr>
<td>ST09</td>
</tr>
<tr>
<td>ST10</td>
</tr>
<tr>
<td>ST11</td>
</tr>
</tbody>
</table>

**HR-SDRD.** The nominal pixel size of the HR system is one-half of the ST system pixel size. If the same exposure factors are used in both systems then the mean number of counts per pixel would be about 4 times smaller in the HR system. Therefore, the tube voltage was increased to 80 kVp for the exposures made with the HR-SDRD in order to maintain high pixel statistics without incurring in tube loading problems. Table 3.3 gives the exposure conditions for the HR images; in this case the mean number of counts at the maximum tube current corresponds to levels of ~5000 and 2500 counts per pixel for THX and ABDO filtrations, respectively.

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Table 3.2. Exposure conditions for TO.10 on the ST-SDRD, 1 mm aft-slit collimator, 70 kVp.

<table>
<thead>
<tr>
<th>Image</th>
<th>Filtration</th>
<th>mA</th>
<th>Air kerma [μGy]</th>
<th>(N_b)</th>
</tr>
</thead>
<tbody>
<tr>
<td>ST12</td>
<td>THX</td>
<td>40</td>
<td>74 ± 7</td>
<td>9203</td>
</tr>
<tr>
<td>ST13</td>
<td>THX</td>
<td>25</td>
<td>46 ± 5</td>
<td>6258</td>
</tr>
<tr>
<td>ST14</td>
<td>THX</td>
<td>10</td>
<td>21 ± 2</td>
<td>2823</td>
</tr>
<tr>
<td>ST15</td>
<td>THX</td>
<td>4</td>
<td>8 ± 1</td>
<td>1328</td>
</tr>
<tr>
<td>ST16</td>
<td>THX</td>
<td>1.6</td>
<td>3 ± 0.5</td>
<td>425</td>
</tr>
</tbody>
</table>

Table 3.3. Exposure conditions for TO.10 on the HR-SDRD, 1 mm aft-slit collimator, 80 kVp.

<table>
<thead>
<tr>
<th>Image</th>
<th>Filtration</th>
<th>mA</th>
<th>Air kerma [μGy]</th>
<th>(N_b)</th>
</tr>
</thead>
<tbody>
<tr>
<td>HR1</td>
<td>THX</td>
<td>40</td>
<td>39 ± 4</td>
<td>5350</td>
</tr>
<tr>
<td>HR2</td>
<td>THX</td>
<td>12</td>
<td>15 ± 2</td>
<td>1850</td>
</tr>
<tr>
<td>HR3</td>
<td>THX</td>
<td>3</td>
<td>5 ± 1</td>
<td>600</td>
</tr>
<tr>
<td>HR4</td>
<td>ABDO</td>
<td>50</td>
<td>45 ± 5</td>
<td>2450</td>
</tr>
<tr>
<td>HR5</td>
<td>ABDO</td>
<td>12</td>
<td>15 ± 2</td>
<td>1200</td>
</tr>
<tr>
<td>HR6</td>
<td>ABDO</td>
<td>6</td>
<td>9 ± 1</td>
<td>700</td>
</tr>
</tbody>
</table>

3.1.3.2. Film-screen images

Radiographic film-screen images were taken at The Middlesex Hospital, London, using a general purpose x-ray unit and a medium speed film-screen combination (Kodak Ortho-G + Curix Ortho-Regular CX). Exposure conditions are given in table 3.4 and correspond to those used with the ST-SDRD. Starting from a visually acceptable image (background optical density within the phantom of approximately 1.0), the exposure was progressively lowered until most image details were lost.

Table 3.4. Exposure conditions for TO.10 images on film-screen at 70 kVp.

<table>
<thead>
<tr>
<th>Image</th>
<th>Filtration</th>
<th>mAs</th>
<th>Air kerma [μGy]</th>
</tr>
</thead>
<tbody>
<tr>
<td>fs1</td>
<td>THX</td>
<td>3.2</td>
<td>51 ± 5</td>
</tr>
<tr>
<td>fs2</td>
<td>THX</td>
<td>2.5</td>
<td>41 ± 4</td>
</tr>
<tr>
<td>fs3</td>
<td>THX</td>
<td>2.0</td>
<td>23 ± 2</td>
</tr>
<tr>
<td>fs4</td>
<td>THX</td>
<td>1.6</td>
<td>20 ± 2</td>
</tr>
<tr>
<td>fs5</td>
<td>ABDO</td>
<td>5.0</td>
<td>109 ± 11</td>
</tr>
<tr>
<td>fs6</td>
<td>ABDO</td>
<td>3.2</td>
<td>66 ± 7</td>
</tr>
<tr>
<td>fs7</td>
<td>ABDO</td>
<td>2.5</td>
<td>50 ± 5</td>
</tr>
<tr>
<td>fs8</td>
<td>ABDO</td>
<td>1.6</td>
<td>32 ± 3</td>
</tr>
</tbody>
</table>
3.1.4. Evaluation of the images

All images taken with the SDRD were displayed for its evaluation on a Sun SPARC station IPX. The display software allowed the operator to window the displayed pixel counts at any point in the full range, and select the grey levels assigned to a given subrange of pixel values. Film-screen images were viewed on standard light boxes.

A group of 10 observers viewed the whole set of images under subdued background lighting conditions. Two of these observers were familiar with the use of T0.10 for the evaluation of fluoroscopy units, and the other eight were 'new' observers. Each observer was free to decide the displayed count levels for the SDRD images, and it took between 20 and 30 minutes, divided in two sessions, to complete the evaluation of all images. The number of objects of a given size that could be detected was recorded. These values have been averaged for all the observers and used along with the dose required for each exposure to produce contrast-detail and contrast-dose diagrams for each set of exposure conditions.

3.2. Results and discussion of the TCDD analysis

An established method of presenting the results from a contrast-detail test is in the form of contrast-detail diagrams, where threshold contrast is plotted as a function of detail diameter. In order to demonstrate some of the performance features of the SDRD threshold contrast was also plotted as a function of the entrance air kerma for objects of a given size (contrast-dose diagrams). The advantages that the SDRD offers over film-screen systems can be fully appreciated when the data is presented in this way.

The results are presented in figures 3.3 to 3.5. Only three sets of data are shown for each system and for each filtration. The three sets cover the complete range of exposure conditions used. Points have only been plotted when all observers claimed to see at least one object in a given row in order to ensure consistent averages for the observed thresholds. Residual point to point fluctuations are due to the finite number of details in each set, the finite number of observers, and the different detectability thresholds set by individuals.
3.2.1. SDRD data

Figures 3.3a and 3.3b show the contrast-detail diagrams for the ST-SDRD at different levels of exposure. As expected, for a given size lower contrast objects can be detected at higher doses. In figure 3.3a it can be observed that increasing the air kerma by a factor of 28 leads to a reduction of threshold contrast by a factor of 5. This approximate relationship holds true over the range of object sizes evaluated and indicates that the limit of detection is set by quantum noise. Significant degradation in contrast sensitivity is only noticed at extremely low dose levels where photon counts per pixel have dropped below 500, which corresponds to an entrance air kerma of the order of 1 μGy. At the highest exposure level the system was able to detect objects down to 0.5 mm diameter, which is half the nominal pixel dimension.

![Contrast-detail diagram](image)

**Figure 3.3.** Contrast-detail diagram for the ST-SDRD: (a) THX filtration; (b) ABDO filtration. The dotted lines correspond to the behaviour predicted by the model of threshold detection proposed by Harrison and Kotre (1986).

The threshold detection model proposed by Harrison and Kotre (1986), which has already been described in Chapter 1, can be used to explain the shape of the contrast-detail diagrams. The dotted lines in figures 3.3a and 3.3b represent the behaviour predicted by equation 1.4. It can be seen from the fit of the dotted lines that there is very good agreement between the SDRD data and the model.

The fits were obtained using the mean output signal \( \langle N_p \rangle \) and the standard deviation \( \langle \sigma_p \rangle \) of the pixel values measured over an area of background in the digital images. In
both images (ST01 and ST07) the pixel counts were observed to follow Poisson statistics (a detailed analysis of image noise is given in §3.4), with \( N_b = 10^4 \) and \( \sigma_e \approx 10^2 \). It was assumed that the value of the total unsharpness was approximately equivalent to the pixel size, that is \( a_u = 1 \text{ mm} \). The best fit was obtained using a threshold signal-to-noise ratio \( (k_{th}) \) of 4.4. This value of \( k_{th} \) is only slightly smaller than the value of 5 originally proposed by Rose (1948).

It should be pointed out that the same curve fits the data from images ST01 and ST07. Both images have the same pixel statistics and, therefore, the same signal-to-noise ratio characteristics. The only difference between these two images is that the entrance dose required to obtain the same level of pixel counts was approximately two times higher for ABDO filtration.

The results of the contrast-detail evaluation of the HR system are shown in figure 3.4. The same model of threshold detectability was used to fit the HR data. Again, it was sufficient to consider the size of the effective sampling aperture to be of the order of the pixel size, that is \( a_u = 0.5 \text{ mm} \). However, in this case the magnitude of the noise measured from the digital images was found to be higher than what would be expected from Poisson statistics (e.g. \( N_b = 4453 \) and \( \sigma_e = 119 \) for image HR1). As a result the threshold signal-to-noise ratio required to fit the experimental data was higher. The best fit for image HR1 was obtained with \( k_{th} = 6.6 \), and the fit for image
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HR4 was obtained with \( k_H = 6.0 \). Since image HR1 has approximately two times higher pixel statistics than image HR4 these results suggest the presence of signal-dependent noise.

3.2.2. Film-screen data

Figures 3.5a and 3.5b show the contrast-detail diagrams for the images taken with the film-screen combination. As it would be expected, the film-screen combination has better intrinsic spatial resolution than the SDRD. Objects down to 0.25 mm diameter were clearly seen over the whole range of exposures used for both filtrations.

The general form of the film-screen contrast-detail curves is similar to those for the SDRD. However, unlike the SDRD the shape of the curves for the f-s system seems to be dependent upon the exposure conditions used. It can be observed that the detectability of low contrast objects degrades slightly more rapidly under ABDO exposure conditions. In the Rose-De Vries region, that is, in the region where the slope of the contrast-detail curve is approximately \(-1\) and the detail diameter is of the order of the total dimension of the unsharpness, the threshold detection model can be used to describe the experimental data. The dotted lines in figures 3.5a and 3.5b were obtained as a least squares fit of equation 1.4 for details of less than 1 mm diameter, with \( a_s = 0.39 \) for image fs2, and \( a_s = 0.45 \) for image fs7. In this case the model can only describe the contrast-detail properties of the system for small objects. For larger
Chapter 3 Evaluation of the MWPC system

details (> 4 mm diameter) \( C_{th} \) tends asymptotically to a minimum value, and the data can no longer be explained in terms of the resolution and noise properties of the images.

3.2.3. Comparison of TCDD performance

3.2.3.1. Effect of collimation

The effect that changes to the aperture of the aft-slit collimator have on the threshold detectability properties of the ST system are shown in figure 3.6. For images taken at the same level of exposure (e.g. ST02 and ST14, 21 µGy entrance air kerma) the use of a 3 mm aft-slit collimator produces consistently lower thresholds for objects down to ~1 mm diameter. For objects smaller than 1 mm the detectability is very similar for both collimators. The gain in resolution obtained when using a 1 mm collimator is lost in terms of detectability because of lower pixel statistics and, therefore, decreased signal-to-noise ratio.

![Figure 3.6](image)

Figure 3.6. Effect of changes in the aft-slit collimator on the contrast-detail performance of the ST-SDRD.

Equivalent performance can be obtained for large diameter objects with either collimator if the same pixel statistics are maintained. At a level of counts of \( 10^4 \) (images ST01 and ST12) the threshold contrast is approximately the same for objects down to 1 mm diameter. For smaller details the 1 mm aft-slit collimator has better detectability properties (see figure 3.6b); objects down to 0.35 mm diameter, which
are less than half the nominal pixel dimension, could be seen in image ST12. This, however could only be achieved at the cost of increasing the entrance dose to the phantom by a factor of 2.6.

3.2.3.2. Contrast-detail performance

A direct comparison of the contrast-detail performance of the SDRD (ST and HR) and the film-screen combination is shown in figure 3.7. All three images were taken at the same level of exposure (20 μGy entrance air kerma) using the same beam quality (70 kVp, THX filtration). It can be seen that both versions of the SDRD have better contrast sensitivity than the f-s system for medium and large, low-contrast objects. On the other hand, film-screen offers better detectability of small, high-contrast objects due to its higher intrinsic resolution.

![Figure 3.7](image)

**Figure 3.7.** Comparison of the contrast-detail curves for the SDRD (ST and HR) and the film-screen combination at the same level of exposure (20 μGy entrance air kerma), using THX filtration.

The HR system performs better than the ST system for all but the largest details. Threshold contrast sensitivity is approximately two times lower for objects below 2 mm diameter, and is only marginally higher for the largest object. At this level of exposure the HR system was able to detect details down to 0.35 mm diameter.
3.2.3.3. Contrast-dose performance

Direct comparison of the ST-SDRD and film-screen performance is best seen in figure 3.8, where the threshold contrast is plotted as a function of dose for rows A, D and H of TO.10 (11.1, 4.0 and 1.0 mm diameter, respectively). The straight lines superimposed on the SDRD data in these figures have slope equal to -1/2, as predicted by the contrast detectability models discussed in Chapter 1. Error bars have been included on one set of data to indicate the magnitude of the fluctuations. It can be seen that the behaviour of the SDRD agrees with the model over a very wide range of exposures. The behaviour of the film-screen data is very different. As the exposure level is reduced there comes a point below which the detected threshold contrast rises quickly. These points are 20 and 60 μGy in figures 3.8a and 3.8b, respectively. Unlike the SDRD, the performance of the film-screen combination degrades very rapidly at lower exposure levels.

![Figure 3.8](image)

Figure 3.8. Threshold contrast as a function of the entrance air kerma for the ST-SDRD and the film-screen system. (a) THX filtration. (b) ABDO filtration.

In the region of exposures where the two systems overlap, the SDRD has better contrast sensitivity for large and medium sized objects compared to film-screen. For the smallest objects the behaviour is reversed, with film-screen giving better images for all but the lowest doses. This is not unexpected, since the inherent resolution of the ST-SDRD is approximately 1 mm and film-screen systems have much better resolution capability.
It is clear from figures 3.8a and 3.8b that, provided there is sufficient inherent contrast in an object, the SDRD has the potential to detect it at very low exposures. It can also be seen that the SDRD has a very wide latitude over which information can be faithfully recorded. However, to be able to resolve very small objects much improved spatial resolution would be required, which can be done only at the expense of dose if the same level of photon counts per pixel is to be maintained.

3.3. Modulation transfer function analysis

3.3.1. Materials and methods

Transfer function analysis was used to study the spatial transfer characteristics of the SDRD. The MTFs of both the ST and the HR systems were determined using the edge spread function method. This method was chosen because it is easy to implement in practice and because it is less sensitive to noise effects at low spatial frequencies when compared to line spread function methods, as discussed in Chapter 1. The effects of collimation and counting rate on the shape of the MTF were also investigated.

Images of a Tantalum edge were taken using a 70 kVp beam and 0.1 mm Cu + 3 cm perspex filtration. Oversampled scans of the edge were obtained by placing it at small angles (<3°) with respect to the pixel sampling grid. Measurements were made in both the horizontal and vertical directions. The edge response data for the ST system were obtained using two different aft-slit collimators (3 mm and 1 mm) and four different tube currents, from 16 mA down to 1 mA, which corresponded to counting rates from $10^4$ down to $10^3$ counts per pixel for the 3 mm aft-slit, and from 5000 down to 300 counts per pixel for the 1 mm aft-slit. The data for the HR system were acquired using a 1 mm aft-slit collimator and a count rate equivalent to 2500 counts per pixel.

Figure 3.9a shows the horizontal edge spread functions obtained with the 3 mm aft-slit collimator. It can be seen that there is sudden increase in the number of counts in the region next to the edge which is followed by a slight decrease in the number of counts further away from the edge. The same edge effect was observed in the 1 mm data, although it was slightly less pronounced due to the lower pixel statistics.
At high counting rates the edge effect is very pronounced. In the image with the highest pixel statistics the number of counts near the edge increases by a factor ~1.3 over the mean number of counts in the background. At lower count rates the edge effect tends to disappear slowly. However, the spatial extent of the edge enhancement remains of the order of 1 mm at all count rates, which corresponds to the distance between anode wires, that is, it is only about one pixel wide.

![Figure 3.9](image.png)

**Figure 3.9.** Edge spread functions of the ST-SDRD at different counting rates, 3 mm aft-slit collimator. (a) Horizontal ESF. (b) Vertical ESF.

The vertical ESF data does not present the same effect. Figure 3.9b shows that the edge response in the vertical direction is a monotonically decreasing function. Rate dependence would not expected in the vertical direction since the effective detector aperture response should depend only on geometrical factors such as the slit size, sampling distance, focal spot size and fore-slit magnification, as discussed in §2.5.3.2.

### 3.3.2. Vertical MTF

#### 3.3.2.1. ST system

Figure 3.10 shows the vertical MTFs that correspond to the 1 mm aft-slit collimator ESF data. The MTFs were calculated by performing a 3-point forward-difference numerical differentiation of the edge spread function to obtain the LSF, followed by a discrete Fourier transform (DFT) of the LSF data (Press *et al* 1988). It can be observed that the shape of the MTF is independent of the counting rate of the
chamber, which confirms that the spatial resolution in the scan direction is determined by geometrical factors only.

The MTF that would correspond to a perfect 1 mm square aperture, which is given by $sinc(v)$, is also shown with a dotted line in the same figure. It can be seen that the experimental data has lower MTF as compared to the sinc function. This is not unexpected, since a certain amount of blurring due to the finite size of the focal spot was anticipated. A very good fit to the experimental data was obtained using an exponential function of the form

$$MTF_v(v) = \exp\left[-2(\pi\sigma_v v)^2\right]$$

with $\sigma_v = 0.46$, and is shown with a dashed line in the same graph. A single curve fits all the measured points very well, despite the fact that there is a difference of one order of magnitude between the highest and the lowest count rates.

![Figure 3.10. Vertical MTF of the ST-SDRD using 1 mm aft-slit collimator. The MTF of a 1 mm square aperture (dotted line) and a Gaussian fit to the experimental MTF data (dashed line) are also shown.](image)

The sampling distance between line scan acquisitions for the ST-SDRD is 1 mm in all cases. Therefore, the vertical Nyquist and cutoff frequencies are $f_{Ny} = 0.5$ mm$^{-1}$ and $f_c = 1$ mm$^{-1}$, respectively (see §2.5.3). From the fitted function it was calculated that the reduction of the measured MTF with respect to the sinc function at the Nyquist
frequency was approximately a factor of 0.44. The measured MTF takes a value of \( \approx 0.35 \) at \( v = f_{by} \), and is considerably non-zero well beyond the Nyquist limit, which means that aliasing effects are present in these images. The MTF is practically zero at 1 mm\(^3\), which coincides with the cutoff frequency of the 1 mm square aperture.

An exponential MTF of the form given by equation 3.1 has a corresponding LSF given by the Gaussian

\[
LSF(y) = \frac{1}{\sqrt{2\pi \sigma_y^2}} \exp\left(-\frac{y^2}{2\sigma_y^2}\right)
\] (3.2)

that is, equations 3.1 and 3.2 are a Fourier transform pair (Bracewell 1986). Figure 3.11 shows the LSF calculated with equation 3.2 using the value of \( \sigma_y \) obtained from the fitted MTF, along with a 1 mm square aperture response function. In this plot the blurring effect of the focal spot can be seen more clearly.

![Figure 3.11. Vertical LSF for a 1 mm collimator calculated from the fit to the MTF data. A 1 mm square aperture is also shown to indicate the extent of the blurring in this direction.](image)

Metz et al (1972) have demonstrated that aliasing errors on an exponential MTF such as the one given by equation 3.1 do not exceed 0.5% at all frequencies below the Nyquist limit if the sampling distance for the Gaussian LSF (equation 3.2) is less than 1/8 of its full width at half maximum (FWHM). In this particular case the edge was
sampled with an effective sampling distance of $\Delta y' = 0.02$ mm. The FWHM of the Gaussian LSF is given by

$$\text{FWHM}_y = 2.35 \cdot \sigma_y$$

Using the value of $\sigma_y$ obtained from the exponential fit the FWHM of the LSF given by equation 3.3 is 1.08 mm, which means that $\Delta y'$ was approximately 7 times smaller than FWHM$_y$. Therefore, the oversampling method used to determine the ESF ensures that practically no aliasing effects are present in the measured MTF.

![Figure 3.12](image.png)

**Figure 3.12.** Vertical MTF of the ST-SDRD, 3 mm collimator. The dotted line and the dashed line are the sinc functions for 3 mm and 1 mm square apertures, respectively. The dot-dashed line is a Gaussian fit of the measured MTF.

The 3 mm collimator data exhibits a similar behaviour to the 1 mm data, as it can be seen in figure 3.12. The form of the MTF does not depend on the count rate, either, and it can also be fitted with a Gaussian function of the form given by equation 3.1, this time with $\sigma_y = 0.78$. The Nyquist limit is the same as in the previous case (i.e. 0.5 mm$^{-1}$), since the scan-line sampling distance is also 1 mm. However, the first zero of the MTF occurs at $\sim 0.65$ mm$^{-1}$, which is approximately halfway between the cutoff frequency of sinc(3v) (0.33 mm$^{-1}$) and the cutoff frequency of sinc(v) (1 mm$^{-1}$). The resolution is obviously worse due to the increased width of the aft-slit collimator.
3.3.2.2. *HR system*

The sampling distance of the HR-SDRD is 0.5 mm, which is equivalent to a Nyquist and cutoff frequencies of 1 mm\(^{-1}\) and 2 mm\(^{-1}\), respectively. The vertical MTF of the high resolution system has the same shape as the ones measured with the ST system. Figure 3.13 compares the results obtained with both systems, together with the sinc function corresponding to a 0.5 mm square aperture. A significant improvement in the resolution properties of the system can be observed. However, it can also be seen that the finite size of the focal spot once again reduces the spatial frequency response of the detector as compared to the square aperture response. The Gaussian fit to the MTF data was obtained using a value of \(\sigma_y = 0.28\).

![Figure 3.13. Comparison of the vertical MTFs of the HR and the ST-SDRD.](image)

Table 3.5 summarises some of the geometrical parameters that determine the vertical MTFs for both modalities of the SDRD, and the parameters used to fit the measured data. The results show that the FWHM of the Gaussian LSF depends on the aperture of the collimators and on the sampling distance (\(\Delta y\)). As it would be expected, a reduction in the aft-slit aperture and/or the sampling distance improves the spatial transfer characteristics of the system. The FWHM for the ST-SDRD with a 3 mm aft-slit is almost twice the nominal pixel size, whereas in the other two cases the FWHM is approximately equivalent to the pixel dimension.
Table 3.5. Geometrical parameters associated with the vertical MTF of the ST and HR-SDRD. The FWHM corresponds to the Gaussian LSF.

<table>
<thead>
<tr>
<th>System</th>
<th>fore-slit</th>
<th>aft-slit</th>
<th>Δy</th>
<th>$f_{Ay} [\text{mm}^{-1}]$</th>
<th>$\sigma_y$</th>
<th>FWHM$_y$</th>
</tr>
</thead>
<tbody>
<tr>
<td>ST</td>
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<td>3</td>
<td>1</td>
<td>0.5</td>
<td>0.78</td>
<td>1.84</td>
</tr>
<tr>
<td>ST</td>
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<td>1</td>
<td>1</td>
<td>0.5</td>
<td>0.46</td>
<td>1.08</td>
</tr>
<tr>
<td>HR</td>
<td>0.5</td>
<td>1</td>
<td>0.5</td>
<td>1</td>
<td>0.28</td>
<td>0.65</td>
</tr>
</tbody>
</table>

3.3.3. Horizontal MTF

3.3.2.2. ST system

The shape of the horizontal MTF is very different to the vertical MTF, as it would be expected from a comparison of the corresponding edge spread functions. The MTF in the horizontal direction is determined not only by the spacing between anode wires, but also by the way the detector registers each event. In principle, the absence of septa in the multiwire chamber could result in a loss of resolution due to crosstalk between neighbouring wires arising from the secondary particles that deposit energy away from the primary interaction site. In the ST-SDRD this problem is avoided by counting single-wire hits only, as explained in Chapter 2. This rejection method causes a dependence of the counting efficiency of the chamber on both the counting rate and on the rate of coincidence rejection. It is therefore not surprising that the edge response functions show an effect related to the counting rate efficiency in the region next to the edge, where neighbouring wires are operating at very different count rates, and that this effect also depends on the exposure level, since the rate of coincidence rejection is a function of the incoming flux.

Figure 3.14 shows the horizontal MTF of the ST-SDRD using a 3 mm collimator, as calculated from the ESF data shown previously in figure 3.9. The function sinc($\alpha$) is also shown as a reference. It can be noticed that there is a big enhancement at frequencies just below the Nyquist limit (0.5 mm$^{-1}$). This enhancement is similar to the adjacency effects observed in the MTF of some film-screen systems (Dainty and Shaw 1974) and in Xeroradiography (Boag 1973). Adjacency effects are often considered to be a desirable feature in an imaging system, as they may produce images which are subjectively sharper than those produced in the absence of these effects (Hay 1982). However, their presence can complicate the description of the input-output relationships of the system.
The shape of the horizontal MTF can be described in terms of an enhancement filter. Enhancement and restoration filters have been used, for example, in SPECT imaging in an attempt to recover the resolution lost in the detection process (Gillard et al. 1988). These filters exceed unit gain over a given frequency range and roll-off to zero gain at higher frequencies in order to avoid noise amplification. One such a filter is the Metz filter, which has a frequency response that resembles the measured horizontal MTF of the SDRD. A general transform based on the mathematical representation of the Metz filter in the spatial frequency domain is given by

$$MTF_H(u) = \frac{1 - [1 - m^a(u)]^a}{m(u)}$$

(3.4)

where $m(f)$ is the MTF before filtering, $MTF_H(f)$ is the filtered MTF. The Metz filter corresponds to the particular case when $a_1 = 2$. The parameter $a_2$ determines the extent to which the filter follows the inverse MTF before rolling-off to zero. A very good fit to the measured data can be obtained by assuming an exponential MTF before filtration of the same form as given by equation 3.1, that is

$$m(u) = \exp[-2(\pi a_H u)^3]$$

(3.5)
and applying to this function the generalised Metz transform given by equation 3.4. The dashed line in figure 3.14 shows how well equations 3.4 and 3.5 fit the experimental data. The parameters used in this case were $\sigma_H = 0.71$, $a_1 = 1.24$ and $a_2 = 1.95$.

Figure 3.15a shows the horizontal MTFs for the 3 mm collimator data at different count rates. For clarity only the fits obtained with equations 3.4 and 3.5 are shown. As the count rate was reduced, the edge enhancement effect was less pronounced. It was found that exactly the same exponential function (i.e. with $\sigma_H = 0.71$) could be used to fit all the MTFs obtained at different count rates. Furthermore, all the curves in figure 3.15a were obtained using $a_2 = 1.95$, and the only parameter that had to be adjusted in order to account for the variation with count rate was $a_1$.

![Figure 3.15. (a) Horizontal MTF for the 3 mm collimator data at different count rates. For clarity only the fits obtained using equations 3.4 and 3.5 are shown. (b) Variation of the fitting parameter $a_1$ as a function of count rate.](image)

The values used for $a_1$ varied from 1.24 for the image acquired at the highest count rate, up to 1.42 for the image acquired using the lowest count rate. A plot of $a_1$ as a function of the mean number of counts per pixel is shown in figure 3.15b. The variation of this parameter can be described by a decreasing exponential of the form $a_1 = b_0 \cdot \exp(-b_1 N_s) + b_2$, with $b_0$, $b_1$, $b_2$ equal to 0.25, 0.20 and 1.21, respectively. This semi-empirical parametrisation of the MTF can therefore be used to characterise the degree of enhancement that will be present in the MTF at a given count rate.
3.3.3.2. HR system

The rate dependence of the horizontal MTF of the SDRD can be explained in terms of the difference on counting rate efficiency between the anti-coincidence (A-) and coincidence (C-) channels in the multiwire chamber, which has already been discussed in Chapter 2 (§2.6.1). Figure 3.16a shows the horizontal ESFs of an A-channel and a C-channel obtained from an uncorrected image of a Ta edge taken with the HR system. The oversampled ESFs correspond to consecutive A- and C- channels. It can be observed that the number of counts per pixel in the C-channel is approximately 30% lower than in the A-channel. More importantly, the shape of the ESFs is also seen to be dependent upon the channel counting mode. The edge enhancement effect is present on the A-channel only, while the shape of the C-channel ESF resembles the ESFs obtained in the vertical direction. Again, the spatial extent of the edge effect is ~1 mm, which corresponds to the separation between anode wires in the chamber.

![Figure 3.16](image)

**Figure 3.16.** (a) Edge response functions of the A- and C-channels. (b) MTF calculated from the ESFs shown in figure (a).

The edge enhancement effect observed in the A-channel is a result of the dependence of the relative counting efficiency of the chamber with counting rate (see §2.6.1, figure 2.9). A gradual decrease in count rate due, for example, to partial obscuration of an A-channel by the edge, means that neighbouring wires will be working with very different efficiencies. A partially obscured A-channel will operate at lower count rates than those being fully irradiated and will have, therefore, higher relative efficiency. An A-channel reaches maximum efficiency when working at a count rate near 250
kHz; any further decrease in count rate (e.g. for channels deep into the shadow produced by the edge) will cause a drop in efficiency. Obviously, the difference in counting rate efficiency between partially obscured neighbouring wires would be smaller at lower photon fluxes than at higher ones. Therefore, the edge enhancement would be expected to be reduced with decreasing counting rate, as is actually observed in the experimental data (see figure 3.9).

Figure 3.16b shows the MTFs of the A- and C- channels calculated from the ESF data shown in figure 3.16a. The MTF of the A-channel has the same shape as the horizontal MTF of the ST system. The C-channel, on the other hand, does not show any edge enhancement effect. Unlike the ST system, both MTFs are practically zero at the Nyquist frequency (1 mm\(^{-1}\)). The C-channel MTF was fitted using a Gaussian function of the form given by equation 3.5, with \(\sigma_H = 0.39\). Applying the generalised Metz transform to the Gaussian MTF, using \(a_1 = 2.3\) and \(a_2 = 2.2\), an excellent agreement was obtained between the A-channel MTF and the transformed data, as shown by the dotted line in the same figure. Therefore, a single function can be used to describe the MTF of the A- and C- channels.

If the horizontal MTF of the HR system is calculated using the ESF data obtained from a renormalised image (see §2.6.1) then the MTF of the C-channel is identical to that of the A-channel. That is, the effect of renormalisation is to reshape the C-channel ESF in such a way that it takes the form of the A-channel ESF, and exactly the same fit as described above can be used to explain the form of the horizontal MTF.

Table 3.6 shows a summary of the nominal values for the horizontal sampling distance and Nyquist frequency of the SDRD, along with the parameters of the Gaussian LSF used to fit the measured data. A change of the aft-slit aperture from 3 to 1 mm has a small effect on the LSF, reducing the FWHM\(_H\) by about 10%. For the HR system the FWHM\(_H\) is at least 38% smaller than that of the ST-SDRD, which represents a significant improvement in the spatial transfer characteristics.

<table>
<thead>
<tr>
<th>System</th>
<th>fore-slit</th>
<th>aft-slit</th>
<th>(\Delta x)</th>
<th>(f_{\Delta x}) [mm(^{-1})]</th>
<th>(\sigma_H)</th>
<th>FWHM(_H)</th>
</tr>
</thead>
<tbody>
<tr>
<td>ST</td>
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<td>1</td>
<td>0.5</td>
<td>0.71</td>
<td>1.67</td>
</tr>
<tr>
<td>ST</td>
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<td>1</td>
<td>1</td>
<td>0.5</td>
<td>0.64</td>
<td>1.51</td>
</tr>
<tr>
<td>HR</td>
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<td>1</td>
<td>0.5</td>
<td>1</td>
<td>0.39</td>
<td>0.92</td>
</tr>
</tbody>
</table>
3.4. Noise and dSNR analysis

The noise properties of the SDRD were determined from a series of images taken under uniform irradiation conditions. The noise samples were acquired under the same exposure conditions used for the contrast-detail measurements. All images were corrected on-line for spatial non-uniformities using the SDRD system software. In the HR system the data were also renormalised in order to account for the difference in counting rate efficiency between A- and C- channels.

Figure 3.17 shows the measured standard deviation as a function of the mean number of counts per pixel as determined from the digital images. For the ST system the standard deviation is observed to vary as the square root of the mean output signal, which means that the system noise follows a Poisson distribution. The square root behaviour is independent of the incident beam quality (THX or ABDO), as it would be expected for a photon counting detector.

![Figure 3.17. Measured standard deviation as a function of the mean number of counts per pixel. (a) ST system. (b) HR system.](image)

On the other hand, the noise in the HR images varies linearly with the mean number of counts. The pixel value histogram was observed to follow a Gaussian distribution with a larger standard deviation that what would be expected from Poisson statistics, which indicates the presence of additional sources of noise. The additional noise contributions may be explained in terms of the renormalisation procedure used to
correct the HR images. Although the correction algorithm reduces the variations due to differences in counting rate efficiency from 35% down to less than 1% (Babichev et al 1992), these additional fluctuations have an effect upon the noise properties of the system.

A series of images of the TO.10 phantom were used to investigate the effect that the difference in count rate and the additional fluctuations in the HR system would have upon the contrast and the detail signal-to-noise ratio characteristics of the digital images. Figures 3.18a and 3.18b show images of a section of TO.10 taken with the ST and the HR systems, respectively. Both images were taken using the same tube voltage and filtration (70 kVp, THX), adjusting only the tube current in order to obtain the same entrance air kerma, which was 20 μGy in this case. No image processing was done in any of these images, apart from overall contrast enhancement. The difference in terms of resolution can be clearly appreciated from these images.

![Figure 3.18](image.png)

**Figure 3.18.** Section of the TO.10 phantom imaged with the SDRD under the same exposure conditions (20 μGy entrance dose). (a) ST system. (b) HR system.

Table 3.7 compares the nominal and measured contrasts for the test detail A1 (11.1 mm diameter) under different exposure conditions. The contrast computed from the measured pixel values in the HR images were found to be within 3% of the nominal contrast values, while the contrasts obtained with the ST system underestimated the nominal contrasts by about 20%. This behaviour might be explained by the difference in counting rate efficiency between the ST- and the HR-SDRD. In the HR-SDRD the number of counts over the detail ($N_1$) and over the background ($N_0$) corresponded to relatively low count rates (< 200 kHz), where the difference in count rate efficiency is small (see figure 2.9). In the ST system the difference in counting rate between $N_0$ and
was much bigger, in such a way that $N_p$ (higher count rate) was reduced due to lower counting rate efficiency, thus decreasing the image contrast.

**Table 3.7.** Comparison of nominal and measured contrasts produced by detail A1 under different exposure conditions with the SDRD.

<table>
<thead>
<tr>
<th>System</th>
<th>kVp</th>
<th>Filtration</th>
<th>C (nominal)</th>
<th>C (measured)</th>
</tr>
</thead>
<tbody>
<tr>
<td>ST</td>
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<td>THX</td>
<td>27.7</td>
<td>21.2</td>
</tr>
<tr>
<td>ST</td>
<td>70</td>
<td>ABDO</td>
<td>16.0</td>
<td>12.4</td>
</tr>
<tr>
<td>HR</td>
<td>70</td>
<td>THX</td>
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<td>THX</td>
<td>23.6</td>
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<tr>
<td>HR</td>
<td>80</td>
<td>ABDO</td>
<td>13.1</td>
<td>13.5</td>
</tr>
</tbody>
</table>

Figure 3.19 shows the measured $dSNR$s of detail A1 as a function of the entrance air kerma for the ST and the HR systems. The dotted lines in the same figure represent the square root behaviour predicted by the model of Motz and Danos (equation 1.43). It can be observed that the $dSNR$s for the HR system follow a square root behaviour despite the fact that the noise is not Poisson, which indicates that the total noise is dominated by the contribution due to quantum fluctuations.

![Figure 3.19. Comparison of the $dSNR$ produced by object A1 (11.1 mm diameter, 27.7% nominal contrast) as a function of the entrance air kerma for the ST and the HR systems.](image)

Since the nominal pixel size of the ST-SDRD is twice that of the HR system, the $dSNR$ of the ST system would be expected to be a factor of 2 higher than that of the
Chapter 3 Evaluation of the MWPC system

HR system, according to the $dSNR$ models described in §1.7. The reduction obtained from the measured data was a factor of 1.9, which matches the expected value.

3.5. Clinical images and doses

3.5.1. Clinical images

The SDRD has been in clinical use for approximately 6 years, and during this time a large number of clinical investigations have been undertaken. There are four systems currently working at hospitals in Novosibirsk and Moscow (Babichev et al 1991a, 1991b). The first two systems were installed in 1986 and 1989 at the Moscow Clinic for Mother and Child Health Protection, where they have been mainly used for specialised studies such as pelvimetry and hysterosalpingography. A third device was installed in 1989 at the Novosibirsk Regional Hospital, where it has been used for studies of lung ventilation in normal and pathological conditions, general chest screening procedures and imaging of the spine and head. The fourth device is at the Budker Institute of Nuclear Physics in Novosibirsk, where it is used by the Health Department of the institute. During 1992 approximately 4000 examinations were carried out between these four devices.

![Figure 3.20](image.png)

**Figure 3.20.** Hysterosalpingography image taken with the ST-SDRD. Exposure factors 70 kVp, 50 mA, 0.3 mm Cu filter. Entrance skin dose 30 μGy.

Figure 3.20 shows an example of a hysterosalpingography study made with the ST-SDRD. Exposure conditions were 70 kVp and 50 mA, 0.3 mm Cu filtration, with an
Chapter 3 Evaluation of the MWPC system

entrance skin dose of $30 \pm 5 \mu\text{Gy}$. This image demonstrates the ability of the system to record fine detail when enhanced by the use of contrast media. In this case an object of less than one millimetre diameter (a partially obstructed fallopian tube) can be seen.

![Figure 3.21](image1.jpg)  
(a)  
(b)

**Figure 3.21.** Images of a lumbar spine study. (a) AP projection, taken at 90 kVp, 50 mA; entrance skin dose 86 μGy. (b) Lateral projection, exposure factors 100 kVp, 80 mA, 0.1 mm Cu filter; entrance skin dose 207 μGy.

![Figure 3.22](image2.jpg)

**Figure 3.22.** Image of a thorax, Postero-Anterior projection. Exposure conditions 70 kVp, 20 mA, 0.1 mm Cu filter; entrance skin dose 21 μGy.

Figure 3.21 shows the Antero-Posterior and the Lateral projections of a lumbar spine examination. The exposure factors were 90 kVp, 50 mA and 100 kVp, 80 mA with 0.1 mm Cu added filtration, giving an entrance skin dose of 86±10 and 207±12 μGy for the AP and the Lateral projections, respectively. Finally, figure 3.22 shows an
image of a Postero-Anterior projection in a thorax examination. This image was taken at 70 kVp, 20 mA, 0.3 mm Cu filtration, and the dose in this case was $21 \pm 7 \mu\text{Gy}$.

The images shown in figures 3.20-3.22 are examples of three common studies carried out with the SDRD in the hospitals where it has been installed. Along with some other common examinations they were used as the basis of a comparison of the dose performance of the SDRD with that of film-screen systems used in normal clinical practice in England.

### 3.5.2. Clinical doses

A comparison has been made between the entrance dose values for a series of clinical examinations made with the SDRD in Russia and with conventional film-screen systems used in routine clinical practice in England. Absorbed dose values for the clinical images were measured using LiF TLD chips placed directly on the patient's skin and include, therefore, the effects of scattering of the beam by the patient. Mean values of entrance skin dose per image for these examinations are shown in table 3.8. The last column of the same table shows the values of the mean entrance skin dose per radiograph obtained by the National Radiological Protection Board (NRPB) in a national survey of doses to patients undergoing routine x-ray examinations in England (Shrimpton et al 1986).

<table>
<thead>
<tr>
<th>Examination</th>
<th>Projection</th>
<th>kVp</th>
<th>mA</th>
<th>Entrance skin dose [mGy]</th>
<th>SDRD</th>
<th>NRPB*</th>
</tr>
</thead>
<tbody>
<tr>
<td>Chest</td>
<td>PA</td>
<td>70</td>
<td>20</td>
<td>0.015</td>
<td>0.23</td>
<td></td>
</tr>
<tr>
<td>Chest</td>
<td>Lat.</td>
<td>80</td>
<td>30</td>
<td>0.032</td>
<td>3.46</td>
<td></td>
</tr>
<tr>
<td>Abdomen</td>
<td>AP</td>
<td>70</td>
<td>50</td>
<td>0.030</td>
<td>8.43</td>
<td></td>
</tr>
<tr>
<td>Skull</td>
<td>AP</td>
<td>80</td>
<td>50</td>
<td>0.058</td>
<td>4.37</td>
<td></td>
</tr>
<tr>
<td>Lumbar spine</td>
<td>AP</td>
<td>90</td>
<td>50</td>
<td>0.086</td>
<td>9.19</td>
<td></td>
</tr>
<tr>
<td>Lumbar spine</td>
<td>Lat.</td>
<td>100</td>
<td>80</td>
<td>0.292</td>
<td>22.80</td>
<td></td>
</tr>
</tbody>
</table>

* from Shrimpton et al 1986

From this table it can be seen that the SDRD doses are consistently lower than the ones reported by the NRPB for the equivalent examinations. Dose saving factors
range from about 15 for the thorax examination to approximately 280 for the abdominal examination.

3.6. Conclusions

This chapter presented the results of an evaluation of the imaging performance of a low-dose digital x-ray system based on a multiwire proportional chamber. Contrast-detail-dose analysis was used to compare the imaging performance of the SDRD with that of a film-screen combination in routine clinical use. Inspection of the contrast-detail and contrast-dose diagrams has shown that performance of the SDRD is better than that of film-screen for medium and large low-contrast objects, but is poorer for the smaller details.

A model of threshold detection was used to explain the shape of the SDRD contrast-detail diagrams. The analysis showed that statistical considerations are sufficient to explain its behaviour. The threshold detection model reproduced the contrast-detail data for all but the largest detail sizes contained in the test object.

The advantages of the SDRD over film-screen were demonstrated using contrast-dose diagrams. High quantum efficiency and virtually zero background in the multiwire chamber allowed the system to work at very low exposure levels. The contrast-dose plots also demonstrated the wide dynamic range over which the SDRD can image in a single examination. No effective dose threshold was observed, since the detector works in counting mode of x-ray detection and, therefore, contrast sensitivity varies simply as the inverse of the square root of the dose.

The resolution properties of the SDRD were investigated using modulation transfer function analysis. It was found that the spatial transfer characteristics of the system depend on different factors for the vertical and horizontal directions. In the vertical direction the MTF was determined by geometrical factors only, such as fore- and aft-slit apertures, and sampling distance. It was also shown that the MTF remained practically the same for counting rates that differed by up to one order of magnitude.

The MTF in the horizontal direction was determined not only by the geometrical characteristics of the chamber, but also by the way in which the detector registers each event. A count-rate dependent edge-enhancement effect was observed in the MTFs measured in this direction. A semi-empirical model was used to demonstrate that this
Chapter 3 Evaluation of the MWPC system

effect can be explained in terms of the dependence of counting rate efficiency on count-rate and on the rate of coincidence-event rejection in anti-coincidence channels.

The noise properties of ST and HR were studied using first order statistics. It was found that the noise in the ST system followed a Poisson distribution, independently of the energy distribution of the incident x-ray beam, as it would be expected for a semi-ideal photon counting detector. The noise in the HR-SDRD was observed to increase linearly with the mean output signal, which indicated the presence of additional sources of noise. This confirms the observation made on the contrast-detail measurements, in which the threshold signal-to-noise ratio for detectability was found to be slightly higher for the HR system.

Clinical doses for the SDRD have been compared to those currently achieved in the UK for the same type of examinations. The most significant differences occurred in studies of the abdomen, where savings up to a factor of ~300 could be made with the SDRD. The difference in dose performance between the two systems is due to the fact that the exposure needed by the film-screen combination for a given imaging task depends not only on the dynamic range of the object to be imaged but also on the non-linear characteristics of film, which limit its usefulness to a certain range of exposures. On the other hand, the SDRD has no effective dose threshold.
Chapter 4

The Charge Coupled Device

4.1. Introduction

The charged coupled device (CCD) is a silicon integrated circuit based on a metal-oxide-semiconductor (MOS) structure. The CCD concept was first proposed in 1970 by Boyle and Smith, and the first experimental verification was given by Amelio and co-workers the same year. Thanks to the fact that highly developed MOS fabrication techniques already existed at the time of its invention very rapid progress was achieved during the first years after its introduction. The intrinsic position sensitive capabilities of CCDs and their good response in the visible region of the spectrum made them immediately attractive for imaging applications. Astronomers were among the first to realise the potential of these detectors for scientific applications. During the last decade CCDs have revolutionised optical astronomy (Fraser 1989).

4.2. CCDs as x-ray imaging detectors

The need of high energy-resolution detectors for imaging spectroscopy in x-ray astronomy (Lumb 1990) prompted the development of scientific grade CCDs with enhanced x-ray response. X-ray sensitive CCDs have also been developed to be used in x-ray crystallography (Fiorucci et al 1990), and the study of synchrotron radiation (Gruner 1989, Clarke 1990).

In general terms, CCDs have been used in two ways as x-ray detectors: by direct conversion of the radiation in the silicon, or by first converting the x-rays to visible light in a phosphor and imaging this light either directly or with light amplification using an image intensifier. The use of CCDs in medical x-ray imaging has been limited primarily by two reasons. First, typical CCD sizes are only of the order of a few square centimetres, and second, radiation damage considerably limits its useful working life. However, during the last few years sophisticated manufacturing
techniques have made possible to construct large area (up to $5 \text{ cm}^2$) scientific grade devices with improved radiation hardness.

In this chapter the main properties relevant to CCD operation are briefly reviewed, and the basic principles of x-ray detection with CCDs are described. Particular emphasis is put on those aspects of x-ray detection that are directly related with the experimental work carried out during this investigation.

4.3. CCD format and operation

CCDs are fabricated by depositing a series of overlapping gate electrodes on the oxidised surface of a silicon wafer. Each electrode is equivalent to the gate of a MOS capacitor (see figure 4.1). The operation of the simplest CCD structure can be briefly summarised as follows: when the electrodes are biased to a suitable voltage a matrix of potential wells (pixel sites) is produced under the oxide-semiconductor surface. Any free charge liberated by the absorption of photons in the silicon is collected in the potential wells.

![Figure 4.1. Cross sectional view of a MOS capacitor.](image)

The amount of charge stored in each well is proportional to the energy deposited by the light (or x-rays) incident on that pixel. After a given image integration time, the electrodes are clocked to move the wells, and their associated pattern of charge, across the surface of the CCD to an output node. A serial train of video pulses is detected and the position of any pulse in the train is uniquely associated with a particular pixel, while its magnitude reveals the intensity of x-rays at that point. The two main processes relevant to CCD operation are called charge storage and charge coupling, and are described in more detail in the next sections.
4.3.1. Charge storage

Before the application of the gate bias ($V_g$) there is a uniform distribution of holes in the p-type semiconductor. If a positive voltage is applied to the gate electrode the holes are repelled from the section of the semiconductor immediately beneath the gate, creating a depletion layer. As the gate bias is increased the depletion region extends further into the semiconductor, and the potential at the insulator-semiconductor interface (called the surface potential) becomes increasingly positive. Eventually, a gate bias is reached at which the surface potential becomes so positive that electrons (minority carriers) are attracted to the oxide interface, where they form a very thin two-dimensional conductivity channel, called the inversion layer. If minority carriers are generated due to photon interactions in the silicon the depletion layer will shrink and the surface potential will fall as the inversion layer charge increases. This mechanism can be interpreted as the creation of a potential well with a depth proportional to the gate voltage. For a given $V_g$ the depth of the well will decrease linearly with the amount of charge present in the inversion layer.

4.3.2. Charge coupling

Charge coupling is the technique used to transfer signal charge from under one electrode to the next one.

Figure 4.2. Movement of collected charge packets by clocking of electrode voltages: a) potential well with collected charge; b) well just created; c) original well collapsing; d) charge transfer completed.

Figure 4.2 shows an arrangement of four closely spaced electrodes. It can be assumed
that some charge is stored under the second electrode, which is biased to 10 V, and that all other electrodes are biased just above the threshold voltage (e.g. 2 V in this case). If the bias on the third electrode is increased to 10 V then both potential wells will merge and the charge stored will be shared. Finally, if the potential on the second electrode is reduced to 2 V then the remaining charge content will be shifted to the third electrode. The whole process might be repeated a number of times, and thus the charge can be transferred along a series electrodes by clocking of the gate voltages.

### 4.3.3. Buried channel CCDs

Most modern frame-transfer CCDs (see §4.3.4.) are based on an alternative MOS structure called Buried (or Bulk) channel CCD. The basic BCDD (see figure 4.3) comprises a lightly doped p-type substrate on top of which there is a thin layer (a few microns) of a more heavily doped n-type semiconductor. The surface is covered with the oxide insulator and a series of metal electrodes, as in the surface channel CCDs.

![Figure 4.3. Formation of the field-induced ($x_j$) and junction ($x_e$ and $x_j$) depletion regions in a buried channel CCD. The position of the inversion layer is indicated by the line $z$-$z'$.](image)

In this CCD structure the potential wells are formed not at the semiconductor-insulator interface, but some distance into the bulk semiconductor. This is done to avoid the effect of charge trapping centres which are located near the surface. BCCDs have the main advantage of a much improved noise performance, but at the expense of reduced charge handling capability, which translates into a smaller dynamic range.
4.3.4. Frame transfer devices

Most CCDs currently under investigation for scientific applications are buried channel, 3-phase, frame transfer devices. The operation and design of this particular CCD architecture are illustrated in figure 4.4, which shows a simplified representation of an $8 \times 5$ pixels frame transfer CCD. In this design a set of horizontal electrodes are grouped into an upper photosensitive section and a lower frame-store section which is covered by an opaque shield. A serial line readout section located below the frame-store is used to transfer image information line by line to the on-chip amplifier for video output.

![Figure 4.4. Simplified representation of a typical 3-phase frame transfer CCD. $I$, $S$ and $R$ represent the clock pulses of the image, store and readout sections, respectively. The arrows indicate the direction of charge transfer.](image)

In a 3-phase CCD the electrodes are connected cyclically in triplets which means that three electrodes constitute the height of one picture element. The pixel width is determined by heavily doped p-type implants, called channel stops, which prevent the diffusion of charge along the length of the electrodes. For example, in figure 4.4 the photosensitive region is formed by 15 electrodes, controlled by the clock pulses $I_1$, $I_2$ and $I_3$. The store section has a similar structure and is controlled by the clock pulses $S_1$, $S_2$ and $S_3$.

During image integration time a positive bias is applied to one set of interconnected
electrodes (say, \( I_i \)) in the image section. If the image section is illuminated some photons will penetrate the electrode structure and generate electron-hole pairs in the underlying silicon substrate. The electrons diffuse to the nearest biased electrode where they are collected as signal, while the holes diffuse into the substrate where they are effectively lost.

Light or x-ray induced signal charges will accumulate as long as the gate voltage is maintained. Once the image integration period finishes the charges are transferred very rapidly into the frame-store section by clocking the image and frame-store electrodes in the way described in section 4.3.2. The frame-store is read out line by line into the output node while a new integration period takes place.

4.3.5. Anti-blooming

The term blooming is used to describe a local overload in the CCD due, for example, to very intense illumination. In some devices a special structure (called anti-blooming) is incorporated within the image section of the device in order to drain off any excess of photogenerated charge before it can spread from the point of overload. Anti-bloomed devices can tolerate extremely high optical overloads, typically between \( \times 100 \) and \( \times 1000 \) of the full-well capacity. However, the drain structure takes up some of the photosensitive area, thereby reducing quantum efficiency and full-well capacity.

4.3.6. Optical response

The effective range of the optical spectrum over which a CCD sensor gives useful output depends on its quantum efficiency (which is a measure of the light to charge conversion process), on the signal handling capability once these charges have been generated and on the noise performance of the device. The numerical value that these parameters can take depends on the particular CCD architecture and on the illumination mode under which they are used. There exist two illumination modes, front-side and back-side. In front-side illuminated devices the light is incident directly upon the electrode structure whereas in back-side illumination the light reaches the sensitive region through the substrate, which has been previously thinned.

4.3.6.1. Quantum efficiency

The quantum efficiency of a CCD determines its sensitivity to incident radiation. In
front-side illuminated devices only those photons that penetrate the electrode layer and generate an electron that is captured on a potential well contribute to the output signal. The penetration depth of light in the visible region increases strongly at larger wavelengths. However, electrons generated more deeply into the silicon can diffuse back towards the potential wells in the front of the CCD where they are eventually collected. Taking the diffusion process into account the quantum efficiency of the sensor for monochromatic light of wavelength $\lambda$ is given by (EEV 1987)

$$
\eta_{\lambda} = \left[1 - \frac{e^{-L_d \alpha(\lambda)}}{1 + L_o \alpha(\lambda)}\right] T_{\lambda}
$$

(4.1)

where $\alpha(\lambda)$ is the silicon optical absorption coefficient, $L_d$ is the depth of the depletion layer, $L_o$ is the electron diffusion length and $T_{\lambda}$ is the electrode transmission. Typical values for $L_d$ and $L_o$ in commercially available CCDs are of the order of 5 $\mu$m and 50 $\mu$m, respectively. Approximate values for $\alpha$ in the visible region of the spectrum range from 5 $\mu$m$^{-1}$ at 400 nm down to 0.1 $\mu$m$^{-1}$ at 800 nm.

Figure 4.5 shows a plot of equation 4.1 using the parameters given above and assuming perfect electrode transmission ($T_{\lambda}=1$). It can be observed that equation 4.1 predicts practically perfect quantum efficiency between 400 and 700 nm, followed by
a sharp decrease towards the near infrared.

### 4.3.6.2. Responsivity

An alternative approach to quantify absorption efficiency consists in measuring both the power of the incident illumination using a radiometer or a photometer and the resulting output signal as an electric current. The ratio of the photogenerated current to the input illumination power at wavelength $\lambda$ is called the *responsivity* of the device and is normally given in units of current per unit power. This parameter is related to the quantum efficiency through the equation

$$ R_\lambda = \frac{q_e \lambda}{hc} \eta_\lambda $$  \hspace{1cm} (4.2)

where $q_e$ is the electron charge ($1.6 \times 10^{-19}$ C), $h$ is Plank's constant ($6.63 \times 10^{-34}$ Js) and $c$ is the speed of light ($3 \times 10^8$ ms$^{-1}$). Since the photogenerated current is the sum of the outputs from all the illuminated elements then no picture information is conveyed by this quantity. If $N_p$ pixels are under illumination and the CCD is operated at a frame rate $F_R$ (Hz) then the average charge per element $Q_{av}$ and the output current $I_o$ are related by

$$ I_o = N_p F_R Q_{av} $$ \hspace{1cm} (4.3)

where

$$ I_o = R_\lambda P_\lambda $$ \hspace{1cm} (4.4)

and $P_\lambda$ is the total illumination power. Using equations 4.2 and 4.3 the average charge per pixel can be calculated in terms of quantities that can be measured experimentally. This in turn permits to calculate the average number of charge carriers $N_e$ (of charge $q_e$) per pixel collected during an integration time $t_I$

$$ N_e = \frac{Q_{av}}{q_e} = \frac{R_\lambda W_\lambda A_{pix}}{q_e F_R} = \frac{R_\lambda W_\lambda A_{pix} t_I}{q_e} $$ \hspace{1cm} (4.5)

where $W_\lambda$ is the power density (or irradiance) and $A_{pix}$ is the CCD pixel area.

Figure 4.6 shows the measured quantum efficiency and responsivity as a function of wavelength for a three-phase, front-illuminated, non anti-bloomed device of the kind used in this study (EEV-02). It can be seen that there is a severe drop in quantum efficiency at lower wavelengths caused by the absorption of visible photons on the
electrode layers. The quantum efficiency peaks at a wavelength near 650 nm and does not exceed 50%. The responsivity of the device is shown in the same graph, and has a peak value of ~250 mA/W.

![Graph showing quantum efficiency and responsivity as a function of wavelength.]

Figure 4.6. Responsivity and quantum efficiency as a function of wavelength for a front-illuminated, non anti-bloomed device (EEV-02).

4.3.7. Resolution

The resolution of a CCD is normally expressed in the frequency domain in terms of the modulation transfer function. The MTF of a bare CCD is determined by three components, the geometric, transfer and diffusion MTFs. The geometric component is due to the finite spatial sampling of the CCD. The transfer component arises because of possible inefficiencies in the charge transfer process. Finally, the diffusion MTF is due to photon absorption below the depletion region.

4.3.7.1. Geometric MTF

In a typical frame transfer device the pixel size \( p \) is practically the same as the distance from pixel to pixel. In this case the geometric MTF is given by

\[
MTF_G(f) = \frac{\sin(\pi pf)}{\pi pf} = \text{sinc}(pf)
\]  

(4.6)
4.3.7.2. Transfer MTF

Resolution can also be degraded by charge transfer inefficiency. A fundamental limitation of CCDs is that, although in principle no charge is lost, some residual charge is left behind after each transfer. The total MTF will be degraded by a factor

\[ MTF_T(f) = \exp\left(-n_\varepsilon[1 - \cos(2\pi f)]\right) \]  

(4.7)

where \( n_\varepsilon \) is the number of transfers from the point of charge generation to the output node and \( \varepsilon \) is the fraction of charge left behind at each transfer (Beynon and Lamb 1980). This effect is very small in buried channel devices since \( \varepsilon \) is normally very small (\( \approx 10^{-5} \)).

4.3.7.3. Diffusion MTF

In practice, the resolution limit is close to the geometric MTF for light of visible wavelengths (400-650 nm). However, at higher wavelengths photons can be absorbed deeper into the silicon, away from the depletion region, and the signal electrons diffuse sideways as they drift before they are collected. Seib (1974) developed a model to describe the effect of minority carrier diffusion on the MTF for both front and back illuminated devices. Only the front illuminated case will be discussed here.

The diffusion MTF is obtained as the solution to the diffusion equation by using the appropriate expression for the carrier generation rate and boundary conditions. The solution obtained by Seib is given by

\[ MTF_D(f) = \frac{1 - \left[ \exp(-\alpha L_\lambda)/(1 + \alpha L_\lambda) \right]}{1 - \left[ \exp(-\alpha L_\lambda)/(1 + \alpha L_\lambda) \right]} \]

(4.8)

where

\[ L^{-2} = L_\lambda^{-2} + (2\pi f)^2 \]  

(4.9)

The total MTF of a CCD would be given as the product of the three components

\[ MTF_{total}(f) = MTF_G(f) \times MTF_T(f) \times MTF_D(f) \]  

(4.10)

Figure 4.7 shows a plot of the geometric and diffusion MTF (equations 4.6 and 4.8) using representative parameters of TV format CCDs and a pixel size of 25 \( \mu \text{m} \). The diffusion MTF is shown for 600 and 800 nm, with \( \alpha(\lambda) = 0.5 \) and \( 0.1 \, \mu\text{m}^{-1} \),
respectively. It can be seen that the geometric MTF of the CCD practically dominates at optical wavelengths. However, the diffusion MTF degrades very rapidly towards the near-infrared. At a spatial frequency of 20 mm\(^{-1}\) the diffusion MTF is reduced from 0.986 at 600 nm to only 0.735 at 800 nm.

![Figure 4.7. Geometric and diffusion components of the MTF for a 25 \(\mu\)m pixel CCD in the optical region of the spectrum. MTF\(_g\) is shown for two different wavelengths, 600 and 800 nm.](image)

### 4.4. Direct x-ray detection

#### 4.4.1. Semiconductor x-ray detection

The result of x-ray absorption in a semiconductor is the production of a number of electron-hole (e-h) pairs which is proportional to the energy \(E\) of the incoming x-ray photon, that is

\[
N_e = \frac{E}{w_i}
\]

where \(w_i\) is known as the ionisation energy and represents the energy required to create one e-h pair. For silicon at room temperature \(w_i\) is approximately 3.65 eV. This quantity is analogous to the number of electron-positive ion pairs in a counting gas. Since \(w_i\) is ~10 times smaller than in gases, ten times as many charge carriers are
Chapter 4 The Charge Coupled Device

liberated by the absorption of one x-ray photon. The statistical fluctuations in \( N_e \) can be quantified in terms of a Fano factor, that is, the variance can be expressed as \( \sigma_{N_e}^2 = FN_e \), with \( F \) of the order of 0.1 for Si at 77 °K (liquid nitrogen temperature). These characteristics, along with their relatively high efficiency at low x-ray energies, mean that semiconductor detectors are capable of much better energy resolution than any form of gas counter and have made them the detector of choice for soft x-ray spectroscopy.

4.4.2. The depletion depth

Most practical semiconductor x-ray detectors consist of a reversed biased p-n or p-i-n junction diode. Fraser (1989) has made a comprehensive review of the most common semiconductor detector designs used in x-ray detection. In all cases it is the region in the vicinity of the junction itself which acts as the active volume of the detector. This region is what it has been called the depletion region in the previous sections.

Within this depletion region an electric field is set up which sweeps out the photogenerated charge carriers towards the sides of the junction in which they are the majority carriers. The width of the depletion region (the depletion depth) is given by

\[
L_d = \sqrt{\frac{2\kappa e V_G}{q_e N_{AD}}} = \sqrt{\frac{2\kappa e \mu_e \rho_s V_G}{q_e N_{AD}}}
\]  

(4.12)

where \( \kappa \) is the dielectric constant of the semiconductor, \( V_G \) the potential difference applied across the junction (gate voltage), \( q_e \) the electron charge, \( N_{AD} \) the dopant concentration on the side of the junction with the lowest impurity level, \( \mu_e \) the electron mobility and \( \rho_s \) the resistivity of the semiconductor. Typical values for Si are \( \kappa = 12 \) and \( \mu_e = 0.135 \, \text{m}^2/\text{V} \cdot \text{s} \).

4.4.3. Quantum efficiency

The x-ray quantum detection efficiency of a semiconductor detector can be written as

\[
\eta_q(E) = \left( \Pi_m e^{-\mu_m l_m E} \right) \left( 1 - e^{-\mu_m l_m E} \right)
\]  

(4.13)

independently of the detailed detector geometry. The first term accounts for the absorption of x-rays before they reach the active region, with \( l_m \) and \( \mu_m \) the thickness
and linear attenuation coefficient of the $m$-th absorbing layer, respectively. This term determines the low energy cutoff response of the detector. The second term is the absorption efficiency of the depletion region, with $\mu_s$ the absorption coefficient of the semiconductor. It should be noticed that equation 4.13 only determines the spectroscopic quantum efficiency of the detector. In cases where the depletion region does not extend throughout the complete detector (i.e., the detector is not "fully-depleted") x-rays can be also be absorbed in the field-free region immediately below the depletion layer and contribute to the total measured signal.

![Mass attenuation coefficients for silicon in the diagnostic energy range.](image)

**Figure 4.8.** Mass attenuation coefficients for silicon in the diagnostic energy range.

The mass attenuation coefficient of silicon at diagnostic x-ray energies is shown in figure 4.8. In the low energy range the photoelectric effect is the dominant interaction. However, as the energy increases the photoelectric contribution decreases very sharply while Compton scattering increases steadily, until it becomes more prominent above ~55 keV. For example, the absorption efficiency of a 100 $\mu$m Si slab drops from 55% at 10 keV to only ~3% at 30 keV.

### 4.4.4. Deep depletion CCDs

Equations 4.12 and 4.13 indicate that quantum efficiency improves by increasing the gate voltage and/or the resistivity of the semiconductor. However, the gate potential of most CCDs is restricted to be of the order of 10 V, mainly due to limitations in
electrode construction and insulation (Flint et al. 1992a). Therefore, the only way to increase x-ray sensitivity is by using thicker epitaxial† (Epi) devices made of higher resistivity silicon.

Commercially available CCDs are normally fabricated from low resistivity silicon ($\rho_s = 20 \ \Omega\text{cm}, \ N_{AD} = 10^{15} \ \text{cm}^{-3}$) grown on a more highly doped $p^+$ silicon substrate ($N_{AD} = 10^{18} \ \text{cm}^{-3}$). An approximately 1 $\mu$m thick electrode layer overlays a 20 $\mu$m epitaxial region, ~5 $\mu$m of which are depleted and the reminder is field-free; the substrate is usually several hundred microns thick.

Such a thin epitaxial layer has very limited x-ray sensitivity (less than 4% at 10 keV), which means that the depletion region is essentially transparent to x-rays with energy above 10 keV. Improved x-ray response at higher energies can be achieved if CCDs are fabricated with higher resistivity silicon, given that the depletion depth varies as the square root of resistivity (see equation 4.12). Epitaxial devices made with very high resistivity silicon (up to 1500 $\Omega$cm, $L_d = 35 \ \mu$m) have been successfully fabricated, and their spectroscopic performance has been studied at both low (<10 keV) and intermediate (60 keV) x-ray energies (Lumb 1990, Francis et al. 1992) in the context of astronomical applications.

It is evident that in such deep depletion devices, with epitaxial regions extending up to 100 $\mu$m, carrier diffusion problems will be more severe. Charge from an x-ray absorbed in the depleted silicon is rapidly collected into the buried channel and normally contributes to the signal of a single pixel. On the other hand, signal charge generated in the undepleted region will diffuse, spreading radially to the pixels nearer to the absorption point. The potential barrier between the epitaxial $p$-type silicon and the $p^+$ substrate reflects most of the electrons (>95%) that diffuse towards it, and consequently nearly all of the charge generated in the epitaxial region is collected. Even in relatively thin epitaxial devices of the kind used in this study (EEV-02, 20 $\mu$m Epi) up to 4 pixels collect charge from a single x-ray conversion (Acton et al. 1991).

Measurements done by Francis et al. (1992) indicate that in a 100 $\mu$m Epi device with only 10 $\mu$m depletion depth the charge can spread as much as 17 pixels for a 60 keV event. The obvious implication for x-ray imaging is that charge spreading can degrade the modulation transfer characteristics (i.e. the spatial resolution) of the system.

† An epitaxy is a growth of crystals in another underlying crystal that determines their orientation.

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4.5. Scintillation x-ray detection

Scintillation detection is based on the intrinsic property that certain materials have to convert energy lost by ionising radiation into visible light. The light pulses emitted by the scintillator are then recorded by a CCD. Good x-ray detection efficiency requires materials with high density and high atomic number. It is also necessary that the scintillator is transparent to its own scintillation light, although some absorption can help to improve the spatial resolution properties of the system.

4.5.1. The scintillation mechanism

Light production in activated alkali halides (NaI, CsI) results from a complex sequence of excitation and relaxation processes. The scintillation mechanism is determined by the band structure and the energy states of the crystal lattice. The absorption of energy elevates electrons from the valence band into the conduction band. In a pure crystal the de-excitation of electrons back into the valence band is normally a very inefficient process, and the energy of the emitted photon lies outside the visible range. The probability of visible photon emission during the relaxation process can be enhanced by adding small amounts of impurities (also called activators) to the scintillator. The role of the impurity is to produce luminescence centres at intermediate energy levels between the valence and the conduction bands in the host crystal.

Figure 4.9 shows a plot of the mass attenuation coefficients of CsI between 10 and 100 keV. It can be observed that photon interactions in this energy range are completely dominated by the photoelectric effect. The attenuation coefficient has discontinuities at the iodine and caesium K-edges, which are located at 33.2 and 36 keV, respectively. Just above the K-edge of Cs the absorption efficiency of a 100 μm CsI screen can be as high as 70%. However, the fluorescent yield of the phosphor is very high (~90%) and it has been observed that for the thicknesses typically used in x-ray image intensifiers about one half of the K-fluorescent photons will escape (Rowlands and Taylor 1983). This means that there could be a significant variation in the energy deposited in the phosphor and, consequently, some degradation of the signal-to-noise ratio transfer properties of the screen. This problem will be treated in more detail in Chapter 6.
Chapter 4

The Charge Coupled Device

At x-ray energies below 100 keV photon interactions in the scintillator are completely dominated by the photoelectric effect. The photoelectrons produced during a photoelectric absorption generate secondary electron-hole pairs which diffuse to the neighbourhood of the luminescent centres and form relatively long-lived (≈ 1 μs) excited states which decay with the emission of visible light.

Table 4.1 lists the physical properties of some of the most commonly used inorganic scintillators. The average energy to produce an optical photon (ωo) varies between 25 and 45 eV. Although NaI(Tl) has the highest energy conversion efficiency, CsI(Tl) has a higher light yield due to its longer wavelength of maximum emission (λmax). The absolute scintillation efficiency (ηs) represents the fraction of energy converted into light for secondary electrons produced by x-ray interactions in the material. The third column of the table gives the refractive index of the phosphor (n) at λmax.

Table 4.1. Physical properties of the most commonly used inorganic scintillators.

<table>
<thead>
<tr>
<th></th>
<th>ρ [g/cm³]</th>
<th>λmax [nm]</th>
<th>n</th>
<th>ωo [eV]</th>
<th>ηs [%]</th>
<th>Light yield [photons/keV]</th>
</tr>
</thead>
<tbody>
<tr>
<td>NaI(Tl)</td>
<td>3.67</td>
<td>420</td>
<td>1.85</td>
<td>25</td>
<td>11.3</td>
<td>38</td>
</tr>
<tr>
<td>CsI(Na)</td>
<td>4.51</td>
<td>420</td>
<td>1.84</td>
<td>30</td>
<td>11.4</td>
<td>38-44</td>
</tr>
<tr>
<td>CsI(Tl)</td>
<td>4.51</td>
<td>550</td>
<td>1.79</td>
<td>44</td>
<td>11.9</td>
<td>52-56</td>
</tr>
</tbody>
</table>
4.5.2. CsI(Tl) light yield and spectral matching

CsI(Tl) is the scintillator with the highest light output of all known inorganic scintillators. It has a broad spectrum of emission that peaks at about 550 nm, which corresponds to the green region of the optical spectrum. In the past this was considered a disadvantage since this spectrum does not match the sensitivity of photomultipliers. However, this output is well suited to the sensitivity of silicon photodiodes and of most commercially available front illuminated CCDs, which have a peak response at around 630 nm. Figure 4.10 shows the normalised light yield of CsI(Tl) as a function of wavelength (Knoll 1989). The dashed line in the same figure represents the optical quantum efficiency of an EEV-02 front-illuminated, non anti-bloomed device, which peaks around 650 nm (see also figure 4.6).

![Figure 4.10. Normalised light yield of CsI(Tl) as a function of wavelength. The quantum efficiency of a front illuminated, non anti-bloomed device is shown for comparison (from figure 4.6).](image)

In any photo-electronic imaging device based on scintillation detection the scintillator material is used in conjunction with a photo-detector (e.g. CCD) capable of converting the luminescent information into an electrical signal. To make full use of the scintillation light the emission spectrum of the phosphor should fall near the maximum sensitivity of the photo-detector. The fraction of light that will be collected at the photo-detector would be given by

\[
\eta_{sp} = \eta_L \int_0^\lambda p_t(\lambda) \eta_\lambda \, d\lambda
\]  

(4.14)
where $\eta_L$ is the optical light collection efficiency, which accounts for losses in the coupling between the scintillator and the detector, $p_L(\lambda)$ is the probability distribution of light emission in the phosphor as a function of wavelength, and $\eta_\lambda$ is the quantum efficiency of the photo-detector at wavelength $\lambda$. The integral term in equation 4.14 represents the degree of spectral matching between the emission spectrum of the scintillator and the photo-detector response.

For the case illustrated in figure 4.10 the spectral matching is 0.35. Even assuming perfect optical coupling ($\eta_L = 1$), the fraction of light would be converted into photoelectrons is about one third of the total yield. For comparison, the spectral matching between the same CCD and a Na(Tl) screen would be only 0.06, since the spectral emission of this scintillator peaks at a shorter wavelength, where the response of the CCD is much lower.

CsI has the added advantage that it can be prepared either as a powder layer or as a vapour-deposited layer. Using the latter method packing densities near 100% can be achieved. The morphological and crystallographic characteristics of vapour-deposited layers of CsI depend very heavily on the type of substrate and its temperature at the time of deposition (Stevels and Schrama-de Pauw 1974). Screens that have a columnar structure can be grown by controlling these variables. Long and very narrow pillars with diameter ~2-5 microns grow roughly perpendicular to the substrate. The outer surface of the screen is very rough and can be considered to act as a diffuse reflector for light generated inside the screen.

![Figure 4.11. Scanning electron micrographs of a vapour deposited layer of CsI. (a) crack structure on the surface, (b) close-up of the columnar structure.](image)
Heat treatment of the screens after deposition can affect the morphology of the phosphor. Some pillars are amalgamated into slightly longer ones, and randomly distributed cracks appear on the surface (see figure 4.11). The areas between cracks are much flatter than in the original (untreated) screen. The presence of cracks and the columnar structure of the screen can help, in principle, to suppress lateral spreading of the visible light, and therefore improve the resolution properties of the screen. Resolution improvements, however, require that the cracks run perpendicular to the substrate and that the mean crack distance is as small as possible. In some systems this can be achieved by using photo-etched substrates or a special gauze.

An estimation of the fraction of light yield that would be collected in a single column of the CsI layer can be obtained by assuming an idealised crack structure. The screen can be considered to be formed by blocks of scintillator separated by very narrow cracks filled with air (see figure 4.12); the interfaces between the phosphor and air are assumed to be smooth.

The collected light yield of each column \(Y_c\) for a totally absorbing overcoat can be calculated (Bigler et al 1986) using

\[
Y_c = \frac{1}{4\pi} \int_{0}^{\omega_c} d\phi \int_{0}^{\omega_c} \sin \theta d\theta = \frac{1}{2} (1 - \cos \theta_c)
\]

(4.15)

where \(\theta_c = 90 - \theta_c\), and \(\theta_c\) is the critical angle, which is given by

\[
\theta_c = \arcsin \left( \frac{n_p}{n_o} \right)
\]

(4.16)
with \( n_p \) and \( n_o \) the refraction indices of the phosphor and air, respectively. Using \( n_p = 1.79 \) equations 4.15 and 4.16 give \( \theta_c = 34^\circ \) and \( Y_c = 22\% \). Therefore, in the ideal case the fraction of the light yield collected in a single column would vary between 22\% and 44\%, depending on the optical properties of the protecting overcoat, and for a given screen thickness an improvement in the spatial resolution of the phosphor would be expected without reducing absorption efficiency.

4.5.3. Coupling methods

One of the main problems faced in the design of a CCD-based electro-optic x-ray imaging detector is the small imaging area offered by the CCD chip itself. Size limitations require to introduce format alterations in the coupling between the x-ray converter and the CCD. Broadly speaking, there are two main coupling methods: either using electron optics (image intensifiers) or light optics (lens or fibre-optic coupling). The main design considerations required by these coupling methods have been evaluated by Deckman and Gruner (1986). In this thesis we are only concerned with light coupling methods, in particular fibre-optic coupling.

4.5.3.1. Lens coupling

Lens systems are the simplest way of coupling the light from a phosphor to the CCD. However, the demagnification ratio (object size to image size) is restricted by the inherent limitations of the light gathering efficiency of lenses. The optical coupling efficiency of a lens under Lambertian illumination (Karellas et al 1992) is given by

\[
\eta_{\text{Lens}} = \frac{T_{\text{Lens}}}{4 f_s (1 + M')^2 + 1}
\]  

(4.17)

where \( T_{\text{Lens}} \) is the transmission efficiency (typically 0.7-0.8), \( f_s \) is the f-number of the lens, defined as the ratio of the focal length to the effective diameter, and \( M' \) is the demagnification ratio. For example, for \( M' = 2 \) and \( f_s = 1.2 \) the light coupling efficiency is only \( \sim 1.3\% \). The small fraction of visible photons relayed by the lens to the detector has a very important effect in the DQE performance of the system. A second problem is that lens coupling can introduce a signal dependent background to the image, arising from light scattered in the anti-reflection coating on optical surfaces and retro-reflection by the optics of light backscattered at the CCD surface.
4.5.3.2. Fibre-optic coupling

In theory, most of the limitations found on lens coupled systems can be overcome by fibre optic coupling. The structure of an optical fibre consists of a core (the inner part of the fibre) and a surrounding material of different refractive index called the cladding. Light transmission through the fibre is confined to the core by total internal reflection at the core-cladding interface, which occurs when the cladding has a lower refractive index than the core. Figure 4.13a shows a schematic diagram of a typical cylindrical fibre. Using Snell's law and the definition of critical angle (i.e. the angle for total internal reflection) it can be demonstrated that the maximum angle of acceptance of a fibre with core and cladding refractive indexes $n_f$ and $n_c$, in contact with a medium of refractive index $n_o$, is given by

$$\sin \theta_{\text{max}} = \frac{\sqrt{n_f^2 - n_c^2}}{n_o} \quad (4.18)$$

The numerical aperture of the fibre is defined as the product $n_o \sin \theta_{\text{max}}$, therefore

$$\text{NA} = n_o \sin \theta_{\text{max}} = \sqrt{n_f^2 - n_c^2} \quad (4.19)$$

![Figure 4.13. Fibre-optic coupling. (a) Numerical aperture of a cylindrical fibre. (b) Schematic diagram of a fibre-optic taper bundle.](image)

The concept of numerical aperture is very important since the square of this quantity is a measure of the light collection efficiency of a single fibre. Since $\sin \theta_{\text{max}}$ cannot exceed the value of 1, the maximum numerical aperture in air ($n_o = 1$) is 1, and...
\[ \theta_{\text{max}} = 90^\circ. \] In this case all light reaching the input face of the fibre is totally internally reflected.

An optical fibre taper (see figure 4.13b) is a spatially coherent fibre optic bundle made from conical fibres. The term coherent is used to specify that the fibres are arranged so that their terminations occupy the same relative positions in both ends of the bundle; as a consequence such kind of arrangement is able to transmit images. A property of conical fibres is that the numerical aperture at the larger end, \( NA_l \), is always smaller than the numerical aperture at the smaller end, \( NA_r \). Therefore, the effective numerical aperture of the fibre optic taper used in the demagnifying orientation is given by

\[
NA_t = \frac{d_e}{d_l} NA_r = \frac{d_e}{d_l} \sqrt{n_r^2 - n_c^2}
\]  

(4.20)

The end face of a fibre taper consists of a group of individual optical fibres closely packed together. Of all the light incident on the surface of the taper only the light that falls onto the taper cores is transmitted. The core packing fraction is defined as the fraction of the cross sectional area of the taper which is occupied by the fibre cores and is given by

\[
\eta_p = k_p \left( \frac{d_f}{D_f} \right)^2
\]  

(4.21)

where \( d_f \) is the core diameter, \( D_f \) is the total fibre diameter (including any coating), and \( k_p \) is a constant that depends on the packing method (see figure 4.14). For square-packed fibres \( k_p = 0.785 \), while for hexagonally packed fibres \( k_p = 0.907 \). Typical core packing ratios range between 0.5 and 0.8.

The total coupling efficiency of a fibre optic taper depends on its effective numerical aperture, the core packing fraction and any other losses that can occur while the light is transmitted through the fibre, such as reflection losses at interfaces and absorption on the core material; typical transmission efficiencies (\( \eta_T \)) are of the order of 0.9. Combining equations 4.21 and 4.22, the total optical energy coupling efficiency of a fibre optic taper under Lambertian illumination (Liu et al. 1993) is given by

\[
\eta_{\text{total}} = NA_t^2 \eta_p \eta_T = \left( \frac{d_e}{d_l} \right)^2 \left( n_r^2 - n_c^2 \right) \eta_p \eta_T
\]  

(4.22)
For example, a $2\times$ reducing fibre optic taper with a packing fraction $\eta_P = 0.8$, transmission efficiency $\eta_T = 0.9$ and a principal numerical aperture $NA_T = 1$ gives an optical coupling efficiency of $\sim 18\%$. Therefore, fibre-optic format alterations are a much more efficient coupling method when compared to a lens system that provides the same demagnification ratio (see section 4.5.5.1).

\[
D_i = \frac{D_f}{\eta_P} \quad \eta_T = 0.9 \quad NA_T = 1 \quad \text{gives an optical coupling efficiency of } \sim 18\%.
\]

Therefore, fibre-optic format alterations are a much more efficient coupling method when compared to a lens system that provides the same demagnification ratio (see section 4.5.5.1).

![Figure 4.14. Packing methods in typical fibre-optic bundles.](image)

A source of contrast degradation in fibre-optically coupled devices is the presence of stray light propagating within the taper. The reduction in image contrast when stray light is present is proportional to the square of the effective numerical aperture and to the packing fraction of the taper (Liu et al 1993). If $C$ represents the original contrast, then the contrast at the output of the taper would be

\[
C_i = NA_T^2 \eta_P C
\]

Since the principal numerical aperture ($NA_T$) is normally designed to be unity in modern fibre-optic tapers, it can be seen from equation 4.22 that contrast degradation is inversely proportional to the square of the taper ratio.

The most common technique used to reduce the effect of stray light is to include a highly absorbing material between the individual fibres, which is called extra mural absorber (EMA). However, a compromise must be reached between the thickness of the EMA, which reduces the effective packing fraction of the taper, and the degree of absorption required.
4.6. Radiation damage

4.6.1. Radiation damage to the CCD

A significant amount of radiation can permanently damage the performance of a CCD. Different performance parameters are affected in different ways by various types of ionising radiation, but in general terms it has been observed that thermal carrier generation and readout noise increase, while charge transfer efficiency decreases. There are essentially two types of radiation damage, those related to ionisation and those caused by displacement in the crystalline lattice of the silicon. The latter is only related to high energy ionising particles. This means that the maximum dose of a particular kind of radiation that can be tolerated by a CCD will depend on nature of the application. As a general guidance for the case of ionising radiation it has been observed that if the dose in the silicon is less that \(-10^2\) Gy the performance parameters remain practically unchanged; at \(-10^5\) Gy a significant change can be observed and finally above \(-10^6\) Gy there is serious degradation of performance (EEV 1987).

Degradation of performance normally occurs slowly as the dose builds up and very rarely the effects are catastrophic. For example, studies carried out by EEV indicate that dark current at room temperature increases at a rate of approximately 1 nA/cm² per \(10^2\) Gy. In comparison, a typical peak signal current level is of the order of 50 nA/cm². Some of the damage can be repaired and the original characteristics partially restored using annealing procedures. However, radiation hardness is normally reduced after annealing.

4.6.2. Radiation damage to the phosphor coating

In general terms doped alkali halides such as CsI(Tl) are also susceptible to radiation damage. Prolonged exposure or intense radiation can change their scintillation characteristics. The damage normally manifests itself as a gradual decrease of the relative signal size. Measurements of light output in evaporated CsI(Tl) layers of the type used in image intensifier tubes indicate a reduction of approximately 30% in signal size after an accumulated dose of \(-10^3\) Gy (Perez-Mendez 1990). Although some radiation damage can be noticeable above 10 Gy, most of the degradation occurs on the optical absorption bands at low wavelengths and therefore it does not produce a very noticeable effect when used with photodiodes or CCDs.
This chapter describes the implementation and calibration of an x-ray imaging system prototype based on charge-coupled devices. The design of the prototype was based in a simple architecture in order to be able to test several design strategies. As such, the system was not optimised for a particular clinical imaging task, but was used only to lay the foundations upon which a clinical system could be designed.

5.1. Description of the system

5.1.1. The CCD camera

The x-ray imaging system used in this work was based on a specially modified asynchronous CCD camera (EEV-Photon P46580) that included a lead-shielded case and accepted two different sensor formats. The camera provided a 50 Hz monochrome standard video output signal, and could be operated either in standard video mode or in asynchronous strobe mode. When the camera operates in standard mode it acquires images continuously, at a rate of 25 frames per second (see figure 5.1). One video frame is formed by two interlaced fields, each one of 20 ms duration, that is, the information contained in a single image corresponds to an integration time of 20 ms.

![Timing diagram for the standard video mode of the CCD camera.](image-url)
The asynchronous strobe mode refers to the combination of an electronic shuttering method with event related triggering to start image acquisition. When the camera operates in this mode the image section of the CCD is continuously reverse clocked into a diode drain situated at the top of the sensor until a trigger pulse, called the *strobe pulse*, is received (see figure 5.2). Following the final clocking sweep the camera integrates for as long as the strobe pulse is low.

![Timing diagram for the asynchronous mode of the video camera.](image)

Figure 5.2. Timing diagram for the asynchronous mode of the video camera. The integration time is controlled by the duration of the strobe pulse, and image capture on the frame grabber is triggered by the framestore pulse.

At the end of the integration period the acquired image is transferred rapidly into the store section of the CCD and the image section returns to reverse clocking mode. The image is stored until the beginning of the next TV field, during which it is read out in the normal way. A TTL pulse is provided by the camera (the framestore trigger pulse) just prior to the readout of the field in which the information is contained. This signal was used to trigger image digitisation on a frame grabber. A characteristic of the asynchronous mode is that image information is contained in a single field, but the camera still produces an interlaced frame in order to remain synchronous with the external equipment. Therefore, the digitised image contains a blank interleaved field. This redundant information was removed after image capture was completed using a post-processing program.
The integration time was controlled using an external electronic circuit based on a monostable-multivibrator and a Wavetek-185 wave generator. This circuit was used to generate a single TTL pulse of variable width that could be triggered manually to start image acquisition. The width of the strobe pulse was monitored using a Gould-OS4040 digital-storage oscilloscope. The circuit was designed to produce strobe pulses that could be adjusted from 10 ms up to 0.5 s duration, which covered the range of exposures necessary for this investigation. Figure 5.3 shows a calibration plot of the measured integration time as a function of the frequency settings in the sweep generator. It can be seen that the circuit was very stable over the whole range of frequencies used.

![Figure 5.3. Measured integration time (strobe pulse width) as a function of the frequency settings in the wave generator. The dotted line is a linear regression fit of the log-log data.](image)

The width of the strobe pulse and, therefore, the image integration time on the camera, were determined from linear regression analysis of the log-log data shown in figure 5.3, and it was found to be $t_i \text{[ms]} = 600 / (f_{\text{sweep}} \text{[Hz]})$. This equation was used to set the integration time simply by adjusting the frequency settings of the sweep generator when the camera was operated in asynchronous mode.

### 5.1.2. Sensors

All the CCDs used in this study were EEV 3-phase, front-illuminated frame transfer devices. Two different modes of x-ray detection were employed: direct x-ray
conversion in the silicon using deep depletion devices, and scintillation detection using CsI(Tl) screens fibre-optically coupled to the CCDs. The main characteristics of these devices are listed in table 5.1. The deep depletion devices were made of high resistivity silicon (100 Ω cm) and had epitaxial layers up to 100 μm thick. The CCDs used for scintillation detection were 20 μm epitaxial devices made of low resistivity silicon.

Table 5.1. Main characteristics of the two types of CCDs used in this study.

<table>
<thead>
<tr>
<th></th>
<th>Coated</th>
<th>Deep depletion</th>
</tr>
</thead>
<tbody>
<tr>
<td>Type (EEV)</td>
<td>CCD02-06</td>
<td>CCD03-06</td>
</tr>
<tr>
<td>Format</td>
<td>2/3&quot;</td>
<td>1/2&quot;</td>
</tr>
<tr>
<td>Image area (mm)</td>
<td>8.5×6.3</td>
<td>6.4×4.8</td>
</tr>
<tr>
<td>Size of array (pixels)</td>
<td>384×288</td>
<td>384×288</td>
</tr>
<tr>
<td>Pixel size (μm)</td>
<td>22×22</td>
<td>16.5×16.5</td>
</tr>
<tr>
<td>Coating</td>
<td>70-130 μm CsI(Tl)</td>
<td>N/A</td>
</tr>
<tr>
<td>Epitaxial depth (μm)</td>
<td>20</td>
<td>up to 100</td>
</tr>
<tr>
<td>Resistivity (Ω cm)</td>
<td>20</td>
<td>100</td>
</tr>
</tbody>
</table>

A total of 9 different CCDs were used during the course of the evaluation. A complete list of all these devices is given in table 5.2. Two non-x-ray sensitive devices (No-1/2) were used for the optical calibration of the system. Of the others, four were coated devices (FO-1/2, FT-1/2) and three were deep depletion devices (DD-1/3). The values of the epitaxial layer, resistivity and phosphor thickness were provided by the manufacturer, and are accurate to within ±15% (Flint 1992b).

Table 5.2. List of all the sensors used for the evaluation.

<table>
<thead>
<tr>
<th>Device</th>
<th>Number</th>
<th>Type</th>
<th>Epi [μm]</th>
<th>ρ [Ω cm]</th>
<th>Notes</th>
</tr>
</thead>
<tbody>
<tr>
<td>No-1</td>
<td>9172/3/59</td>
<td>02</td>
<td>20</td>
<td>20</td>
<td>Non anti-bloomed (NAB)</td>
</tr>
<tr>
<td>No-2</td>
<td>9012/6/48</td>
<td>02</td>
<td>20</td>
<td>20</td>
<td>Anti-bloomed (AB)</td>
</tr>
<tr>
<td>FO-1</td>
<td>9392/7/50</td>
<td>02</td>
<td>20</td>
<td>20</td>
<td>NAB FO 130 μm CsI</td>
</tr>
<tr>
<td>FO-2</td>
<td>A0375/24</td>
<td>02</td>
<td>20</td>
<td>20</td>
<td>NAB FO 130 μm CsI</td>
</tr>
<tr>
<td>FT-1</td>
<td>A037/47</td>
<td>02</td>
<td>20</td>
<td>20</td>
<td>AB FT 70 μm CsI</td>
</tr>
<tr>
<td>FT-2</td>
<td>A037/47</td>
<td>02</td>
<td>20</td>
<td>20</td>
<td>AB FT 100 μm CsI</td>
</tr>
<tr>
<td>DD-1</td>
<td>A0425/13</td>
<td>03</td>
<td>100</td>
<td>100</td>
<td>NAB Deep depletion (DD)</td>
</tr>
<tr>
<td>DD-2</td>
<td>A0425/9</td>
<td>03</td>
<td>100</td>
<td>100</td>
<td>NAB DD</td>
</tr>
<tr>
<td>DD-3</td>
<td>7233/4/5</td>
<td>03</td>
<td>70</td>
<td>100</td>
<td>NAB DD</td>
</tr>
</tbody>
</table>

FO: Fibre-optic faceplate coupling. FT: Fibre-optic taper coupling
Devices FT-1 and FT-2 in table 5.2 refer to the same CCD chip, which was fitted with a 1.6× fibre-optic reducer taper and could accept two different CsI screens. Figure 5.4 shows a schematic diagram of the tapered device. The CsI(Tl) screens were grown on separate fibre-optic faceplates that could be optically coupled to the taper using an index-matching oil.

![Figure 5.4. Schematic diagram of the fibre-optic taper coupled device. Two replaceable CsI screens mounted on fibre-optic faceplates could be used with a single CCD.](image)

All the fibre-optic couplers (faceplates and tapers) were made of square-packed fibres and included extra mural absorber to reduce the effects of stray light. However, no data was available about the packing fractions of the couplers. All the faceplates had a numerical aperture of 1, and the effective numerical aperture of the taper was 0.625.

### 5.1.3. The video-frame grabber

Images produced by the Photon camera were digitised using a Data Translation DT-3852 programmable video frame grabber board. The frame grabber consists of a plug-in card connected to a PC compatible computer. The board can operate either in continuous or in single-frame grabbing mode and it is fitted with 2 Mb of on-board video memory.

The function of the frame grabber was to strip the video signal of the CCD camera from all the sync pulses that carry the video information and to digitise that part of the signal that contains image data. The DT-3852 includes some programmable circuitry that could be used to adjust the sensitivity and range of digitisation according to the characteristics of the input signal. Three parameters were used to set the maximum
attainable resolution of the ADC for a given low-level input: the frame grabber gain, the signal zero-offset and the digitisation range of the analogue-to-digital converter.

The programs to control the board were written in C, and include routines for initialisation and setup of the input circuits, trigger detection, image acquisition and on-board buffer manipulation. All the ADC and video-acquisition setup parameters were controlled from within the program.

5.1.3.1. Video input clock and format memory

Video signals are complex waveforms that carry information about the brightness at each point in the image, along with timing pulses that indicate the end of each line (horizontal sync) and of each display frame (vertical sync). Accurate digitisation of the video signal requires the specification of several parameters in the ADC section of the frame grabber that are used to describe the timing format of the video waveform.

![Schematic diagram of the video output signal of the Photon camera.](image)

**Figure 5.5.** Schematic diagram of the video output signal of the Photon camera. The duration of the different pulses is given in microseconds, and the maximum amplitude (0.68 volts) corresponds to saturation.

The timing format of the video signal was measured with an oscilloscope and compared to the nominal values specified by the manufacturer. Figure 5.5 shows a schematic diagram of a typical video output waveform for a single line, and the corresponding values obtained from the measurements, which agree to within 1% with the nominal values.
The format memory provides control signals for the video-input circuitry. Four parameters are used to specify the digitisation window in the horizontal and vertical directions, and two parameters are used to designate the position within the horizontal blanking period (the sync and back-porch) where the AD converter is zeroed and clamped to a reference level.

The digitisation rate of the board is controlled by a video-input clock. The analogue-to-digital conversion is carried out by an 8-bit flash ADC which has a maximum sampling frequency of 18 MHz. The correct value of this parameter is set in terms of the timing format of a horizontal line of the video signal and the number of pixels per line in the sensor. The Photon camera operates at a refresh rate of 50 Hz and produces 625 lines per interlaced frame. The interlaced mode of operation effectively doubles the number of CCD pixels (384x288) in the final image, therefore each interlaced frame in the video display was made of 768 pixels per line and 576 active lines per frame. From this value and the length of the active video per line shown in figure 5.5 the digitisation rate (video input clock) was calculated as

\[ f_{\text{sampl.}} = \frac{768 \text{ pixels/line}}{52 \text{ \mu s/line}} = 14.77 \text{ MHz} \quad (5.1) \]

This value differs slightly from the video output clock for a typical 50 Hz signal, which establish 15 MHz as the normal pixel readout rate. An experimental validation of the digitisation parameters used with the frame grabber will be presented in §5.2.2.

5.1.3.2. Frame grabber gain, zero offset and AD-reference

The Photon CCD camera provides a fixed-gain video output signal in the range 0 to 0.68 volts above the blanking level (see figure 5.5). When the system was operating with low-level inputs, all the signal amplification had to take place in the frame grabber board. The frame grabber gain \( G_y \) acts as a multiplier for the incoming signal, and it can take the values of 0.5, 1, 2 or 4. Gain factors greater than 1 increase the overall amplitude of the signal, causing the image to become brighter.

The function of the zero offset value is to shift the input voltage to a given level. The effect of this shift is to determine the signal level below which all the data is digitised to a value of zero. The zero offset value shifts the signal with respect to a fixed AD-reference point, and is controlled by a 12 bit DAC. The nominal zero offset for a 50
Hz video signal corresponds to the blanking level. Therefore, at low illumination levels this parameter remains the same, in order to avoid possible signal loss.

The AD-reference value ($AD_{ref}$) is used to set the upper threshold for digitisation in the ADC and, along with the zero offset, defines the input range. This parameter is controlled by a 6 bit DAC which provides 64 possible values for $AD_{ref}$ representing the range from 0 to 1.2 volts in 19 mV steps. The lower limit for this parameter is determined by the precision of the ADC, and the upper limit is given by the maximum output voltage of the camera. The Photon camera has a nominal saturation value of 680 mV, therefore the maximum $AD_{ref}$ was chosen as 38 to cover the whole range of output. The minimum $AD_{ref}$ was set to 10 since lower values could degrade ADC accuracy (see §5.2.3.2).

### 5.1.4. ADC calibration considerations

Since the precision of the ADC was limited to 8 bits, for a given level of illumination the optimal parameters for digitisation had to be found. After the nominal values were set an image was acquired and evaluated. Typically gross adjustments were first made to the frame grabber gain. Fine adjustments were then made using the AD-reference. Adjustments to the AD-reference value change the upper threshold (white level) of digitisation. This adjustment is logically different from the zero offset and gain adjustments in that both zero offset and gain affect the input signal directly, while the AD-reference affects the analogue-to-digital converter.

In the remaining of this chapter the concepts of "dark" and "flat" images will be frequently used. A dark image is obtained by shielding the image section of the CCD in such a way that there is no optical nor x-ray input signal. A dark image contains only the background composed of thermally generated carriers and any offset introduced by the camera electronics. A flat image is obtained using constant, uniform illumination across the entire surface of the detector. As such, it conveys no picture information apart from any structure associated with the phosphor screens or with the CCD itself.

### 5.2. Radiometric measurements

A complete characterisation of the conversion processes leading to a given output by the CCD system requires the determination of the conversion constant $G_e$ that relates
the number of electrons per pixel (the CCD signal) to the digital number (in analogue to digital units, ADU) produced by the frame grabber. The two most common methods used to determine this constant are either radiometric techniques using visible light, or x-ray methods based on monoenergetic x-ray sources.

In radiometric methods the incident irradiance from a uniform, ideally monochromatic source is measured using a power meter, and then either the output current is measured, or tabulated values of responsivity are used to calculate the required constant. Monoenergetic x-ray sources can also be used since it is known that the energy required to produce an electron-hole pair in silicon is a well defined constant (3.65 eV per e-h pair). However, the use of monoenergetic x-ray sources requires to use very active sources and/or to acquire many frames to obtain meaningful statistics, since the conversion efficiency, even for deep depletion devices (100 μm epitaxial layer) is only of the order of 10% at 20 keV and diffusion of the charge generated in the undepleted silicon also introduces errors.

5.2.1. Materials and methods

An integrating sphere coupled by an optical fibre to a HeNe laser was used as a diffuse light source for the radiometric measurements. The inner surface of an integrating sphere is uniformly coated with a layer of material that has a high diffuse reflectance. In this case the sphere provides a volume for collecting optical radiation and redirect it to a given target area. When light from a source enters the integrating sphere it looses all its direction and polarization information. The light intensity at the exit port of the sphere is uniform and Lambertian, that is, the radiant flux coming out from the sphere is the same in all directions. The integrating sphere used for this project (see figure 5.6) was developed in-house (Essenpreis 1993). It has an internal diameter of 48 mm, input port diameter of 1 mm and an output port diameter of 5 mm.

A special port adapter was made in order to connect the integrating sphere to the Photon camera. The adapter was also used to hold a set of optical filters between the output port of the sphere and the CCD. A 400 μm diameter silica optical fibre was used to collect and guide the light from a 1 mW HeNe laser (λ = 630 nm) into the input port of the sphere. The system (see figure 5.6) was mounted on an optical bench in order to facilitate the alignment of the optical fibre and the laser.

† Energy flux per unit projected area per unit solid angle
The output irradiance from the sphere was measured with a Coherent-212 power meter. This power meter uses a silicon cell detector with a spectral sensitivity from 440 nm to 1.1 µm, and it has an active sensor area of 49 mm². The power meter could operate in the range 0.1 µW to 0.1 W, with an overall accuracy of ±5%. A maximum output power density of approximately 40 nW/mm² was obtained under typical experimental conditions. This level of illumination was enough to saturate a normal (EEV-02) non anti-bloomed device which has a nominal full well capacity of $3 \times 10^5$ electrons per pixel.

![Schematic diagram of the integrating sphere and of the experimental setup used for the radiometric measurements.](image)

**Figure 5.6.** Schematic diagram of the integrating sphere and of the experimental setup used for the radiometric measurements.

A set of optical filters was made from a piece of uniformly exposed and developed radiographic emulsion with an optical density of 0.04±0.01. The incident irradiance to the CCD was varied by placing different numbers of these filters on the sphere port-adapter. Figure 5.7 shows a semi-log plot of the measured output power as a function of the number of filters used to attenuate the diffuse light distribution. It can be observed that the reduction of intensity was linear, which means that several filters could be used to attenuate the output power by known amounts.
5.2.2. Experimental validation of the digitisation parameters

5.2.2.1. Non-uniformity of the video signal

In order to verify the values for the digitisation parameters three cameras, the Photon EEV camera and two other normal video cameras, all conforming to the CCIR standard and from different manufacturers, were compared. The video signals were first measured using an oscilloscope and subsequently a series of images were taken. All AD conversion parameters apart from the digitisation rate and format memory were kept to their default values ($AD_{ref} = 63, \ G_f = 1$). At low levels of illumination the EEV camera was observed to give non-uniform output as opposed to the other two cameras, which showed good uniformity.

Figure 5.8 shows a plot of the horizontal profiles corresponding to zero input (dark) images for the three cameras tested. An arbitrary offset was added to the data to avoid overlapping between the profiles. Since all the images were taken without illuminating the sensor and all cameras produce the same (standard) output, the data indicated that the non-uniformity observed in the EEV camera was due to the readout electronics, and was not an effect of the digitisation procedure. Consequently any quantitative measurements done with this camera had to take into account any non-uniformity effects.
5.2.2.2. Analysis of non-uniformities by Fourier methods

The presence of spatial non-uniformities in the image data was investigated using Fourier techniques. It is well known that some artefacts present in the digitised image can occur due to the interaction between the readout electronics of the camera and the signal acquisition electronics (Boreman 1987).

The conversion of continuous scene data into an array of digitised values involves a complex sequence of sample-and-hold operations. Differences between the pixel readout clock and the ADC sampling rate can give rise to spurious frequency content on the digitised image. Figure 5.9a shows an image of a uniformly illuminated field that was taken using a sampling rate of 16 MHz. A beating frequency arising from the asynchronicity between the pixel readout clock and the input clock that controls the digitisation process can be clearly seen. It is interesting to note that non-uniformities are present only in the raster (horizontal) direction, while the image has good uniformity in the vertical direction.

The information given by the spatial frequency content of the digitised images can be used to determine the correct value for the sampling frequency of the ADC without knowing the exact time format of the video waveform. A series of images was taken under the same uniform illumination conditions using different sampling frequencies.
Figure 5.9b shows the corresponding horizontal profiles for images taken at 15, 16, 17 and 18 MHz. An arbitrary offset of 20, 40 and 60 ADU was added to the profiles taken at 16, 17 and 18 MHz in order to avoid overlapping. It can be seen that an increase on the digitisation frequency also increases the frequency of the beats.

Estimates of the spatial-frequency content of the video signal in the raster direction were obtained by taking the Discrete Fourier Transform (DFT) of the profiles shown in figure 5.9b. The results of the calculation are shown in figure 5.10a, where the modulus squared of the Fourier transforms are plotted as a function of spatial frequency for all four image profiles. The spikes in the spectra correspond to the harmonic component of the beating frequency present at different sampling rates, and are indicated with an arrow for each of the four different sampling frequencies used.

Figure 5.10b shows a plot of the spatial frequency of the beats, determined from the power spectrum estimates, as a function of the frame grabber sampling frequency. Linear regression analysis of the data gives

\[
    f_{\text{beat}}[\text{mm}^{-1}] = 2.38 \cdot f_{\text{sampl}}[\text{MHz}] - 35.06
\]  

The correct sampling rate is given by the frequency for which \( f_{\text{beat}} = 0 \), that is, by the interception of the regression line with the sampling frequency axis. Therefore, using the result from the linear fit given by equation 5.2 the correct sampling frequency is 14.73 MHz. This value agrees to within 0.3% with the sampling frequency calculated
in §5.1.3. It was observed that variations of ±0.1 MHz around this value had no effect in the output signal. Furthermore, it was noticed that dark signal and background non-uniformity depend on the particular CCD being used.

These results indicate that non-uniformities in the raster direction are related to the video circuitry of the camera, and that they are not artefacts introduced by the digitisation process. Although it is possible, in principle, to adjust the camera electronics to minimise these effects, a calibration procedure of this kind requires specialised equipment. Besides, the nature of this assessment in which several CCDs were being tested using a single camera made this procedure impracticable.

5.2.3. Frame grabber linearity and ADC resolution

5.2.3.1. Gain and AD-reference linearity

The experimental setup described in §5.2 was used to investigate the effects that changes in the frame grabber gain and/or AD-reference would have on the measured signal. For the gain linearity measurements the AD-reference value was set to 63, which corresponds to the default setup configuration after frame grabber initialisation, and the gain was set to its minimum value (0.5). Starting from an illumination level sufficient to saturate the CCD the output power was reduced until the measured signal
reached a level of \(-25\) ADU, which is about ten times smaller than the maximum output value of the 8-bit ADC.

Four images were taken using all the possible gain settings in the frame grabber (0.5, 1, 2 and 4) without changing any other parameters, or the illumination conditions. Figure 5.11a shows the values of the measured signal, normalised at unit gain, as a function of the gain settings. It can be seen that the gain amplifier of the frame grabber had very good linearity, to within \(2\%\) of the expected linear relationship, which is plotted with a dashed line in the same graph.

The precision of the AD-reference value was determined using the same illumination conditions as in the previous section. First, the peak output voltage of the camera was measured and found to be \(V_{\text{out}} = 0.22 \pm 0.04\) V. Since each \(AD_{\text{ref}}\) step is equivalent to 19 mV then the optimum range of digitisation would correspond to an AD-reference value of \((0.22\text{ V} / 0.019\text{ V}) = 12\). After setting the gain to 1, a series of ten images were taken changing \(AD_{\text{ref}}\) from 63 down to 12 in small steps. Figure 5.11b shows the results of the AD-reference linearity measurements, normalised with respect to \(AD_{\text{ref}} = 63\). The dotted line is a linear regression of the experimental data and the dashed line represents the expected values, which should be proportional to the inverse of \(AD_{\text{ref}}\). The deviation from the expected values does not exceed 15% down to \(AD_{\text{ref}} \approx 10\), therefore, this value was chosen as the lower limit for this parameter.
5.2.3.2. Effective ADC resolution

Changes in exposure conditions require to readjust the digitisation parameters in order to match the output signal of the video camera to the full 8-bit precision of the ADC. As a consequence, direct comparison of signals measured under different exposure conditions can only be made if the measured values are renormalised to compensate for changes in the gain and AD-reference. Taking into account the deviation from the expected values observed in the figure 5.11b the renormalised signal is given by

\[ S[\text{ADU}] = \left( \frac{AD_{\text{ref}}}{38} \right)^{0.9} \frac{S(G_f, AD_{\text{ref}})}{G_{\text{ref}}} \]  

(5.3)

The effective ADC resolution of the frame grabber is therefore determined by the combined effect of the gain amplifier and the AD-reference DAC. For low level input signals the maximum attainable gain is given approximately by

\[ \frac{AD_{\text{max}}}{AD_{\text{min}}} \times G_f = \frac{38}{10} \times 4 = 15.2 \]  

(5.4)

In the rest of this thesis the term ADU applies to signals renormalised with respect to \( G_f = 1 \) and \( AD_{\text{ref}} = 38 \) using equation 5.4, while ADU' applies to signals measured using the maximum attainable gain, that is 1 ADU = 15.2 ADU'.

5.2.4. Optical calibration – coated devices

The conversion factor \( G_s \) for the fibre-optically coupled devices was determined by measuring the response of the CCD to optical light using devices No-1 and No-2, and the experimental setup described in §5.2.1. The \( AD_{\text{ref}} \) value was set to 38 to cover the range from 0 up to 0.68 Volts, and the frame grabber gain was set to 1. The incident irradiance was varied in small steps from ~1 nW/mm² up to complete saturation of the CCD, which occurred at about 35 nW/mm². For a given incident irradiance the corresponding output voltage and mean output signal from the ADC were measured. Values of the mean and standard deviation of the output signal were determined over a circular region of interest (ROI) of radius 60 pixels in the centre of the image.

Figure 5.12 shows the measured output signal as a function of the incident irradiance for device No-1. The dark signal for this device measured over the same ROI was
20.7 ± 0.9 ADU. The uncertainty of the measurement is approximately of the order of the symbol size. The signal is observed to be linear up to ~11 nW/mm\(^2\) where the device begins to saturate. The slope of the data below saturation corresponds to a conversion gain of 13.40 ± 0.36 ADU/(nW/mm\(^2\)).

![Figure 5.12](image)

**Figure 5.12.** Mean output signal as a function of the incident irradiance for device No-1. The dotted line is the linear regression of the data excluding the last two points. The dashed line was calculated with the saturation model that will be described in §5.2.5.

Using equation 4.5 the number of charge carriers generated in the CCD per unit incident irradiance can be expressed as

\[
\frac{N_e}{W_\lambda} = \frac{R_\lambda A_{\text{pix}} t_f}{q_e}
\]  

(5.5)

In this case \(A_{\text{pix}} = 4.84 \times 10^{-4} \text{ mm}^2\), \(t_f = 20 \text{ ms}\) and \(R_\lambda = 230 \text{ mA/W}\) at \(\lambda = 630 \text{ nm}\), thus \(N_e/W_\lambda = 1.39 \times 10^4 \text{ e}^-/(\text{nW/mm} ^2\)\). The overall conversion factor for the coated devices based on EEV-02 non anti-bloomed CCDs is therefore

\[
G_e = \frac{13.4}{1.39 \times 10^4} \text{ ADU/e}^- = 9.64 \times 10^{-4} \text{ ADU/e}^-
\]  

(5.6)

that is, approximately \(10^3\) electrons are needed to generate an output of 1 ADU. Using the maximum frame grabber gain \(G_f = 4\), \(AD_{\text{ref}} = 10\) the conversion factor would be increased up to \(1.47 \times 10^{-2} \text{ ADU'}/\text{e}^-\), which is equivalent to ~68 e^-/ADU'.
The conversion factor for an anti-bloomed device (No-2) was determined using the same method as detailed above. The results of the calibration are shown in table 5.3, along with the values for device No-1. The values of the full-well capacity were calculated using a saturation value of 250 ADU in both cases. These figures are in agreement with the nominal full-well capacities quoted by the manufacturer, which are $3 \times 10^5$ and $1 \times 10^5 \pm 15\%$ for the NAB and AB devices, respectively (Flint 1992b).

### Table 5.3. Conversion factors relating AD units to the number of charge carriers per pixel for EEV-02 non anti-bloomed and anti-bloomed devices.

<table>
<thead>
<tr>
<th>Device</th>
<th>Type</th>
<th>$R_s$ [mA/W]</th>
<th>$G_s$ [ADU/e]</th>
<th>Full-well capacity [e]</th>
</tr>
</thead>
<tbody>
<tr>
<td>No-1</td>
<td>EEV-02 NAB</td>
<td>230</td>
<td>$9.64 \times 10^{-4}$</td>
<td>$2.6 \times 10^5$</td>
</tr>
<tr>
<td>No-2</td>
<td>EEV-02 AB</td>
<td>135</td>
<td>$2.65 \times 10^{-3}$</td>
<td>$9.4 \times 10^4$</td>
</tr>
</tbody>
</table>

* for $\lambda=630$ nm.

#### 5.2.5. Saturation effects

The gradual departure of linearity of the measured output signal when the CCD approaches saturation can be explained in terms of the statistical properties of the incident photon flux, using a statistical model that describes the input-output relationships of a semi-ideal array of photon counters (Dainty and Shaw 1974). In this model the system is assumed to be linear and to have unit gain. It is also assumed that each pixel registers the incident quanta with perfect efficiency and independently of its neighbours. After a given integration time each pixel reaches an image state that depends only on the number of incident quanta. Each receptor is restricted to have a finite number of discrete image states, such that any further incident quanta do not contribute to the signal once the pixel is saturated.

It has already been seen in Chapter 1 that the incident photon flux can be described by a Poisson probability distribution. Then, for an illumination level such that the average number of incident quanta is $\overline{q}$, the mean output signal can be expressed as the sum of the products of each value with its probability of occurrence using the definition of the mean value of a probability distribution

$$
\langle S \rangle = \sum_{r=1}^{S_{\text{max}}} \overline{q}^r \frac{e^{-\overline{q}}}{r!} + \sum_{r=S_{\text{max}}}^{\infty} \overline{q}^r \frac{e^{-\overline{q}}}{r!}
$$

(5.7)
In this equation the first sum represents the contribution from those pixels that have received $r$ quanta, and the second sum corresponds to the pixels that received a number of counts equal to the saturation level $S_{\text{max}}$. Equation 5.7 can be rearranged such that the mean signal can be expressed as

$$\langle S \rangle = S_{\text{max}} \left[ 1 - \left( g(q, S_{\text{max}}) \cdot e^{-\bar{q}} \right) \right]$$  \hspace{1cm} (5.8)

where

$$g(q, S_{\text{max}}) = \frac{1}{S_{\text{max}}} \left[ 1 + \sum_{n=1}^{S_{\text{max}}} \left( \sum_{r=0}^{n} \frac{q^r}{r!} \right) \right]$$  \hspace{1cm} (5.9)

A short routine was written using Mathematica (Wolfram 1991) to evaluate equations 5.8 and 5.9 using as a saturation level $S_{\text{max}} = 100$. The output of the program was then renormalised to match the gain, offset and saturation level of the experimental data obtained in the previous section. The dashed line in figure 5.12 shows the results of the calculation. It can be seen that the calculated values correspond very well with the experimental data. These results could have important implications in the appropriate selection of the ADC resolution for a CCD with a given full well capacity in an optimised system. Even if the system is quantum limited the linearity and dynamic range are affected by the gradual saturation of the CCD.

5.3. X-ray measurements

5.3.1. Exposure techniques

The source for all the x-ray measurements was a Machlett x-ray tube controlled by an AED D-44 generator. A series of quality assurance tests were made to determine the main characteristics of the x-ray tube, such as output linearity and stability, peak kilovoltage, half value layer (HVL) and focal spot size. Air kerma measurements were made with a Keithley 35050A dosimeter fitted with a 15 cc ionization chamber, which has a nominal overall accuracy of $\pm 2\%$.

The output of the x-ray tube was observed to be linear with exposure time. After an adequate warming up period the output consistency of the tube under fixed exposure conditions was of the order of 8%. The half value layer was determined using a Keithley 115A aluminium HVL set and a 70 kVp beam, and it was found to be 2.1 mm Al. This corresponds to a total filtration of 2.0 mm Al for an x-ray tube operated
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A CCD-based x-ray imaging system

at 70 kVp constant potential (Birch et al. 1979). The focal spot size was determined from images of a 30 µm pinhole taken with a conventional film-screen combination. A focal spot size of 1.9 x 1.3 mm was obtained.

Two different methods were used to control the entrance exposure to the CCD. For a given beam quality and tube current the entrance exposure was varied either by operating the camera in asynchronous mode while keeping the focal spot-detector distance (FDD) fixed, or by changing the distance between the x-ray tube and the detector, while keeping a constant integration time. The variation of the mean output signal with integration time and with focal spot-detector distance is shown in figures 5.13a and 5.13b, respectively. Both data sets were taken with device FO-1 at 70 kVp. In the first case the FDD remained fixed at 80 cm, and in the second case the integration time was set to 20 ms.

![Figure 5.13. Mean carrier signal as a function of: (a) integration time for constant FDD (80 cm); (b) distance, for constant integration time (20 ms).](image)

It can be observed that in asynchronous mode the exposure range was limited due to saturation effects. The dashed line in figure 5.13a was calculated with the model described in §5.2.5. In this case the mean signal varies linearly with increasing integration time between 20 and 80 ms, which was equivalent to an entrance air kerma between 90 and 480 µGy, before the CCD began to saturate. Since the CCD was operated at room temperature the output signal includes the contribution of thermally generated carriers.
On the other hand, the experimental conditions restricted the distance between the x-ray tube and the CCD to be varied between 146 and 30 cm. This in turn corresponded to an entrance air kerma between 25 and 650 μGy, which represent an exposure range approximately five times bigger than in the previous case. Note that the data in figure 5.13b follow a perfect inverse-square behaviour, as indicated by the dotted line.

The exposure conditions used for the quality assurance tests were chosen as a reference for most of the CCD x-ray measurements. In what follows the “standard” exposure conditions will refer to a 70 kVp beam, 2 mm Al total filtration, 200 mA tube current, 51.5 cm FDD and 20 ms exposure time. Unless otherwise specified, beam quality will be quoted in terms of tube potential only, and a total filtration of 2 mm Al will be assumed.

5.3.2. Linearity and sensitivity

The sensitivity of the system with respect to x-ray exposure was measured in terms of AD conversion units per unit entrance air kerma at the input plane of the detector. A 70 kVp beam was used and the entrance exposure to the detector was varied by changing the distance between the x-ray tube and the CCD; all other system parameters remained constant. Two images were taken with each device at six different exposure levels covering the range from 100 to 450 μGy. The mean signal at each exposure was determined from the average of the two images after background (dark signal offset) subtraction.

Figure 5.14 shows the results of the measurements after data renormalisation to compensate for differences in the frame grabber gain. In all cases the response of the system was linear with x-ray exposure. The sensitivity of the system was determined from the slopes of the straight lines obtained from linear regression analysis, and found to be 0.60, 0.29, 0.19 and 0.15 ADU/μGy for FO-2, FT-2, FT-1 and DD-3 respectively. These values would correspond to a maximum sensitivity (i.e. using maximum frame grabber gain) of 9.09, 4.47, 2.87 and 2.31 ADU/μGy.

As it would be expected, the coated devices have better sensitivity than the deep depletion device. Also, the sensitivity of the coated devices increases with phosphor thickness. However, it is interesting to note that the sensitivity of the coated devices depends as well on the coupling efficiency of the intervening fibre-optics. For example, the quantum efficiency of a 130 μm and a 100 μm CsI layers averaged over
a 70 kVp spectrum are 56% and 47%, respectively; the ratio of the quantum efficiencies is therefore ~1.19. On the other hand the sensitivity of FO-2 is about twice than that of device FT-2. This problem will be treated in more detail in Chapter 6.

The sensitivity values found for this system compare well with those reported recently by Roehrig et al (1993) for three commercial digital x-ray camera-prototypes developed for stereotactic breast needle biopsy, which vary between 11.16 and 3.45 ADU/μGy, depending on the coupling method (fibre-optic taper or lens) and on the phosphor screen used.

5.3.3. X-ray calibration – deep depletion devices

5.3.3.1. The mean-variance technique

The overall conversion gain $G_e$ that relates the number of electrons generated in the CCD to the analogue-to-digital conversion signal (in ADU) of a quantum limited system can be determined by statistical analysis of the mean and variance of the observed signal. Janesick et al (1987c) and Holdsworth et al (1990) have described a general technique based on the determination of these quantities that can be applied to systems with noiseless gain.

Figure 5.15 shows a schematic representation of all the transfer functions involved in the photon detection process using a CCD camera. The system can be described in
terms of five general transfer functions, three of which are related to the CCD and two related to the signal processing circuitry (video camera and ADC). The input to the camera is a given photon flux and the final output of the system is a digital number in analogue-to-digital conversion units.

The average output signal of the system (in ADUs) can be expressed as the product of the successive conversion gains

$$ S [\text{ADU}] = \Phi \eta_0 \eta_e S_v A_1 A_2 $$

(5.10)

where $\Phi$ is the mean incident photon fluence per pixel, $\eta_0$ is the quantum efficiency of the CCD and $\eta_e$ is the number of electrons collected per absorbed photon. At optical wavelengths ($\lambda > 300$ nm) only one electron-hole pair is generated per interaction, but at x-ray energies $\eta_e$ is a function of the energy deposited and can also vary depending on the charge collection efficiency of the device. Complete charge collection requires that there are no charge-trapping centres within the CCD that would cause charge loss due to recombination. If perfect charge collection efficiency is assumed then $\eta_e$ is equal to the total number of e-h pairs generated after the absorption of an x-ray photon, that is $\eta_e = N_e$. The conversion factors that relate the number of ADUs per electron ($G_\text{e}$) and the number of ADUs per absorbed x-ray photon ($G_q$) are given respectively by

$$ G_\text{e} [\text{ADU/e}^{-}] = S_v A_1 A_2 $$

$$ G_q [\text{ADU/x-ray}] = \eta_e S_v A_1 A_2 $$

(5.11)
It can be demonstrated that for a quantum limited detector that has linear response and noiseless gain the mean observed signal and the variance are related through

\[ \sigma_S^2 = G_q \langle S \rangle + \sigma_R^2 \]  

(5.12)

where \( G_q \) is the total system gain and \( \sigma_R \) is the signal independent readout noise. Mean-variance analysis leads to the determination of the total conversion gain of the system even if the individual gains of each component are not known. It also has the added advantage that it can be performed with a very simple experimental setup, consisting only of the detector system and a variable signal source.

### 5.3.3.2. Calibration of a deep depletion device

The conversion gain of a deep depletion device was determined using the mean-variance technique. Since the effective quantum yield depends on the energy deposited, the total conversion gain of a deep depletion device will change depending on the active collecting depth (the epitaxial thickness) and on the energy distribution of the incident x-ray beam. Table 5.4 shows the results of a Monte Carlo simulation of energy deposition in deep depletion devices with epitaxial thicknesses 50, 70 and 100 \( \mu \text{m} \), at different tube voltages in the range 40 to 90 kVp. It can be observed that the mean energy deposited (\( \bar{E}_{\text{dep}} \)) changes by up to 7 keV, depending on the incident x-ray beam. No significant dependence of \( \bar{E}_{\text{dep}} \) on the epitaxial thickness was observed, but the fraction of the energy deposited is seen to change by a factor of more than five.

<table>
<thead>
<tr>
<th>kVp :</th>
<th>40</th>
<th>50</th>
<th>60</th>
<th>70</th>
<th>80</th>
<th>90</th>
</tr>
</thead>
<tbody>
<tr>
<td>( \bar{E}_{\text{dep}} ) [keV] :</td>
<td>23.0</td>
<td>25.0</td>
<td>26.8</td>
<td>27.7</td>
<td>29.0</td>
<td>30.3</td>
</tr>
<tr>
<td>( L_{\text{epi}} ) [( \mu \text{m} )]</td>
<td>Percent energy deposited</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>50</td>
<td>2.4</td>
<td>2.0</td>
<td>1.6</td>
<td>1.3</td>
<td>1.1</td>
<td>0.8</td>
</tr>
<tr>
<td>70</td>
<td>3.3</td>
<td>2.8</td>
<td>2.0</td>
<td>1.8</td>
<td>1.5</td>
<td>1.2</td>
</tr>
<tr>
<td>100</td>
<td>5.5</td>
<td>4.6</td>
<td>3.6</td>
<td>3.0</td>
<td>2.5</td>
<td>1.7</td>
</tr>
</tbody>
</table>

Table 5.4. Mean energy deposited and percent energy deposition for different kVps and epitaxial thicknesses calculated with the Monte Carlo code.

Experimental validation of the Monte Carlo results was made by comparing the spectrum of energy deposition obtained with the program with the distribution of pixel values of a flat field image taken at a sufficiently low exposure such that the number
of interactions per pixel was small. The dashed line in figure 5.16 represents the normalised energy deposited per unit energy interval on a 70 μm thick epitaxial CCD for an incident 90 kVp spectrum as calculated by the Monte Carlo program. This spectrum is compared to the normalised pixel value histogram of an image taken with device DD-3 at approximately 10 μGy entrance air kerma, after binning along the energy axis in 3 keV steps in order to reduce random fluctuations. The incident 90 kVp spectrum used for the calculation is shown as a reference with a dotted line in the same figure. It can be observed that there is a very good agreement between the measured data and the Monte Carlo results.

![Figure 5.16. Normalised energy deposited per keV on a 70 μm Epi CCD for an incident 90 kVp spectrum calculated with a MC program, and measured pixel value histogram from an image taken with device DD-3.](image)

Two mean-variance plots for device DD-3 obtained at 50 kVp and 70 kVp are shown in figure 5.17. The mean and variance data were obtained from a series of flat images taken at different exposure levels. The entrance exposure was varied by changing the distance between the x-ray tube and the CCD; two images were taken at each kVp and incident exposure. The mean signal was obtained from the average of the two images, and the signal variance was determined from a subtracted image to ensure that there were no contributions from non-stochastic noise sources. It can be observed that the total variance increased linearly with the output signal, which indicates that random fluctuations were indeed the dominant source of noise. The straight lines in figure 5.17 were obtained from linear regression analysis of the measured data. From
equation 5.12, the slope of the straight lines gives the total conversion gain of the system in ADU per x-ray, therefore

\[
\overline{G}_q = \begin{cases} 
1.57 \pm 0.04 \text{ ADU/x-ray at 50 kVp} \\
2.17 \pm 0.07 \text{ ADU/x-ray at 70 kVp}
\end{cases}
\] (5.13)

The average quantum yield can be calculated using equation 4.11 in terms of the mean energy deposited given in table 5.4, and the ionisation energy of Si at room temperature \( (\hbar \nu) = 3.65 \text{ eV per electron-hole pair} \),

\[
\bar{N}_e = \frac{\bar{E}_{dep} \text{ [eV]}}{3.65 \text{ [eV/e\textsuperscript{-}]} }}
\] (5.14)

which gives 6849 and 7589 e\textsuperscript{-}/x-ray, respectively. The average conversion factors are therefore given by

\[
\overline{G}_e = \begin{cases} 
2.29 \times 10^4 \text{ ADU/e\textsuperscript{-} at 50 kVp} \\
2.86 \times 10^4 \text{ ADU/e\textsuperscript{-} at 70 kVp}
\end{cases}
\] (5.15)

that is, each AD unit was equivalent to approximately 4367 and 3497 electrons when using these particular exposure conditions. The corresponding values for maximum system sensitivity would be reduced to 287 and 230 e\textsuperscript{-}/ADU\textsuperscript{’}.

![Figure 5.17](image)

**Figure 5.17.** Mean-variance plots obtained with device DD-3 at 50 and 70 kVp. The total conversion gain of the system was determined from the linear regression analysis of the measured data.
5.4. Noise measurements

5.4.1. Fixed pattern noise

The two main sources of non-stochastic noise in a CCD are local variations of pixel response, and localised regions of high dark current generation. Pixel to pixel non-uniformities arise during the manufacturing process. As a consequence each pixel has its own collection volume and its own characteristic quantum efficiency (Blouke et al 1987). The response variations from element to element in terms of size and spacing do not change with time and, therefore, they constitute a stationary source of noise. At high signal levels this variations of pixel-to-pixel sensitivity can become the dominant source of system noise (Janesick et al 1987a). Regions of high dark current generation in the vicinity of crystal defects in the semiconductor lattice also constitute a source of fixed pattern noise, since they lead to non-uniformities in the spatial distribution of thermally generated carriers (Beynon and Lamb 1980).

![Figure 5.18](image.png)

**Figure 5.18.** Standard deviation as a function of the mean output signal. (a) Device No-1, optical illumination. (b) Device FO-1, x-ray illumination. The linear behaviour of $\sigma_s$ with signal is associated with fixed pattern noise.

Figures 5.18a and 5.18b show the variation of the measured standard deviation as a function of the mean output signal for a series of flat images taken with devices No-1 and FO-1 using optical and x-ray illumination, respectively. The linear behaviour of the data is associated with the presence of fixed pattern noise, which dominates over
the stochastic noise fluctuations of the incident quanta. In device No-1 the fixed pattern noise is related to responsivity variations from pixel-to-pixel, as described above. In the fibre-optic coupled device non-stochastic noise arises from the physical structure of the phosphor screen.

5.4.1.1. Monte Carlo modelling of fixed pattern noise

A simple Monte Carlo simulation was used to test the validity of the assumption that the linear behaviour of standard deviation with signal was due to fixed pattern noise (FPN). The program generated a series of images of a flat field signal including only quantum (Poisson distributed) noise and increasing amounts of FPN, which was introduced in the form of a normalised "responsivity" value for each pixel. It was assumed that the responsivity followed a Gaussian probability distribution with a mean value of 1. The variance of the Gaussian distribution determined the amount of FPN present in the images, and was varied from 0 (no FPN) to 0.05.

![Figure 5.19](image_url)

**Figure 5.19.** Standard deviation as a function of mean signal level for a series of simulated images containing different amounts of fixed pattern noise.

Figure 5.19 shows a plot of the total standard deviation in the simulated images as a function of the mean signal. For $\sigma_{\text{FPN}} = 0$ the standard deviation varies as the square root of the mean signal, as it would be expected for a Poisson distributed random variable. As fluctuations in the responsivity are increased $\sigma_s$ departs very quickly from the square root behaviour. For $\sigma_{\text{FPN}}^2 = 0.05$, which corresponds to a standard
deviation of about 20% in the responsivity, the data show a perfect linear behaviour, as was actually observed on the experimental data (see figure 5.18).

5.4.1.2. Correction of pixel non-uniformities

The presence of pixel non-uniformities in the CCD makes it necessary to introduce some form of post-processing in the digital images in order to correct for response and dark current variations. It has already been mentioned that fixed pattern noise can be the dominant source of noise at high signal levels. In addition, scintillation detection requires the use of a phosphor screen that, as in this case, may introduce image artefacts. This is particularly important if CCDs are to be used in medical applications, where the imaging task consists of detecting an abnormality without a priori knowledge of its localisation, shape or size. Image artefacts could increase the number of false positive diagnostics and consequently contribute to an increased dose to the patient if repeat exposures are necessary.

The fact that the CCD response is linear over most of its dynamic range allows to use simple computer algorithms to eliminate pixel-to-pixel non-uniformities. The correction process makes use of dark and flat-field images to compensate for background trends and for fixed pattern noise, respectively. In this thesis the corrected pixel data $S_c(i)$ were calculated on a pixel by pixel basis using the equation

$$S_c(i) = \left( S_f(i) - S_d(i) \right) \frac{S_e(i)}{S_f(i)}$$

(5.16)

where $S_f(i)$ are the corresponding pixel values of the calibration frame (flat image), $S_c(i)$ are the uncorrected pixel data, and $S_d(i)$ are the dark frame pixel data. Similar algorithms have been reported by Blouke et al (1987) and Karellas et al (1992).

Using the general equation for error propagation it can be demonstrated that if the uncorrected and the calibration images contain only random noise then the variance in the corrected image would be given by

$$\sigma_c^2 = \left( \frac{\delta}{S_f} \right)^2 \left[ \sigma_e^2 + \left( \frac{S_e}{S_f} \right)^2 \sigma_f^2 \right]$$

(5.17)
where $\bar{S}_u$ and $\bar{S}_f$ are the mean values of the uncorrected and flat images, respectively, and $\delta = (\delta_f(i) - \delta_u(i))$.

Figures 5.20a shows an uncorrected image of a flat field taken with device FO-2 using the experimental arrangement described in §5.2.1. Visible illumination was used as opposed to x-rays in order to avoid extra sources of random noise such as direct hits in the silicon. The crack pattern of the CsI screen can be clearly seen in this image. Figure 5.20b shows the corresponding corrected image, as obtained using the algorithm described above. The phosphor structure practically disappears after correction, and the total measured standard deviation (see table 5.5) is reduced by a factor of approximately 5.

(a)  
(b)

Figure 5.20. Correction of fixed pattern noise in a coated device. (a) Noise in the uncorrected image is totally dominated by the crack structure of the phosphor. (b) Noise is reduced by a factor ~5 after correction.

(a)  
(b)

Figure 5.21. Correction of fixed pattern noise in a deep depletion device. The total noise increases by a factor ~1.3 after correction (see table 5.5).
Figures 5.21a and 5.21b show a similar example of the correction procedure for a deep depletion device. In this case an artefact due to a partially blocked column of pixels can be clearly seen in the image. The artefact disappears after correction, although the corrected image looks slightly noisier. The mean signal and standard deviation of all the images presented in figures 5.20 and 5.21 are given in table 5.5. The values for device No-1 are also given in the same table for comparison. It can be observed that in all cases the measured standard deviation in the corrected images ($\sigma_c$) is smaller than the value calculated with equation 5.17 ($\sigma'_c$), which indicates that all images contained at least a small contribution of non-stochastic noise.

| Table 5.5. Mean values and standard deviations for a series of flat images taken with three different devices. |
|---|---|---|---|---|---|---|---|
|   | $\bar{\delta}_f$ | $\sigma_f$ | $\bar{\delta}_d$ | $\bar{\delta}_u$ | $\sigma_u$ | $\bar{\delta}_c$ | $\sigma_c$ | $\sigma'_c$ |
| FO-2 | 172.3 | 18.0 | 15.7 | 172.5 | 18.1 | 156.1 | 3.4 | 23.3 |
| DD-3 | 168.6 | 11.2 | 12.3 | 172.7 | 11.4 | 156.4 | 14.4 | 15.0 |
| No-1 | 181.9 | 7.2 | 17.1 | 180.6 | 6.7 | 164.8 | 8.6 | 8.9 |

5.4.2. Thermal noise

Thermal excitation of valence electrons in the semiconductor lattice gives rise to the generation of some electrical current in the CCD. The thermal contribution to the output signal is present even in the absence of an input signal. When the CCD is operated at room temperature this current not only introduces a dark signal offset, thereby reducing the dynamic range of the CCD, but it also increases the background noise, and therefore limits the system sensitivity at low-signal levels. The main contribution to dark signal current in a CCD is the thermal generation of electron-hole pairs due to surface states at the Si-SiO$_2$ interface (Blouke et al 1987). This current can be expressed as (Knoll 1989)

$$J_s = C_k T_k^{3/2} \exp \left( -\frac{E_g}{2kT_k} \right)$$

(5.18)

where $C_k$ is a proportionality constant characteristic of the device, $T_k$ is the absolute temperature (in °K), $\kappa$ is Boltzmann’s constant and $E_g$ is the silicon band gap, which is also temperature dependent and is given by

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\[ E_e [\text{eV}] = 1.1557 - \left( \frac{(7.021 \times 10^{-4} T_i^2)}{(1108 + T_i)} \right) \]  \hspace{1cm} (5.19)

The dark signal of the CCD can therefore be expressed as (Holdsworth 1990)

\[ S_{\text{dark}} [\text{ADU}] = (G_e q_e^{-1} J_{ss} A_{\text{pix}} t) + S_{\text{offset}} \]  \hspace{1cm} (5.20)

where \( S_{\text{offset}} \) is any deliberate signal offset introduced in the system. This equation indicates that dark signal is expected to increase linearly with integration time. An estimation of the dark current contribution to the dark signal in device FO-1 was obtained by shielding the image section of the CCD from any visible or x-ray input and measuring the output signal as a function of the integration time. The results of these measurements are shown in figure 5.22.

![Figure 5.22. Measured dark signal as a function of integration time at room temperature. The dotted line is a linear regression of the data.](image)

The signal offset and the proportionality constant of equation 5.20 were determined from linear regression analysis of the measured data, and found to be \( S_{\text{offset}} = 9.0 \pm 0.5 \) ADU and \( G_e q_e^{-1} J_{ss} A_{\text{pix}} = 0.20 \pm 0.01 \) ADU/ms, respectively. The pixel area of this device is \( 4.84 \times 10^{-4} \text{ mm}^2 \) and the conversion constant is \( G_e = 10^{-3} \text{ ADU/e}^- \), therefore the dark current contribution was \( J_{ss} = 6.69 \text{ nA/cm}^2 \).

The Poisson shot noise associated with the dark current in terms of charge equivalent carriers is given by

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When the camera is operated in normal video mode the integration time is 20 ms, therefore the thermal noise per field calculated from equation 5.21 is \( \sigma_{\text{thermal}} = 64 \, \text{e}^- \). This figure compares well with values reported in the literature. Flint \textit{et al} (1992a) reported a thermal noise of \( \sim 150 \, \text{e}^- \) for a 100 \( \mu \text{m} \) Epi deep depletion device of the same kind as used in this investigation. Holdsworth \textit{et al} (1990) have measured a thermal noise of \( \sim 100 \, \text{e}^- \) on a time-delay and integration CCD at 5 \( ^\circ \text{C} \) using an integration time of 0.79 s, which would be approximately equivalent to 50 \( \text{e}^- \) at room temperature using an integration time of 20 ms.

An important consequence of the dark signal contribution to the total measured signal when the camera is operated in asynchronous mode is that the dynamic range of the system can be severely limited. At the maximum integration time used (\( t_I = 140 \, \text{ms} \)) the dark signal already accounted for 15% of the total 8 bit ADC range, and full saturation was reached in about 1 second.

5.5. Summary and conclusions

In this chapter a digital x-ray imaging system prototype based on an asynchronous CCD camera and a programmable frame grabber combination has been described, and its main operational parameters have been determined. The system was used in two different x-ray detection modes: direct conversion in the silicon with deep depletion devices, and scintillation detection using standard CCDs fibre-optically coupled to CsI(Tl) screens.

The basic characteristics of the individual components of the system were analysed using standard methods. Although the system is based on a 8-bit ADC, the total gain can be adjusted to extend the effective range of the ADC by a factor of approximately 15. The parameters that control the total gain of the system were found to be accurate to within 3%. Once the system was calibrated under specific illumination conditions the calibration remained stable to better than 1%, but recalibration was required if camera response or illumination conditions changed over time.

Two different calibration methods were used to determine the conversion factors that relate the CCD signal, expressed in terms of charge carriers per pixel, to the frame
A CCD-based x-ray imaging system

grabber signal, which is given in terms of analogue-to-digital conversion units. The conversion factors depend on the CCD architecture and on the x-ray detection mode. For scintillation detection the conversion factor at maximum gain was found to be approximately $26 \text{ e}^-/\text{ADU}'$ for EEV-02 anti-bloomed devices, and $68 \text{ e}^-/\text{ADU}'$ for EEV-02 non anti-bloomed devices. For deep depletion devices the conversion factor depends on the incident x-ray energy. A statistical method was used to determine the total conversion gain of a 70 $\mu$m epitaxial device using polyenergetic x-ray spectra, and it was found to vary between 287 and 230 $\text{ e}^-/\text{ADU}'$ for 50 and 70 kVp spectra, respectively.

The system was observed to have very good linearity of response to both visible and x-ray illumination. The sensitivity of the system to x-ray radiation was dependent upon the CCD used. As expected, coated devices showed higher sensitivity than deep depletion CCDs. The maximum sensitivity of the system ranged between 2.31 ADU'/\mu GY for a 70 $\mu$m deep depletion device up to 9.09 ADU'/\mu GY for a 130 $\mu$m CsI(Tl) coated device. The sensitivity of coated devices was governed not only by the phosphor thickness, but also by the coupling efficiency of the intervening optics.

First order statistics were used to analyse non-stochastic noise and thermal noise. It was observed that at high signal levels fixed pattern noise was the dominant source of noise in both bare CDDs and coated devices. For bare devices under visible illumination fixed pattern noise was associated with sensitivity variations from pixel to pixel, whereas in coated devices the main source of non-stochastic fluctuations was the physical structure of the phosphor screen. A simple correction algorithm that eliminates fixed pattern noise was described and examples of its use were given.
CHAPTER 6

Evaluation of the CCD system

This chapter presents the results of an evaluation of the imaging performance of the CCD-based x-ray imaging system that was described in Chapter 5. The spatial transfer properties and the noise performance characteristics of the system are determined. These quantities are then used to obtain the spatial-frequency dependent detective quantum efficiency using the methods that were discussed in Chapter 1. Comparisons are drawn between scintillation and direct x-ray detection methods and the relative merits of these two different imaging modalities are discussed.

6.1. Transfer Function analysis

6.1.1. Materials and methods

Two different methods were used in the determination of the modulation transfer function of the CCD system. The MTFs of the fibre-optic coupled CCDs were obtained from the square wave response function (SWRF) of the system measured from images of a resolution test pattern, and the MTF of a deep depletion device was obtained using the extended-edge technique. As discussed in §1.5, these methods require that the test object (the bar pattern in the first case or the edge in the second case) are sampled above the Nyquist frequency of the digital system in order to avoid aliasing effects due to the finite sampling aperture. Therefore, the test objects were placed at a slight angle with respect to the pixel sampling grid; in this way the information from several lines could be used to oversample the test object.

In order to minimise the effect of the focal spot on the detector MTF measurements the x-ray tube was placed as far as possible from the detector, while maintaining the test object in close contact with CCD. The exposure conditions were set to 70 kVp, 200 mA and 20 ms integration time. The frame grabber settings were adjusted accordingly for each of the CCDs tested to ensure good image contrast.
6.1.1.1. Square wave response

A commercially available bar pattern phantom based on a Huttner type 18 grid (Hay et al. 1985) was used for the square wave measurements. The phantom is made of 100 μm thick lead foil embedded in a 1 mm thick plastic case. It contains 21 groups of bar patterns, each group comprising 4.5 cycles (line pairs) of a given spatial frequency. The spatial frequencies range from 0.5 up to 5.0 lp/mm.

![Image of the Huttner grid taken with device FT-1 and DD-3.](https://example.com/image)

Figure 6.1. Images of the Huttner grid taken with (a) fibre-optic coupled device FO-1; (b) deep depletion device DD-3.

Figure 6.1a shows an image of a section of the Huttner grid taken with device FT-1. Four bar pattern groups can be clearly seen, which correspond to 4.0, 2.8, 2.0 and 1.4 lp/mm. For comparison, figure 6.1b shows an image of the bar pattern groups corresponding to 2.5, 3.55 and 5 lp/mm taken with device DD-3. It can be observed that the image taken with the coated device is slightly more blurred, while in the image taken with the deep depletion device the maximum spatial frequency of the phantom (5 lp/mm) can be very easily resolved, although image noise is also quite noticeable.

Two image profiles parallel to the raster scan direction are also shown in figure 6.1a. The values of the maxima and minima of the grid profiles, $S_{\text{max}}$ and $S_{\text{min}}$, were determined by applying a running average to an oversampled profile obtained from 150 registered profiles such as these. An example of the result from this calculation is shown in figure 6.2. The effect of amplitude modulation on the square wave as the frequency of the bar pattern increases can be clearly seen. $S_{\text{max}}$ and $S_{\text{min}}$ were measured for the 3 central cycles of each spatial frequency and the average values were used to
calculate the modulation of the square wave according to equation 1.6. The results were normalised with respect to the modulation of the lowest frequency bar pattern (0.5 mm⁻¹) to obtain the SWRF of the system. The MTF was determined by applying a 4-term expansion of the Coltman correction formula (equation 1.11) to the SWRF.

![Figure 6.2. Average of 150 registered profiles taken from the image of the Huttner grid phantom shown in figure 6.1a. Four spatial frequencies can be clearly distinguished, from left to right: 4.0, 2.8, 2.0 and 1.4 lp/mm.](image)

6.1.1.2. Edge spread function

The resolution limit of the deep depletion device was well beyond the maximum spatial frequency of the bar pattern phantom, therefore a different approach was used in order to measure its MTF performance. The extended-edge technique (see §1.5.3) combined with scan-line averaging was used to overcome the problem of noise in single image profiles.

A finely machined Cu edge was placed directly on top of the CCD at an angle nearly perpendicular to the raster direction, and an image was taken using the same exposure conditions as described in the previous section. The angle of the edge with respect to the CCD columns was measured from the digital image and found to be $\theta = 5.25^\circ$. The displacement of the edge between consecutive horizontal profiles (in pixel units) is given by $\tan \theta = 0.092$ pixels. The number of profiles needed to complete a single pixel shift was, therefore, $1/(\tan \theta) = 11$. A series of 220 consecutive profiles parallel to the raster direction was then taken and divided into 20 groups. Each group was
registered using the pixel shift calculated above. The 20 registered profiles were averaged to obtain a single oversampled profile. Finally, an 11 point running-average was applied to the oversampled profile to obtain the ESP (see figure 6.3). The MTF was calculated from the ESP using numerical differentiation followed by a discrete Fourier transform of the LSF. The MTF values were corrected for the effect of the finite-element differentiation using the method described in §1.5.2.1.

![Figure 6.3. Comparison of the edge spread functions of devices DD-3 and FO-2 obtained using the extended-edge technique.](image)

### 6.1.2. Results and discussion

#### 6.1.2.1. Coated devices

Figure 6.4 shows the results of the MTF measurements for all three fibre-optic coupled devices. Although all devices were able to resolve all the spatial frequencies in the bar pattern phantom, in all cases the MTF was reduced to values between 10% to 15% at $f = 5 \text{ mm}^{-1}$. Since the geometrical MTF due to the finite CCD pixel size (see §4.3.7.1) drops only to -0.95 and -0.98 at this spatial frequency for pixel sizes of 35.2 μm (FT-1 and FT-2) and 22 μm (FO-2), it is evident that the geometrical effect of the CCD sampling aperture is not important in this case. A comparison between devices FT-1 and FT-2 (figure 6.4a) shows that the device with the thinner phosphor screen (FT-1) has better MTF performance, as it would normally be expected. On the other hand, both devices use the same fibre-optic taper and the only difference
between them is the thickness of the phosphor screen. Therefore, these results indicate
that light channelling in the screen is not perfect, and suggest that the MTF is
-dominated by the propagation of visible light in the phosphor.

Figure 6.4. Modulation transfer functions of the fibre-optic coupled CCDs. (a) comparison between the tapered devices FT-1 and FT-2. (b) comparison between FT-2 and device FO-2.

As discussed in §4.5.5.2, the presence of stray light propagating within the taper
degrades the contrast in fibre-optic coupled devices. For a given imaging situation the
image contrast in a tapered device can be significantly reduced since the contrast
degradation is proportional to the square of the effective numerical aperture. Thus, for a
1.6× taper (NA^2 = 0.39) the degradation factor is 2.56 times larger than that of a
fibre-optic faceplate which has the same packing fraction. It was also mentioned that
the most common technique used to reduce contrast degradation is to introduce an
extra-mural absorber (EMA) between the individual fibres. However, the reduction of
contrast degradation is more effective at lower spatial frequencies since the bulk
absorption of the EMA is not perfect (i.e. it is less than 100%). As a consequence the
MTF of an optical-fibre taper is expected to have a steeper drop off at high spatial
frequencies when compared with a fibre-optic faceplate. This effect can be observed
in figure 6.4b, where the MTF of device FT-2 is compared with that of device FO-2.
Up to a spatial frequency of 4 mm\(^{-1}\) FT-2 has better MTF performance, as it would be
expected in terms of phosphor thickness only. However, this trend seems to be
reversed just above 4 mm\(^{-1}\), where FO-2 seems to have slightly better MTF.
6.1.2.2. Theoretical analysis of the MTF of coated devices

The MTF of a phosphor-coated device can be characterised using a model originally developed by Swank (1973a) for a transparent phosphor with exponential absorption and a non-reflective backing. Westmore and Cunningham (1993) have modified the model to include the effects of an index-matching overcoat placed between the phosphor and a fibre-optic faceplate, and of the optical coupling efficiency of the phosphor-fibre-CCD system.

![Diagram of the geometry used for calculating the MTF of the phosphor-fibre-CCD system.](image)

**Figure 6.5.** Schematic diagram of the geometry used for calculating the MTF of the phosphor-fibre-CCD system.

Figure 6.5 shows a diagram of the geometry used to calculate the LSF for the coupling between the phosphor and the fibre optic. A line-beam of x-rays is assumed to be incident from $z = \infty$ towards the $xy$ plane along the $Y$ axis. The LSF of the system is determined by finding the number of light quanta from an emitting element $dydz$ of the screen that pass through the area $dA$ of the fibre-optic faceplate and are then transmitted by the fibre optic to the CCD. If $D_p$ and $D_o$ represent the thicknesses of the phosphor and the overcoat, then the LSF is given by

$$LSF(x) = k \int_{D_o}^{D_p} \int_{-\infty}^{\infty} \frac{z}{r^2} e^{-w[D_p-d]} E_p^f(\theta_p, \theta_f) E_p^o(\theta_o, \theta_f) dy dz$$ (6.1)

where $E_p^f(\theta_p, \theta_f)$ and $E_p^o(\theta_o, \theta_f)$ are the transmittances from the phosphor to the fibre optic and from the fibre optic to the CCD, respectively, and $k$ is a normalisation constant. Since the light is unpolarised, $E_p^f$ and $E_p^o$ have to be evaluated for two
orthogonal components and the results have to be averaged. In general, for a wave passing from a medium with refraction index \( n_0 \) to a medium with refraction index \( n_i \), the two components are given by

\[
E_\perp = \frac{4n_0 n_i \cos \theta_0 \cos \theta_i}{(n_0 \cos \theta_0 + n_i \cos \theta_i)^2}
\]

(6.2)

\[
E_\parallel = \frac{4n_0 n_i \cos \theta_0 \cos \theta_i}{(n_i \cos \theta_0 + n_0 \cos \theta_i)^2}
\]

(6.3)

The transmittances were evaluated using the refraction indexes \( n_\text{phosphor} = 1.79 \), \( n_\text{fibre core} = 1.81 \), and \( n_\text{CCD} = 4.12 \) for the phosphor, fibre core and CCD (silicon), respectively. The maximum angle of acceptance at the phosphor-fibre interface was calculated in terms of the numerical aperture of the fibre-optic faceplate (equation 4.20) and found to be 34°. The results of the calculation are shown in table 6.1. The average transmittances at both interfaces are practically 1, and as a first approximation can be considered to have no effect on the system LSF.

<table>
<thead>
<tr>
<th></th>
<th>( \theta_0 )</th>
<th>( \theta_1 )</th>
<th>( E_\perp(\theta_0, \theta_1) )</th>
<th>( E_\parallel(\theta_0, \theta_1) )</th>
<th>( \bar{E} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>phosphor-fibre</td>
<td>34.0</td>
<td>33.5</td>
<td>1</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>fibre-CCD</td>
<td>33.5</td>
<td>14.0</td>
<td>0.79</td>
<td>0.90</td>
<td>0.85</td>
</tr>
</tbody>
</table>

Equation 6.1 was integrated numerically to find the LSF. The discrete Fourier transform of the LSF data was then taken to obtain the MTF of the system. The parameters used in the calculation were \( \mu = 8.33 \, \text{mm}^{-1} \), which is the linear attenuation coefficient of CsI at 35 keV, a phosphor thickness of 130 \( \mu \text{m} \) and an overcoat thickness of 20 \( \mu \text{m} \). The results are shown with a dotted line in figure 6.6, were they are compared to the measured MTF of device FO-2. It can be observed that there is excellent agreement between the measured data and the MTF model, particularly at high spatial frequencies. These results offer further evidence that the phosphor screen is the dominant factor in the MTF performance of the system.
6.1.2.3. Deep depletion device

Figure 6.7 shows the MTF of device DD-3, as determined using the extended-edge technique. Unlike the MTFs of the fibre-optic coupled devices it can be observed that noise fluctuations are present at practically all spatial frequencies. The dotted line in the same figure represents a least-squares fit of the experimentally measured data. For comparison, the geometrical MTF of the square sampling aperture associated to the CCD pixel size (16.5 μm) is shown with a dashed line. In this case the Nyquist limit of the CCD is 30.3 mm⁻¹, however the spatial frequencies for which the MTF drops to 50% and 10% are only 6 mm⁻¹ and 18 mm⁻¹, respectively.

The degradation in MTF performance with respect to the geometrical limit can be attributed to carrier diffusion effects in the undepleted silicon following the absorption of one x-ray photon. As discussed in §4.4.2, the depletion depth of the CCD is determined by the resistivity of the epitaxial layer and the gate voltage applied to the metal electrodes. Using the physical parameters given in table 5.2 and equation 4.12, with a gate voltage \( V_g \) of 10 volts (which is typical for these devices), the depth of the depletion layer was estimated as 17 μm. Since the epitaxial thickness of DD-3 is 70 μm this means that the CCD would collect signal from ~53 μm of undepleted silicon. It has been shown using both analytical (Hopkinson 1987) and Monte Carlo...
methods (Janesick et al 1987b, Lumb et al 1987) that the maximum diameter of an electron cloud diffusing in the field-free region is equal to approximately twice the depth of the undepleted layer. Using the fitted MTF data shown in figure 6.7 the LSF of the system was obtained by means of the inverse Fourier transform. The full width at tenth maximum of the LSF was then measured and found to be \( \sim 120 \, \mu\text{m} \). This value compares well with the approximate size of the electron cloud predicted by the models mentioned above, which would be \( \sim 106 \, \mu\text{m} \).

\[ \text{Figure 6.7. Modulation transfer function of the deep depletion device DD-3. The dashed line represents the geometrical component of the MTF due to the square sampling aperture of the CCD.} \]

6.1.2.4. Comparison of MTF performance

A comparison of the MTF performance of the CCD system with that of two typical film-screen combinations based on Kodak Lanex screens (Doi et al 1982) is shown in figure 6.8. Both film-screen systems use double screens based on \( \text{Gd}_2\text{O}_2\text{S(Tb)} \), each screen having a nominal thickness of 40 \( \mu\text{m} \) (Lanex-Fine) and 80 \( \mu\text{m} \) (Lanex-Regular). The deep depletion device (DD-3) is seen to have the best overall performance of all the systems in this comparison, followed by the Lanex-Fine system. The MTF of device FT-1 is closer to that of Lanex-Fine, and is much better than that of Lanex-Regular. In general terms it was observed that the spatial transfer characteristics of the coated devices were better than those of the general purpose film-screen system, but did not match the high resolution system.
6.1.3. Limiting resolution and noise equivalent aperture

The spatial resolution limit of an imaging system is defined as the highest spatial frequency of a unit-contrast test object that can be detected visually on a radiological image (Hay et al. 1985). The limiting resolution of the CCD system was determined from images of a Type 9 star pattern phantom. All the images were taken under the same exposure conditions used for the MTF measurements. The phantom is made of 0.03 mm thick Pb, and contains 4 groups of 14.5 line pairs each. A single line pair has an angular aperture of 2°, which is equivalent to 180 line pairs in a full circle. The phantom has an external diameter of 45 mm, and therefore at the periphery of the circumference the spatial frequency is \(\frac{180}{(\pi \times 45 \text{ mm})} = 1.27 \text{ lp/mm}\); at any smaller diameter \(D\) the spatial frequency is \(1.27 \times \frac{45}{D}\). Since the innermost diameter of the bar pattern is 3.2 mm, then the maximum spatial frequency that can in principle be observed is 17.90 lp/mm. The limiting resolution \(R_L\) was obtained by measuring the radial distance from the innermost diameter of the bar pattern to the point where the last frequency in the image could be seen.

Figures 6.9a and 6.9b show two images of one group of the star pattern phantom taken with devices FO-2 and DD-3, respectively. The image taken with the coated device corresponds to a single exposure and was corrected for fixed pattern noise, while the
image taken with the deep depletion device is the result from averaging four frames in order to reduce random noise. The limiting resolution was determined from at least four such images for each CCD taken at different orientations of the bar pattern. However, no significant dependence on the values of the limiting resolution with the orientation of the phantom was observed.

![Image](a) ![Image](b)

**Figure 6.9.** Images of the resolution test pattern phantom taken with (a) a coated device (FO-2), and (b) a deep depletion (DD-3).

Table 6.2 lists the results of the measurements for all the fibre-optic coupled devices and one deep depletion device. The corresponding pixel size ($p$), Nyquist frequency ($f_N$) and the value of the spatial frequency at which the modulation transfer function is reduced to 5\% ($f_m$) for each CCD are also given in the same table. As it might be expected from the MTF measurements the deep depletion device had the highest limiting resolution of all the CCDs tested, while FT-1 was the coated device with the best resolution performance.

It can be observed in table 6.2 that, apart from device DD-3, there is very good agreement between the value of $R_\ell$ and the limiting resolution estimated from the MTF measurements, which is given by $f_N$. On the other hand, the value of $R_\ell$ is much lower than the theoretical resolution limit, which is given by $f_N$. This indicates that the spatial transfer properties of the system are determined by secondary processes, such as visible light diffusion in the phosphor screen, or diffusion of the photo-generated charge in the undepleted region in deep depletion devices.
The noise equivalent aperture (NEA, see §1.5.4) of the CCDs was calculated with equation 1.21 restricting the upper integration limit to the limiting resolution of the system, and using the MTF data from §6.1.2. The results are shown in the last column of table 6.2. The smallest NEA corresponds to the deep depletion device. For the coated devices the NEA increases with phosphor thickness. Although FO-2 has a higher spatial resolution limit than FT-2, its NEA is bigger, which reflects the fact that FT-2 has a slightly better MTF at lower frequencies.

<table>
<thead>
<tr>
<th>Device</th>
<th>$p$ [µm]</th>
<th>$f_N$ [mm$^{-1}$]</th>
<th>$f_R$ [mm$^{-1}$]</th>
<th>$R_L$ [lp/mm]</th>
<th>$a_e$ [mm$^2$]</th>
</tr>
</thead>
<tbody>
<tr>
<td>DD-3</td>
<td>16.5</td>
<td>30.3</td>
<td>20</td>
<td>14.7 ± 0.7</td>
<td>0.011</td>
</tr>
<tr>
<td>FT-1</td>
<td>35.2</td>
<td>14.2</td>
<td>10</td>
<td>9.7 ± 0.3</td>
<td>0.083</td>
</tr>
<tr>
<td>FT-2</td>
<td>35.2</td>
<td>14.2</td>
<td>7.8</td>
<td>8.0 ± 0.3</td>
<td>0.122</td>
</tr>
<tr>
<td>FO-2</td>
<td>22.0</td>
<td>22.7</td>
<td>8.2</td>
<td>8.4 ± 0.3</td>
<td>0.146</td>
</tr>
</tbody>
</table>

Workman and Cowen (1993) have recently reported values of $R_L$ and NEA for three different film-screen (f-s) combinations used in routine clinical practice. They found that the spatial resolution limit and NEA of the fastest system (DuPont Quanta III screen + DuPont Cronex 10S film) were 8.0 lp/mm and 0.2 mm$^2$, respectively, while for the system with highest definition (DuPont Quanta Detail + Cronex 10S film) were 12.5 lp/mm and 0.038 mm$^2$. The resolution of the coated devices is, therefore, generally better than that of the general purpose f-s system, but considerably lower than that of the high resolution f-s system. The deep depletion device, on the other hand, has much better resolution characteristics than any of the two f-s systems.

6.2. Noise spectral analysis

6.2.1. Method of measurement

Unlike MTF measurements, which provide information about the relative performance of a particular device at different spatial frequencies, Wiener spectral analysis represents an absolute quantity of system performance. For this reason all the results in this section are given in terms of relative noise fluctuations, in order to avoid renormalisation problems due to the different gains employed with different CCDs.
The determination of the Wiener spectrum was based on the scanning slit method, which has already been described in Chapter 1 (§1.6.4.3). The image data used to calculate the noise power spectrum (NPS) were acquired by uniformly irradiating the entire detector area. A two-dimensional noise matrix was then obtained by subtracting the average background intensity from each pixel on the flat field image. Only the Wiener spectrum in the raster direction, $W_{\text{asis}}(u,0)$, was calculated; therefore the scanning slits were synthesised in the vertical direction. All the spectra were corrected by the finite length of the scanning slit.

A series of computer-generated images containing only uncorrelated Poisson noise were used to validate the programs to calculate the Wiener spectrum. Since the images have a finite size a compromise had to be reached between the slit length and the maximum number of synthesised slits that could be used to calculate the ensemble average. It was observed, using both experimental and computer-generated images, that amplitude variations on the NPS estimates were more sensitive to the number of slits averaged than to the slit length; therefore the slit length was set to 12 pixels in order to reduce the uncertainty in the spectral values. Since the CCD images were 288 pixels wide this was equivalent to 24 synthesised slits for the ensemble average. The error in the NPS estimates calculated in this way was less than 10%. The scale value was verified by calculating the total noise power, which is given by the integral of $W_s(f)$, and it was found to be within 5% of the total signal variance ($\sigma_s^2$) measured on the digital images.

The relative contribution of quantum and electronic noise per frame was obtained from two single-frame images taken under the same exposure conditions. The images were subtracted and the Wiener spectrum of the subtracted frame was calculated. The Wiener spectrum was divided by two and then it was renormalised with respect to the mean signal obtained from the flat field images. The Wiener spectrum of dark noise, which includes the contributions from thermal and electronic noise, was obtained with two dark frames using the same subtraction method. For comparison, $W_{s\text{,dark}}$ was renormalised in terms of the same mean signal used to obtain the quantum and electronic noise contribution.

### 6.2.2 Magnitude and shape of the Wiener spectrum

The total noise in CCD images is determined by many independent contributions,
such as the input quantum noise, thermal noise and electronic noise. In addition, the presence of spatial non-uniformities in the detector contributes with non-stochastic fluctuations, as discussed in §5.4.1. The magnitude and shape of the NPS depend on the nature of the stochastic fluctuations during the process of image formation and on the spatial transfer characteristics of the imaging system. Figure 6.10 shows the relative contribution of different noise sources to the total system noise for a coated device. The spectra were obtained from images taken with device FT-2 under standard exposure conditions. The total system noise was derived from a single-frame, uncorrected image; therefore, it contains the fixed pattern noise associated with the structure of the CsI layer. The relative contributions from quantum noise and from dark noise were obtained using the subtraction method described above.

![Figure 6.10. Contribution of individual noise sources to the noise power spectrum of a coated device. The spectra were calculated from images taken with device FT-2.](image)

In figure 6.10 the total system noise is dominated by the structure mottle of the CsI screen. The Wiener spectrum of the total noise has very large components at low spatial frequencies, with a difference of almost three orders of magnitude between the lowest and highest frequency spectral values. Only above $f=6 \text{ mm}^{-1}$ the total noise and quantum noise are comparable. Although the texture of the phosphor screens varied considerably, the Wiener spectra of uncorrected images obtained with fibre-optic coupled devices were very similar in terms of the spectral distribution. Only the magnitude of the fluctuations (i.e. the total noise power) changed, with a total...
standard deviation between 4 and 7 times higher than that of quantum and electronic noise, depending on the structure of the screen. For example, the total relative variance \( \left( \frac{1}{SNR_{\text{tot}}} \right) \) for device FT-2 increased by a factor of \( \sim 23 \) after image correction, which represents a reduction in the measured noise by a factor of \( \sim 5 \).

In the same figure the WS of dark noise is observed to be independent of spatial frequency, which indicates that it is composed by random fluctuations only. The dark noise includes the effect of readout, frame-independent electronic noise and of thermal noise, and it establishes the noise floor of the system. Since both thermal and readout noise depend on the particular CCD architecture, the magnitude of the total noise power (in absolute units) was different for each of the devices tested. Typical values were 128, 425 and 735 (±15%) noise equivalent electrons for FT-2 (anti-bloomed), FO-2 (non anti-bloomed) and DD-3 (deep depletion), respectively.

The shape of the WS of quantum and electronic noise is similar to the WS of quantum mottle and film granularity in a typical film-screen combination. In both cases two noise levels can be distinguished. For the CCD system the magnitude of the WS at lower spatial frequencies is determined by the quantum fluctuations corresponding to the detection of individual x-ray photons in the CsI screen. As the frequency increases the spectrum gradually levels off, and finally becomes dominated by electronic noise above \( f = 10 \text{ mm}^{-1} \) in this case. The transition between these two noise levels is modulated by the spatial transfer properties of the system.

Figure 6.11 compares of the Wiener spectrum of the fibre-optic coupled devices as determined from images taken using the same entrance air kerma (\( \sim 220 \mu \text{Gy} \)). In this plot the spatial frequency has been normalised with respect to the Nyquist frequency \( (f_N) \) for each device, and the Wiener spectral values were multiplied by \( f_N \) in order to preserve the total area under the curve. It can be observed that the spatial distribution of the noise fluctuations is very similar in all cases. The relative contribution due to the scintillation detection process dominates in all the spectra up to \( f = 0.7f_N \).

The total relative variance of FO-2 was found to be about 4 times higher than that of FT-2. Since the difference in sensitivity between these two devices was about a factor of two (see §5.3.2), it means that noise performance was approximately equivalent. Given that the dark noise of FO-2 was higher, this variation can only be accounted for by the difference in conversion and coupling efficiency between these two systems. The relative variance of FT-2 was only marginally better than that of FT-1. These two
devices were based on the same CCD chip and fibre-optic taper, therefore the noise was also very similar and the difference is due to the higher sensitivity of FT-2.

As it would be expected, the magnitude of the relative noise fluctuations of quantum and electronic noise per frame was observed to increase as the exposure level was decreased. Figure 6.12a shows the variation in $W_{\Delta SV}$ obtained with device FT-2 when the entrance air kerma was reduced from 209 μGy to 44 μGy, that is, a reduction by a factor of ~5. The total relative variance increased accordingly by a factor of 5, which indicates that after correction for FPN the system was quantum limited. However, it can be observed that the spatial frequency range over which quantum fluctuations dominate changed with the entrance air kerma. At 209 μGy quantum noise dominates up to $f = 10 \text{ mm}^{-1}$, while at 44 μGy the quantum noise component has a steeper drop off, in such a way that electronic noise dominates above $6 \text{ mm}^{-1}$. Therefore, at lower doses the spatial frequency range over which the system is quantum limited would be reduced.

Unlike coated CCDs, the spatial distribution of the relative noise fluctuations in the deep depletion device does not change with the exposure level, as it can be seen in figure 6.12b; a 16-fold increase in the entrance air kerma leads to a decrease in the Wiener spectral values by approximately the same factor. In this case the stochastic fluctuations associated with the detection process totally dominate the spectrum at all
spatial frequencies. The low quantum efficiency and high conversion gain for direct x-ray conversions in the silicon lead to big variations in energy deposition between adjacent pixels. These fluctuations generate a distribution of pixel values that is not necessarily Gaussian, but that reproduces the spectrum of energy deposition in the device (see §5.3.3.2). The magnitude of the NPS is closely related to the form of this distribution, while the shape is determined by the spatial transfer characteristics of the system, which are dominated by charge diffusion effects, as discussed in §6.1.2.3.

A direct comparison of the Wiener spectra obtained with FT-2 and DD-3 at approximately the same level of exposure reveals that the magnitude of the spectral values of DD-3 is about 2 orders of magnitude higher than those of FT-2. Therefore, although it has been shown that DD-3 has much better spatial resolution than any of the fibre-optic coupled devices, its noise performance would severely limit the ability of the system to work at low doses.

6.2.3. Effect of subtraction and averaging

The effect of subtraction and averaging on the NPS can be seen in figure 6.13, which compares the Wiener spectra of a single-frame, uncorrected image \( W_{s_{\text{unc}}} \) with those of a subtracted image \( W_{s_{\text{sub}}} \) and a 10-frame average image \( W_{s_{\text{avg}}} \) taken with device DD-3 using an entrance air kerma rate of 14 \( \mu \)Gy per frame. Once again, the shape of
the NPS remained approximately the same in all cases. The magnitude of $W_{\text{sub}}$ was found to be twice that of $W_{\text{sng}}$, which means that any FPN present in the image would be negligible compared to random noise at this exposure level, as indicated by equation 1.38. However, $W_{\text{avg}}$ was found to be only about 6 times lower than $W_{\text{sng}}$, as opposed to the 10-fold reduction that would be expected if there were no contribution from FPN. It can easily be shown using equations 1.38 and 1.39 that if $N_f$ frames are averaged, including the contribution from FPN, the Wiener spectrum of the averaged image should be given by

$$W_{\text{avg}}(f) = \frac{1}{N_f} W_{\text{Q}}(f) + W_{\text{FPN}}(f)$$

where $W_{\text{Q}}$ represents the contribution from quantum and electronic noise. Therefore, an estimation of the relative contribution of fixed pattern noise to the total measured noise in the CCD images could be obtained by averaging a large number of frames.

![Figure 6.13](image)

Figure 6.13. (a) NPS for single-frame (Sng.), subtracted (Sub.) and 10-frame average (Avg.) images taken with device DD-3. (b) Standard deviation as a function of number of frames averaged for the same device.

Figure 6.13b shows the measured standard deviation as a function of the number of frames averaged for a series of images taken with device DD-3. The dotted line is a least squares fit of the measured data, and represents the inverse square root behaviour that would be expected for the reduction of random noise as the number of averaged frames increases. However, in the limit when the number of frames is very large $\sigma_s$. 

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tends asymptotically to a constant value that is bigger than zero. This value, which represents the fixed pattern noise, was found to be 0.9 ADU' for this device, which would be equal to 207 e\(^{-}\) using the conversion factor determined in §5.3.3.

6.3. Detective quantum efficiency

The spatial-frequency dependent DQE models described in Chapter 1 (§1.8) were used to investigate the DQE performance of the CCD system. The x-ray detection modes used, scintillation and direct x-ray detection, involve a different sequence of events during the process of image formation, as discussed in Chapter 4. Therefore, a different approach was taken to obtain the DQE characteristics of the system.

Direct x-ray detection with deep depletion devices can be considered a simple two-stage detection process: binary selection followed by scattering. The DQE can therefore be obtained directly from the measured MTF and NPS, using the definition given by equation 1.55. Using this method it is only necessary to determine the incident x-ray fluence at the entrance plane of the detector, apart from the spatial transfer and noise characteristics.

Scintillation detection with fibre-optic coupled CCDs involves a more complex sequence of events. Binary selection in the CsI screen (\(\eta_q\)) is followed by non-deterministic amplification during the process of light emission, which can be described by the mean number of visible photons (\(\bar{m}\)) and the Poisson excess factor of the screen (\(\varepsilon_m\)). The transport of the optical photons from the point of generation in the screen to the CCD is described by the MTF of the system, along with any losses due to optical coupling efficiency (\(\eta_{op}\)). Finally, the visible photons are detected by the CCD with optical efficiency \(\eta_v\), which constitutes another binary selection process. In this case the model for multi-stage imaging systems described in §1.8.3 can be used to determine \(DQE(f)\). Using equation 1.60 and the detection sequence described above the expression for \(DQE(f)\) is given by

\[
DQE(f) = \eta_q \left[ 1 + \frac{\varepsilon_m}{\bar{m}} + \frac{1}{\eta_v \eta_{op} \bar{m} \text{MTF}^2(f)} \right]^{-1}
\]  

(6.5)

It can be demonstrated that for a phosphor screen with high conversion gain (i.e. \(m >> 1\)) the Poisson excess factor is related to the Swank factor by \(1 + (\varepsilon_m/\bar{m}) \approx 1/I_x\).
Equation 6.5 can be used to calculate the DQE of the fibre-optic coupled devices in terms the measured MTF if all of the other parameters are known. A calculation of the quantum efficiency and the Swank factor using Monte Carlo methods is presented in §6.3.1. The optical coupling efficiency is determined in §6.3.2.

6.3.1. Image transfer properties of the CsI screen

The EGS4 code system (Nelson et al 1985) was used to write a Monte Carlo program to simulate the x-ray interaction processes in the CsI layer. The Monte Carlo simulation considered a monoenergetic pencil beam incident on a semi-infinite, homogeneous slab of CsI of uniform thickness. The incident energy of the monoenergetic beam was varied from 10 to 90 keV in 1 keV steps in order to cover the energy range used in the experiments. $10^5$ photons were used for each incident energy. An estimation of the errors involved in the calculation was obtained by performing 10 independent runs, each one with $10^4$ incident photons, and was found to be no more than 5%.

For each incident energy the program calculated the distribution of energy deposition in the phosphor, normalised the corresponding spectrum to unit area and calculated the first three moments of the normalised absorbed energy distribution (AED). Figure 6.14 shows some examples of the AEDs obtained with the Monte Carlo program. The spectra show the photopeak, centred at the energy of the incident x-ray beam $(E_i)$ and a series of K-escape peaks, which are located at $E_i - E_k$, where $E_k$ is the energy of the K-fluorescent photon.

The effect of changes in the incident energy (35 and 40 keV) on the AED for a 130 μm thick CsI screen is shown in figure 6.14a. Figure 6.14b shows the effect of phosphor thickness (70 and 130 μm) on the AED for an incident energy of 35 keV. It can be observed that the magnitude of the K-escape peak is comparable to that of the photopeak, which means that the variation in energy deposition for incident energies above the K-edge would be significant.

Tables 6.3 and 6.4 show the number of counts integrated over the K-escape peak $(n_k)$, the photopeak $(n_\gamma)$ and the total AED $(n_T)$ for several incident x-ray energies and phosphor thicknesses. The K-escape fraction is given by $K_f = n_k/n_\gamma$. As indicated in §4.5.1, about one half of the K-fluorescent photons escape from the screen. For energies above of the Cs edge (36 keV) the K-escape fraction remains fairly constant,
decreasing only by ~4% when $E_i$ changes from 40 keV to 80 keV. Phosphor thickness has a greater effect on the K-escape fraction; it can be seen in table 6.4 that $K_f$ is reduced by ~18% when the phosphor thickness is increased from 70 to 130 µm.

![Figure 6.14. Monte Carlo simulated scintillation spectra. (a) 35 keV and 40 keV incident x-ray energy and 130 µm CsI screen. (b) incident x-ray energy 35 keV, phosphor thicknesses 70 and 130 µm CsI.](image)

### Table 6.3. K-escape peak ($n_k$), photopeak ($n_p$) and K-escape fraction ($K_f$) for a 130 µm CsI screen at different incident x-ray energies.

<table>
<thead>
<tr>
<th>$E_i$ [keV]</th>
<th>$n_k$</th>
<th>$n_p$</th>
<th>$n_r$</th>
<th>$K_f$</th>
</tr>
</thead>
<tbody>
<tr>
<td>35</td>
<td>24919</td>
<td>40312</td>
<td>65231</td>
<td>0.382</td>
</tr>
<tr>
<td>40</td>
<td>30952</td>
<td>42503</td>
<td>73455</td>
<td>0.421</td>
</tr>
<tr>
<td>60</td>
<td>14734</td>
<td>20722</td>
<td>35909</td>
<td>0.411</td>
</tr>
<tr>
<td>80</td>
<td>7489</td>
<td>10272</td>
<td>18489</td>
<td>0.405</td>
</tr>
</tbody>
</table>

### Table 6.4. K-escape peak, photo-peak and K-escape fraction for three different screen thicknesses at different incident x-ray energies.

<table>
<thead>
<tr>
<th>$D_p$ [µm]</th>
<th>35 keV</th>
<th>40 keV</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$n_k$</td>
<td>$n_p$</td>
</tr>
<tr>
<td>70</td>
<td>20563</td>
<td>22776</td>
</tr>
<tr>
<td>100</td>
<td>23659</td>
<td>31884</td>
</tr>
<tr>
<td>130</td>
<td>24919</td>
<td>40312</td>
</tr>
</tbody>
</table>
Table 6.5 shows the average moments of the AED as calculated with the Monte Carlo program for two different x-ray beam qualities, 70 kVp and 90 kVp (2 mm Al). The average noise equivalent absorption ($\bar{A}_n$), energy absorption efficiency ($\bar{A}_E$) and Swank factor ($\bar{I}_x$) obtained from the average moments are given in the same table. These data were compared with the equivalent monoenergetic values corresponding to the mean energy of the incident x-ray beam (38 keV and 45 keV) and it was found that quantum efficiency ($\eta_Q = M_0$) and noise equivalent absorption can be overestimated by as much as 60% and 30%, respectively, if the monoenergetic moments are used.

Table 6.5. Averaged moments of the absorbed energy distribution for 70 kVp and 90 kVp (2 mm Al total filtration) spectra.

<table>
<thead>
<tr>
<th>$D_p$ [µm]</th>
<th>$\bar{M}_0$</th>
<th>$\bar{M}_1$</th>
<th>$\bar{M}_2$</th>
<th>$\bar{A}_n$</th>
<th>$\bar{A}_E$</th>
<th>$\bar{I}_x$</th>
</tr>
</thead>
<tbody>
<tr>
<td>70 kVp</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>70</td>
<td>0.37</td>
<td>9.84</td>
<td>334.12</td>
<td>0.29</td>
<td>0.26</td>
<td>0.79</td>
</tr>
<tr>
<td>100</td>
<td>0.43</td>
<td>13.27</td>
<td>463.37</td>
<td>0.38</td>
<td>0.35</td>
<td>0.80</td>
</tr>
<tr>
<td>130</td>
<td>0.56</td>
<td>16.23</td>
<td>579.03</td>
<td>0.45</td>
<td>0.43</td>
<td>0.81</td>
</tr>
<tr>
<td>90 kVp</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>70</td>
<td>0.33</td>
<td>9.61</td>
<td>367.05</td>
<td>0.25</td>
<td>0.21</td>
<td>0.77</td>
</tr>
<tr>
<td>100</td>
<td>0.42</td>
<td>13.13</td>
<td>517.43</td>
<td>0.33</td>
<td>0.29</td>
<td>0.78</td>
</tr>
<tr>
<td>130</td>
<td>0.51</td>
<td>16.24</td>
<td>656.51</td>
<td>0.40</td>
<td>0.36</td>
<td>0.79</td>
</tr>
</tbody>
</table>

$\bar{A}_n$, which represents the fraction of photons contributing to the signal, is seen to vary between 29% and 45%, while $\bar{M}_0$ varies between 37% and 56% for the 70 kVp beam. As it would be expected, quantum efficiency is slightly lower at 90 kVp. The statistical factor, on the other hand, does not change considerably for the phosphor thicknesses and beam qualities used in this investigation, being of the order of 0.79 in all cases.

A validation of the Monte Carlo results was made by comparing the fractional energy absorption efficiency ($A_f$) with values calculated using a semi-empirical formula derived by Rowlands and Taylor (1983). Figure 6.15 shows a plot of $A_f$ as a function of the incident x-ray energy for a 100 µm CsI screen. The difference between both curves is due to the fact that the analytical model used by Swank (1973b), on which Rowlands and Taylor’s results are based, does not include the effect of Compton interactions in the AED. This effect is more noticeable just below the K-edge and at
high energies, where the relative contribution of Compton interactions is more important. Similar results have been obtained by Chan and Doi (1984).

![Fractional energy absorption efficiency for a 100 μm CsI screen as a function of energy. The dashed line was calculated using a semi-empirical formula proposed by Rowlands and Taylor (1983), and the circles are the result of the Monte Carlo calculation.](image)

**Figure 6.15.** Fractional energy absorption efficiency for a 100 μm CsI screen as a function of energy. The dashed line was calculated using a semi-empirical formula proposed by Rowlands and Taylor (1983), and the circles are the result of the Monte Carlo calculation.

### 6.3.2. Quantum accounting diagrams

Table 6.6 lists the six stages associated to the x-ray detection process with the fibre-optic coupled system, along with appropriate values for the number of quanta per pixel for three different CCDs. These values correspond to images obtained using the standard exposure conditions (§5.3.1), with an entrance air kerma at the plane of the detector of the order of 243±26 μGy.

The incident x-ray photon fluence per pixel was calculated using the method described in Appendix 1 and the measured entrance air kerma. Stage B represents the number of detected photons contributing to the signal, that is, the number of noise equivalent quanta, which was obtained using the results of the Monte Carlo simulation presented in §6.3.1. This number is given by the product of the average quantum efficiency, the Swank factor and the incident photon fluence per pixel. The mean optical gain of the system was calculated using mean light yield values for CsI screens reported in the literature (see §4.5.1). The fraction transmitted was calculated in terms of the measured output signal (given in ADU), the calibration factors for the conversion of signal charge per pixel to AD units obtained in §5.2.4, and the effective
quantum efficiency of the CCDs for the optical emission of the phosphor (i.e. the spectral matching), calculated as described in §4.5.4. The corresponding optical coupling efficiencies were 3% for FT-1, 4% for FT-2 and 22% for FO-2.

Table 6.6. Estimated number of quanta per pixel at different stages in the fibre-optic coupled devices for a typical exposure at 70 kVp, 4 mAs, 51.5 FDD.

<table>
<thead>
<tr>
<th>Stage</th>
<th>Description</th>
<th>FT-1</th>
<th>FT-2</th>
<th>FO-2</th>
</tr>
</thead>
<tbody>
<tr>
<td>A.</td>
<td>Incident x-ray flux per pixel</td>
<td>5016</td>
<td>5016</td>
<td>1960</td>
</tr>
<tr>
<td>B.</td>
<td>Number detected on CsI screen</td>
<td>1466</td>
<td>1755</td>
<td>882</td>
</tr>
<tr>
<td>C.</td>
<td>Optical gain</td>
<td>2.90×10^6</td>
<td>3.47×10^6</td>
<td>1.74×10^6</td>
</tr>
<tr>
<td>D.</td>
<td>Optical coupling</td>
<td>8.40×10^4</td>
<td>13.2×10^4</td>
<td>38.3×10^4</td>
</tr>
<tr>
<td>E.</td>
<td>Number detected in CCD</td>
<td>1.85×10^4</td>
<td>2.90×10^4</td>
<td>15.0×10^4</td>
</tr>
<tr>
<td>F.</td>
<td>Digital output [ADU]</td>
<td>45</td>
<td>72</td>
<td>146</td>
</tr>
</tbody>
</table>

The number of quanta per pixel can be plotted in terms of the different stages in which the system can be subdivided for the purpose of studying the transfer of signal and noise. These plots are normally referred to as quantum accounting diagrams (QAD) or nomograms, and are useful to determine the steps during the process of image formation that mostly affect the final image characteristics. The step for which the number of quanta is minimum (i.e. the one that introduces the largest statistical uncertainty) is called the quantum sink.

Figure 6.16a shows the QADs corresponding to the values presented previously in table 6.6, normalised with respect to the entrance air kerma. It can be observed that the system is quantum limited, since the quantum sink occurs at the x-ray detection stage. The effect of the optical coupling efficiency can be seen between stages C and D, where the curves for FT-1 and FT-2, which have much lower coupling efficiency, cross that of FO-2. If the digital output of the camera-frame grabber combination is also taken into consideration (stage F) then there is clearly a secondary quantum sink at this stage, which can make it difficult to achieve quantum limitation.

The analysis presented so far is only valid for zero spatial frequency. Recent work (Westmore and Cunningham 1993, Cunningham et al 1994) has shown that it is necessary to include the effect of the MTF into the QAD, since quantum limitation can be restricted at higher spatial frequencies by the spatial transfer characteristics of the system. Figure 6.16b shows the spatial-frequency dependent QAD for device FO-
2, in which the effect of the MTF is demonstrated for three different spatial frequencies. It can be observed that the quantum limitation of the system quickly disappears at higher frequencies due to the blurring effect of the CsI screen. At $f = 5 \text{ mm}^{-1}$ the number of quanta per pixel in stage E is already of the same order of magnitude as that of stage B, which means that at higher frequencies the system noise is no longer dominated by the quantum fluctuations of the incident x-ray fluence, but by the statistics set by the detection of the visible photons in the CCD.

![Quantum accounting diagrams of all three fibre-optic coupled CCDs.](image)

- **Figure 6.16.** (a) Quantum accounting diagrams of all three fibre-optic coupled CCDs. (b) Frequency dependent QAD of device FO-2. The effect of the MTF on the effective number of quanta per pixel is shown at three different spatial frequencies ($0$, $2$ and $5 \text{ mm}^{-1}$).

### 6.3.3. Frequency dependent $DQE$

The spatial frequency dependent $DQE$ of all three fibre-optic coupled devices is shown in figure 6.17 as calculated with equation 6.5 using the MTF data obtained in §6.1 and the performance parameters described in the two previous sections. A comparison of the $DQE$ of devices FO-2 and FT-2 is shown in figure 6.17a. These two devices have similar MTF characteristics, especially at higher spatial frequencies, but different coupling efficiencies (22% and 4%) and optical quantum efficiencies (39% and 22%). Figure 6.17b compares the $DQE$ of devices FT-1 and FT-2, which have approximately the same coupling efficiency, but different MTFs.

Equation 6.5 indicates that the zero frequency $DQE$ is given by the product of the quantum efficiency and the statistical factor. Therefore, the difference in $DQE$ at low
Spatial frequencies is determined by the absorption characteristics of the CsI screen. Consequently, devices with thicker coatings have better low frequency $DQE$ performance. At higher spatial frequencies the spatial transfer characteristics of the system determine the overall shape of the $DQE$. However, it can be seen in figure 6.17a that higher coupling and optical quantum efficiencies have a greater effect in improving $DQE$ at higher frequencies than the increase observed in figure 6.17b, which is due to better MTF performance.

![Figure 6.17. Comparison of the frequency dependent $DQE$ of the coupled devices.](image)

(a) Similar MTF and different coupling efficiency (FO-2 and FT-2). (b) Similar coupling efficiency and different MTF (FT-1 and FT-2).

The difference between the noise performance characteristics and the modulation in the transfer of signal is best seen in figure 6.18a, which compares the square of the MTF of device DD-3 with the noise Wiener spectrum for the same device normalised at zero spatial frequency. It can be observed that the spectral components of $W_s$ are much higher over the whole spectrum. While $MTF^2$ is practically zero at $f = 10 \text{ mm}^{-1}$, $W_s$ decreases by only 30% over the same frequency range. Therefore, the shape of $DQE(f)$ would be essentially determined by $MTF^2$.

The solid line in figure 6.18b shows a plot of $DQE(f)$ as obtained from the measured values of the MTF and the NPS. The Wiener spectrum was obtained from an image taken under standard exposure conditions; the entrance photon fluence in this case was $\Phi_e = 1102 \pm 118$ photons per pixel. The mean output signal and the standard deviation, obtained directly from the digital image, were $\langle S \rangle = 39.29$ and $\sigma_s = 9.50$ ADU. Therefore $SNR_{out} = 4.13$ and, correspondingly, the zero-frequency $DQE$ would be...
be 1.55%. The dotted line in the same figure was obtained by renormalising $MTF^2$ using the value for the zero-frequency $DQE$, and is shown for comparison.

![Figure 6.18.](image)

**Figure 6.18.** (a) Comparison of $MTF^2$ and the normalised NPS of device DD-3. (b) Frequency dependent $DQE$ obtained with the data shown in figure a).

Using the same formalism previously employed with the coated devices in §6.3.1, the quantum efficiency and the noise equivalent absorption of DD-3 were calculated in terms of the moments of the AED for a 70 μm Epi CCD and a 70 kVp beam. Although in this case there are no fluctuations in the energy signal due to K-escape fluorescence, a significant variation in energy deposition due to the relative contribution of Compton interactions in Si for this energy range would be expected (see §4.4.3). The average quantum efficiency was found to be 1.9%, and the statistical factor was 0.82, giving a noise equivalent absorption of 1.56%, which agrees very well with the measured value.

Thus, despite having a much better spatial resolution, the $DQE$ performance of DD-3 is much lower than that of any of the fibre-optic coupled devices due, essentially, to its lower quantum efficiency and very poor noise characteristics. However, it would still compare favourably with a silicon photodiode array used for micro-tomography, reported recently by Reid and Cunningham (1993), which has a zero-frequency $DQE$ of only 0.4% under similar exposure conditions.
6.4. Contrast-Detail analysis

6.4.1. Method of measurement

The contrast-detail performance of the CCD system was determined from images of the Leeds TO-10 phantom, which has already been described in Chapter 3. Due to the small size of the CCDs used for this assessment in most cases it was not possible to image more than a few details of the phantom in a single frame. A simple scanning method was devised in order to overcome this problem. First, the CCD camera was carefully aligned with the central axis of the x-ray beam. A raster scan of the phantom was then made by moving it in discrete steps, in such a way that the area to be imaged lay directly above of the CCD. Figure 6.19 shows a schematic diagram of the experimental setup for the contrast-detail measurements.

A continuous scan of the total phantom area (~ 1.1×10^4 mm²) would require more than 350 images with the smaller devices. For this reason a selective scan of individual details was done. Objects up to ~4 mm diameter (row D) could be imaged within a single CCD frame using this method and, therefore, the whole of the...
interesting range of detail sizes could be covered. Since the actual physical
distribution of details within the phantom (see figure 3.1, Chapter 3) makes it difficult
to scan the groups with diameters larger than 1.4 mm, it was decided to restrict the
scanning area to the central section of TO.10, which includes rows G to M only, with
detail diameters in the range 1.4 mm down to 0.25 mm. Depending on the size of the
CCD and the detail size to be observed, up to three objects could fit within a single
image, thereby considerably reducing the total number of images needed for one
study. Although strictly speaking this method of assessment does not conform to the
Leeds protocol, it would still provide some information about the ability of the system
to detect simple objects of known diameter and contrast.

6.4.1.1. The scanning system

A computer-controlled scanning system was used to move the phantom with respect
to the CCD camera. The scanning system consisted of two translation units that could
run along perpendicular directions (see figure 6.20); each unit was connected to a
stepper motor. The phantom was mounted on a bracket that was in turn fixed to one of
the translation units. The stepper motors were controlled by a Digiplan electronic
motion control system, which consisted of a pre-wired rack fitted with an RS-232
(serial) interface and two independent stepping motor drivers, one for each axis.

![Figure 6.20. Top view of the scanning system showing the translation units, the phantom and the CCD camera.](image-url)
The stepper motors had an angular resolution of 400 steps per revolution, and the translation units had a resolution of 1 mm per revolution, which means that the phantom could be moved in 2.5 µm steps in either direction.

A map of the relative positions of the details was obtained from a conventional film-screen radiograph of TO.10. These data were used to create a look-up table for the scanning program. The radiograph was used as a mask placed on top of TO.10 in order to locate a reference detail (G1) to be lined up with the x-ray tube and the camera. Fine adjustments to the alignment were made by taking images of G1 and moving the phantom until the object was exactly at the centre of the image. Once the reference detail was precisely localised all other details were quickly found using the look-up table.

6.4.1.2. Exposure conditions

Three different beam qualities were used for the contrast-detail measurements. THX and ABDO filtrations at 70 kVp were used in order to compare the results of the evaluation with the film-screen data presented in Chapter 3. However, these two filtrations produce x-ray spectra that do not favour the energy response of the detector (particularly the deep depletion device). Therefore, a 90 kVp beam at 0.08 mAs with no additional filtration was also used to test the performance of the system at lower exposures.

From the point of view of detector response and sensitivity it is the incident spectrum at the detector plane (i.e. the exit spectrum from the phantom) that determines the final signal-to-noise ratio properties of the image. Table 6.7 shows the measured air kerma at the detector plane for a 70 kVp spectrum at 51.5 cm FDD, using different filtrations and including the effect of TO.10. The corresponding x-ray fluence for each filtration was calculated using the method described in Appendix 1.

It can be observed in table 6.7 that the air kerma was reduced by 2 orders of magnitude when ABDO filtration was used, as compared to the use of TO.10 alone. Also, the photon fluence per pixel for ABDO was about 4 times smaller than for THX. Since the maximum sensitivity of the system varied between 2.31 and 9.13 ADU/µGy, depending on the CCD used, only the most sensitive device (FO-2) was able to work at such low exposure levels.
Table 6.7. Measured air kerma and corresponding x-ray fluence at the entrance plane of the detector for a 70 kVp spectrum at 51.5 cm FDD, 4 mAs.

<table>
<thead>
<tr>
<th>Added filtration (+TO.10)</th>
<th>Air kerma [μGy]</th>
<th>$\Phi_s$ [mm$^2$]</th>
<th>$\Phi_x$ [photons/pixel]</th>
</tr>
</thead>
<tbody>
<tr>
<td>None</td>
<td>199</td>
<td>$3.32 \times 10^6$</td>
<td>904</td>
</tr>
<tr>
<td>0.1 mm Cu</td>
<td>90</td>
<td>$2.05 \times 10^6$</td>
<td>558</td>
</tr>
<tr>
<td>THX</td>
<td>14</td>
<td>$3.59 \times 10^5$</td>
<td>98</td>
</tr>
<tr>
<td>ABDO</td>
<td>2.7</td>
<td>$9.0 \times 10^4$</td>
<td>25</td>
</tr>
</tbody>
</table>

The measured entrance air kerma for the 90 kVp spectrum was 11.5 μGy, and the corresponding x-ray fluence was 63, 111, and 285 photons per pixel for DD, FO and FT devices, respectively. In terms of exposure and photon fluence these figures lie between those of THX and ABDO. However, this spectrum has a higher proportion of low energy components, which favours the response of the system.

6.4.2. Results and discussion

6.4.2.1. Evaluation of the images

Once an image was captured it was stored on the PC hard disk until a complete scan was finished. All the images corresponding to one scan were then transferred via Ethernet to a Sun workstation for further processing and analysis. After correction for fixed pattern noise several individual images could be joined to form a mosaic of a whole section of the phantom, or were simply displayed one by one to determine the threshold contrast.

Figure 6.21 shows two sample images of different sections of TO.10 taken with device FT-2 at 70 kVp without any additional filtration. The entrance air kerma to the phantom in this case was ~240 μGy. Figure 6.21a shows the first three details of rows G and H (1.4 and 1.0 mm diameter, respectively) and figure 6.21b shows the first four details of rows L and M (0.35 and 0.25 mm diameter). At this level of exposure this device was able to resolve practically all details in rows G to M.

The analysis of the images was made either by two observers, with only one independent reading per observer, or by a single observer, with two independent readings. In all cases the observers had some previous experience with the use of
TO.10. The statistical error associated with the threshold contrasts determined in this way was 11% for a two-observer experiment, and 16% for a one-observer experiment, following the work of Marshall et al (1992).

Figure 6.21. Images of different sections of TO.10 taken with device FT-2. (a) Details G1 to H3, 1.4 and 1.0 mm diameter. (b) Details L1 to M4, 0.35 and 0.25 mm diameter.

6.4.2.2. Contrast-Detail diagrams

A summary of all the contrast-detail measurements made with the CCD system is shown in table 6.8. Given the large number of CCD configurations and exposure conditions used only a few representative cases were considered. The entrance air kerma was varied by changing the focal spot-detector distance and it was measured at the entrance surface of the phantom. All the threshold contrast values were corrected for changes in the x-ray beam quality using the method described in Chapter 3 (§3.1).

Table 6.8. Exposure conditions for all the contrast-detail scans. The air kerma was measured at the entrance plane of the phantom.

<table>
<thead>
<tr>
<th>Device</th>
<th>Filtration</th>
<th>kVp</th>
<th>Air kerma [μGy]</th>
<th>Cm range [%]</th>
</tr>
</thead>
<tbody>
<tr>
<td>FO-2</td>
<td>ABDO</td>
<td>70</td>
<td>112</td>
<td>10.4 - 50.0</td>
</tr>
<tr>
<td></td>
<td>ABDO</td>
<td>70</td>
<td>221</td>
<td>6.6 - 35.0</td>
</tr>
<tr>
<td>FT-1</td>
<td>THX</td>
<td>70</td>
<td>112</td>
<td>12.0 - 51.0</td>
</tr>
<tr>
<td>FT-2</td>
<td>THX</td>
<td>70</td>
<td>113</td>
<td>7.8 - 42.6</td>
</tr>
<tr>
<td>FO-2</td>
<td>None</td>
<td>90</td>
<td>4.2</td>
<td>9.9 - 55.6</td>
</tr>
<tr>
<td></td>
<td>None</td>
<td>90</td>
<td>8.3</td>
<td>8.4 - 33.9</td>
</tr>
<tr>
<td></td>
<td>None</td>
<td>90</td>
<td>17.1</td>
<td>7.2 - 27.0</td>
</tr>
<tr>
<td>DD-3</td>
<td>None</td>
<td>90</td>
<td>17.3</td>
<td>27.0 - 94.7</td>
</tr>
<tr>
<td></td>
<td>None</td>
<td>90</td>
<td>33.3</td>
<td>16.8 - 79.1</td>
</tr>
</tbody>
</table>
The results of the evaluation are presented in the standard way of plotting the threshold contrast as a function of detail diameter for different levels of exposure. Figure 6.22a shows the results of the threshold contrast evaluation for device FO-2 using ABDO filtration at 112 and 221 μGy entrance air kerma. The film-screen data set fs5 (see §3.1.3.2), which was taken at 109 μGy, is also shown in the same graph for comparison. It can be observed that the threshold contrast of the CCD system varies as the inverse of the detail diameter, as it would be expected for a noise limited system. An increase in the incident exposure to FO-2 by a factor of two reduces $C_n$ by a factor of $\sqrt{2}$, which is also an indication of quantum limitation.

![Graph](image)

**Figure 6.22.** Comparison of Contrast-Detail performance between the CCD system and film-screen. (a) device FO-2, ABDO filtration. (b) devices FT-1 and FT-2, THX filtration.

Figure 6.22b shows the contrast-detail plots for devices FT-1 and FT-2 obtained at the same entrance air kerma (~112 μGy) using THX filtration, and compares them with the film-screen data set fs1, which was taken at 51 μGy. The threshold contrasts obtained with device FT-2 are, on average, 1.4 times lower than those obtained with FT-1. In §6.2.2 it was shown that these two devices have very similar noise characteristics and, from §5.3.2, it is also known that the sensitivity of FT-2 is higher by a factor ~1.6. Therefore, the difference in the contrast-detail performance of these two devices approximately scales with sensitivity.

The film-screen system shows better contrast-detectability properties than the CCD system for both filtrations used and for all the detail sizes considered. However, the
threshold contrast of film-screen degrades more rapidly for small details. For 0.25 mm
diameter objects the fibre-optic coupled CCDs have nearly equivalent performance to
the film-screen system, although an entrance exposure about twice as high is required.
The dashed curves in figures 6.22a and 6.22b were obtained with equation 1.4, and
correspond to an effective sampling aperture (i.e. the total size of the unsharpness, \( a_u \))
of 0.39 mm and 0.28 mm for fs5 and fs1, respectively. These are considerably higher
than the limiting resolutions of FT-1 and FT-2 (see §6.1.3), which are 0.05 mm and
0.06 mm, respectively.

\[ C_{th} = \frac{K}{E} \]

\( K \) is a constant, \( E \) is the entrance air kerma, and \( C_{th} \) is the threshold contrast.

\[ C_{th} \propto \frac{1}{E} \]

\( C_{th} \) decreases as the entrance air kerma increases. A comparison of the TCDD performance between devices FO-2 and DD-3 using a 90 kVp beam and TO.10 without any additional filtration is shown in figure 6.23. The threshold contrast data for the deep depletion device has similar characteristics to the contrast detail curves obtained with the coated devices, which indicates that contrast
detail performance is also noise limited. As it would be expected from the resolution
properties of these devices, which were determined in §6.1, both CCDs were able to
detect objects down to 0.25 mm. It can be observed that the coated device achieves
lower threshold contrasts at lower entrance air kerma. At the same level of exposure
(\(-17 \mu Gy\) \( C_{th} \) is approximately four times lower for FO-2. This difference can be
explained once again in terms of the difference in sensitivity between FO-2 and DD-2,
which is also a factor of 4.
6.5. Summary and conclusions

An evaluation of the imaging performance of a CCD-based x-ray imaging system has been presented in this chapter. The spatial resolution of the system was studied using modulation transfer function analysis. The results have shown that spatial resolution was mainly determined by secondary processes in the imaging chain. In fibre-optic coupled devices the phosphor screen determined the resolution of the whole system, while in deep depletion devices the absorption and diffusion in the undepleted silicon reduced the maximum attainable resolution. The blurring effect caused by these secondary processes totally dominated the spatial frequency modulation, reducing the MTF to values between 10% and 15% at spatial frequencies of less than half the Nyquist limit. This means that the geometrical limit imposed by the finite pixel size of the CCDs did not restrict significantly the spatial transfer characteristics of the system and that aliasing effects were unlikely to occur.

The spatial resolution limit of the fibre-optic coupled devices varied between 8.0 and 9.7 lp/mm, depending on the phosphor thickness and coupling method (taper or faceplate). These figures were compared with the spatial resolution of modern film-screen combinations and found to be generally better than that of a general purpose film-screen system (8.0 lp/mm), but much lower than that of a high resolution film-screen system (12.5 lp/mm). The spatial resolution of the deep depletion device was 14.7 lp/mm, which is higher than that of any of the two film-screen systems.

Wiener spectral analysis was used to investigate the effect of the relative contribution of different noise sources on the magnitude and texture of the total system noise. The noise components considered were structure mottle (fixed pattern noise), quantum noise, and dark noise. The effects of some image processing operations such as image subtraction and averaging on the measured Wiener spectrum were also investigated.

The results of the noise analysis have shown that both the magnitude and the spatial distribution of noise are strongly dependent on the detection mode. The dominant source of noise in fibre-optic coupled devices was the structure mottle associated with the phosphor screens. After correction for fixed pattern noise this factor was practically eliminated, and the images contained only the stochastic contributions from quantum, electronic and thermal noise. On the other hand, the total system noise in deep depletion devices was dominated at all spatial frequencies by the random
fluctuations associated with detection process, with only a small contribution from fixed pattern noise due to small variations in responsivity from pixel to pixel.

The spectrum of quantum and electronic noise per frame of fibre-optic devices was observed to be very similar for the three CsI screens tested. At low spatial frequencies the spectrum was dominated by the quantum fluctuations corresponding to the detection of individual x-ray photons in the CsI screen. At higher frequencies the relative contribution of quantum noise slowly decreased due to the spatial transfer properties of the system, until electronic and thermal noise became the dominant contribution.

Dark noise, which includes only electronic and thermal noise, was observed to be independent of spatial frequency. However, the magnitude of dark noise was dependent on the CCD architecture. Typical values were 128, 425 and 735 noise equivalent electrons for anti-bloomed, non anti-bloomed and deep depletion devices, respectively. A comparison of these numbers with the values for thermal noise obtained in §5.4.2 indicates that electronic noise was the dominant contribution to dark noise.

The MTFs of the fibre-optic coupled devices were used to calculate the frequency dependent $DQE$ of the system using a model for multi-stage imaging systems. For the deep depletion device the $DQE$ was obtained directly from the MTF and the noise power spectrum. A Monte Carlo program was used to simulate the x-ray interaction processes in each detector in order to determine the noise equivalent absorption. The results of the Monte Carlo program were compared with a direct calculation of the zero-frequency $DQE$ of the deep depletion device and were found to be in very good agreement. The results have shown that the $DQE$ of the fibre-optic coupled devices was at least 20 times higher than that of the deep depletion device. The excellent MTF performance of the deep depletion CCD were totally offset in terms of $DQE$, given its poor absorption efficiency and high noise characteristics.

The threshold contrast of the system varied as the inverse of the detail diameter in all the experiments, which indicates that performance of the CCD system was noise-limited. Although the dose performance of the film-screen system was observed to be generally better for the exposure conditions used, the CCD prototype was not optimised in any respect.
Chapter 7

Conclusions and future work

The results of an evaluation of the imaging performance of two x-ray imaging detectors that could be suitable for the development of new digital radiographic systems have been presented in this thesis. An experimental procedure for the assessment of the most important imaging characteristics of the x-ray detectors was set up. The evaluation process included subjective and objective measurements of image quality and took into account the effect of various detector parameters upon the imaging performance of the digital radiographic systems.

The systems investigated have some basic characteristics in common, such as being based in position sensitive detectors and producing digital output in real time. However, there are also some fundamental differences in terms of format and mode of operation. The SDRD is based on a photon counting detector and uses a scanned projection irradiation geometry, while the CCD system works in energy integration mode and uses an area exposure technique. Also, the stage of development of these devices as x-ray imaging detectors is very different. The SDRD has been developed over the last 10 years and has already undergone clinical trials. The CCD system, on the other hand, was not build as a clinical system, but only as a prototype that could lay the foundations upon which a clinical system could be developed. This means that, although the evaluation of both systems was made using the same basic techniques, a very different approach to data collection and analysis had to be taken.

7.1. Evaluation of the SDRD

The evaluation of the imaging performance of the SDRD was mainly based on threshold-contrast detail-detectability studies. Since the detector works in photon counting mode and has very good scatter rejection capabilities, particular emphasis was put in the dose performance of the system as compared to a modern film-screen combination. The effect of different geometrical parameters and modes of operation of the multiwire chamber on the detectability of simple objects was also studied.
The results of the evaluation have shown that the high quantum efficiency of the chamber combined with high counting rate capability and practically zero background permit the system to work at very low exposure levels. The contrast sensitivity varied as the inverse of the square root of dose, which translates in a very wide dynamic range. The contrast-detectability of the system was generally better than that of the film-screen combination for objects bigger than the pixel size.

The main characteristic of the SDRD is its ability to produce clinical images at very low exposures due to its wide dynamic range and linear response. Its main disadvantage is that it has limited spatial resolution. Taking these factors into account the areas in which this device could be useful are:

i) Examinations where low resolution information conveys the pathology

ii) Follow-up assessment of known pathology

The system has proved to be useful in pelvimetry and in examinations of the abdomen using contrast media. Other possible areas in which it could be used include intravenous urography and barium studies, facio-maxillary reconstruction planning and osteoporosis assessment.

The evaluation of the SDRD has shown that it is possible to use counting x-ray detectors in digital radiography. Further work has been initiated in the improvement of present chambers and in the development of new systems using gaseous micro-strip chambers (Shekhtman 1993), which could offer better spatial resolution while maintaining high quantum efficiency and count rate capability.

Electrostatic and mechanical constraints in the design of the multiwire chamber limit the minimum anode wire separation to \(-1\) mm. Therefore no further improvements in the spatial resolution of the system with respect to what has already been achieved using the method of separate counting of coincidence events would be expected. However, improvements in the count rate capability can be achieved by reducing the anode-cathode distance. A system prototype with the anode plane separated only 1 mm from the cathode plane has been recently build and is currently undergoing initial tests. It is expected that in this system the difference in counting rate efficiency between anti-coincidence and coincidence channels will be reduced, thus improving the noise performance of the system. A second modification of the chamber geometry, which consists in increasing the anode wire length to 10 cm, has been planned for the near future. This would enhance the quantum efficiency of the chamber and would
help to maintain the low dose performance of the system. The evaluation of such systems could be the subject of future work.

7.2. Evaluation of the CCD system

The evaluation of the imaging properties of x-ray sensitive CCDs showed that the spatial transfer and noise characteristics are strongly dependent on the x-ray detection mode. In general terms both modalities of x-ray detection could offer good spatial resolution, comparable to that modern film-screen technology. MTF measurements showed that secondary processes determined the spatial-frequency modulation of the system. In fibre-optic coupled devices the geometrical resolution limit imposed by the finite sampling aperture of the CCD pixels did not match the resolution properties of the scintillator. Therefore, if the full resolution of the CCDs is to be used then the MTF of the screens would have to be improved. This, of course, would require a task dependent optimisation of the system in order maximise the detail signal-to-noise ratio while maintaining adequate resolution.

The full-well capacity has to be considered when choosing an appropriate CCD and designing the optical coupling characteristics of the system. The saturation levels of typical CCD image arrays vary normally between $10^5$ and $10^6$ electrons per pixel, and can be a limiting factor in the detector design. For a given optical coupling efficiency the adequate full-well capacity would depend on many other parameters related to the particular imaging task and exposure conditions.

The size and format of these detectors is appropriate for applications that require a small field of view. The application of CCDs to digital mammography is being pursued very actively by several groups around the world, and some prototypes have already been reported in the literature (Maidement 1993a, Roehrig et al 1993). Recent theoretical calculations of dynamic range requirements for digital mammography (Maidement et al 1993b) have shown that even for a moderately low coupling efficiency of 16 electrons per pixel per x-ray absorbed, a dynamic range of 3000 would decrease the number of grey levels that could be resolved by only 2% compared to a detector with ideal coupling and no inherent noise. The experimental results obtained in this thesis indicate that such values for the coupling efficiency can be attained using fibre-optically coupled CCDs, since it varied between 12 to 170 electrons per x-ray absorbed. If the inherent detector noise is of the order of 200
electrons per pixel then a maximum full-well capacity of $6 \times 10^5$ would be needed in order to obtain the necessary dynamic range. The CCDs used during this investigation had full well capacities between $1 \times 10^5$ and $3 \times 10^5$, which means that either the inherent detector noise should be decreased to levels of the order of 50 electrons per pixel, or CCDs with higher full well capacity should be used.

An alternative application of x-ray sensitive CCDs would be in studies where knowledge of the spatial distribution of radiation intensity would be of interest. Work is currently in progress to use the system to obtain detailed measurements of radiation intensity distribution of focal spots and of collimator alignment in x-ray diffraction studies. The traditional method used to measure the focal spot distribution is to obtain an image using a pinhole and radiographic emulsion. However, quantitative information can only be obtained from these images if the exposure and development are carefully controlled. The CCD camera described in this thesis could provide an accurate method of measuring the radiation intensity distribution of the focal spot in real time. Figure 7.1a shows an image of the focal spot obtained with a 30 μm pinhole and a deep depletion device. A surface plot of the same focal spot as obtained with a fibre-optic coupled device is also shown in figure 7.1b.

X-ray tubes that use micro-focal spot technology capable of working at up to 30 kVp and 0.2 mA with a focal spot of the order of 5 μm are readily available. This could open the possibility of using x-ray sensitive CCDs for non-destructive testing. Deep depletion devices could be used as very high resolution detectors for the study of small objects using computed microtomography. However, the low quantum
efficiency of these devices at x-ray energies above 20 keV would require large doses and/or long acquisition times, which may limit their usefulness to the imaging of *in-vitro* specimens. Some interest has recently been expressed in the development of such systems for the inspection of breast lumpectomy samples (Machin and Webb 1994). Another possible application would be in bone densitometry, where it would be useful to image the structure of trabecular bone. Moreover, *in-vivo* measurements of trabecular bone structure could possibly be made over small areas of the body, such as the wrist, using coated devices with a high resolution phosphor screen without incurring in severe radiation dose to the patient.
Appendix 1

Determination of the incident x-ray photon flux

The incident x-ray photon fluence to the detector can be determined from the entrance air kerma measured at the detector plane. The total air kerma ($K_{air}$) for a broad beam x-ray energy distribution can be calculated by adding the individual contributions from the energy components of the incident x-ray spectrum, that is

$$K_{air} = \int_{0}^{E} c_k(E) \Phi_x(E) dE \quad (A1.1)$$

where $\Phi_x(E)$ represents the differential photon fluence distribution and $c_k(E)$ is the photon fluence to kerma in air conversion factor at energy $E$ (Johns and Cunningham 1983). Tabulated values of $c_k(E)$ have been given by Birch et al. (1979). Figure A1.1 shows a plot of $c_k^{-1}(E)$ in the energy range from 0 to 100 keV.

![Figure A1.1](image.png)

**Figure A1.1.** Air kerma to photon fluence conversion factors as a function of x-ray energy, from Birch et al. (1979).

The x-ray energy distribution can be expressed as the product of the total number of incident photons per unit area and the normalised x-ray energy distribution, that is
\[
\Phi_x(E) = \Phi_x \hat{\Phi}_x(E) \quad (A1.2)
\]

where
\[
\int_0^\infty \hat{\Phi}_x(E) \, dE = 1 \quad (A1.3)
\]

Using equations A1.1 and A1.2 the incident x-ray fluence can be calculated in terms of the total measured kerma in air and the normalised x-ray energy distribution

\[
\Phi_x = \frac{K_{air}}{\int_0^\infty c_x(E) \hat{\Phi}_x(E) \, dE} \quad (A1.4)
\]

For example, the measured entrance air kerma at the detector plane for the typical exposure conditions used in the evaluation of the CCD system (70 kVp, 4 mAs and 51.5 cm focal spot-detector distance) was \( K_{air} = 243 \pm 26 \, \mu \text{Gy} \). Using the spectral distribution for a 70 kVp spectrum with 2 mm Al total filtration (Birch et al. 1979) the air kerma per photon per unit area is \( 6 \times 10^{-5} \, \mu \text{Gy}/(\text{photon/mm}^2) \). Substituting these values into equation A1.4 the incident photon fluence to the detector is \( \Phi_x = (4.05 \pm 0.43) \times 10^6 \, \text{photons/mm}^2 \). For a 22 \, \mu\text{m} pixel size this would be equivalent to an incident x-ray fluence of approximately 1960 \pm 208 photons per pixel.
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