Optimization Of Portal Imaging Systems

Jean-pierre Bissonnette

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Optimization of Portal Imaging Systems

by

Jean-Pierre Bissonnette

Department of Medical Biophysics

Submitted in partial fulfillment
of the requirements for the degree of
Doctor of Philosophy

Faculty of Graduate Studies
The University of Western Ontario
London, Ontario
September 1996

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ABSTRACT

Many portal imaging devices have been developed to verify the geometric accuracy of radiation therapy treatments. Portal imaging devices are used to take images of the patient during radiation therapy treatments. These images are used to detect patient positioning errors which may jeopardize the outcome of conventional and high-precision radiotherapy treatments. Unfortunately, the quality of portal images obtained with such devices is disappointing, resulting in sparse clinical use of these devices.

Researchers have been substituting various imaging components on these portal imaging systems in the hopes of optimizing portal image quality. This empirical approach has led to some successes. However, choosing imaging system components on the basis of one desirable parameter while ignoring the impact of the change on overall system performance wastes time, money, and effort. Clearly, a more efficient approach is required.

This thesis presents approaches for the optimization of both the design and use of portal imaging devices. These approaches require understanding of the fundamental physics of portal imaging, such as the size and shape of the x-ray sources of medical linear accelerators and the interaction of x-rays within typical portal imaging detectors. The use of existing portal imaging systems (i.e., portal films and video-based systems) has been optimized by finding the radiographic magnification which provides the best image quality for a particular system/linear accelerator combination. It has been found that, for portal films, radiographic magnification is undesirable. On the other hand, a radiographic magnification of 1.5-1.7 is optimal for video-based systems. Therefore, the image quality from an existing imaging system can be improved without
changing the system design. The design of portal imaging systems has been optimized using a theoretical approach known as quantum accounting diagram (QAD) theory. Using this theory, a detailed analysis of a video-based portal imaging system has permitted the theoretical derivation of the detective quantum efficiency (DQE) of the imaging system. The analysis has shown that the video-based portal imaging system suffers from severe secondary quantum sinks for non-zero spatial frequencies, resulting from sub-optimal system design. Furthermore, the theoretical DQEs have been compared with experimental measurements - the first experimental verification of the QAD theory. The QAD theory has been expanded to include the physical parameters involved with the human visual system and allow the computation of indices of perceived image quality. This approach enables the optimization of imaging devices using a single figure of merit, and has been used to optimize the phosphor screen thickness used by two different designs of portal imaging systems. We have demonstrated that the QAD approach allows us to predict the change in overall system performance for any modification in imaging system design. In future, we believe that this tool will be vital to the development and optimization of improved portal imaging systems for radiation therapy.

Keywords: Radiation therapy, medical imaging, portal imaging, MTF, NPS, DQE.
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Cancer is second only to heart disease as a cause of death and accounts for 22% of all deaths in North America.¹ In Canada, the 1996 estimates from the National Cancer Institute of Canada predict 129 200 new cases of cancer in Canada as well as 61 800 deaths resulting from cancer.² However, the increase in the average age of the population, improved cancer detection and better case reporting produce a yearly increase of the incidence of all cancers combined. In Ontario, the projected increase rate of cancer incidence is 4.5% per year, leading to a doubling of the number of cases by the year 2010.³ Thus, the radiation therapy caseload is expected to double within 15 years.

1.1 External beam radiation therapy

External beam radiation therapy, used alone or in conjunction with chemotherapy or surgery, has been proven to help increase cancer patient survival.¹ ⁴ The goal of the external beam radiation therapy is to administer high doses of radiation to a diseased volume with minimal radiation dose to surrounding healthy tissues, resulting in (i) a reduction of the number of tumor cells to a level that achieves permanent local tumor control and (ii) a higher quality of life.⁴ The volume treated with radiation is separated into three distinct volumes: the gross tumor volume (GTV), the clinical target volume (CTV), and the planned target volume (PTV).⁵ The GTV contains palpable, visible, or demonstrable malignant growth. The CTV contains the GTV and any volumes where microscopic malignant disease is suspected. The PTV is a static volume which
encloses the CTV as well as margins that must be added to compensate for organ, tumor, and patient motion in addition to uncertainties in beam and patient setup. Thus, the PTV can be considered as a three-dimensional envelope within which the CTV may move, but cannot exit. Therefore, the irradiated volume (which also involves other tissues intercepting the entire path of the treatment beam) often includes a significant amount of disease-free tissues. Naturally, higher doses of radiation result in better tumor control probability, but also in increased normal tissue damage and thus morbidity.

External beam radiation therapy is performed using machines consisting of a source of high-energy x rays or electrons mounted on a rotating gantry. This gantry can be rotated 360° around the patient such that the central axis of the treatment beam describes a plane perpendicular to the gantry axis of rotation. The intersection of the gantry axis of rotation and the plane described by the central axis of the treatment beam at all gantry angles is known as the isocentre. Placing the tumor volume at the isocentre of the treatment machine ensures that the tumor volume is always in the central axis of the treatment beam. Therefore, one can direct beams from several directions to the tumor so that tumor dose is high while minimizing the dose to the surrounding healthy tissues.

A typical external beam radiotherapy treatment occurs as follows. Following the definition of the GTV and the surrounding normal structures, radiation oncologists plan radiotherapy treatments using devices called radiotherapy simulators. These devices consist of special fluoroscopic or radiographic units that can reproduce the geometry of treatment machines, but use a diagnostic x-ray source instead of a high-energy therapy source to acquire high-quality radiographs of the patient. During simulation, radiation oncologists define the PTV, determine the gantry angles, and define the size and shape of the radia-
tion beams to be used. Once these parameters are determined, radiographs of the patient are taken, one for each of the intended treatment beams. The intended position of the radiation therapy beam is clearly indicated on these radiographs, which are called “simulator films”. These films demonstrate visually the intended relationship between the patient’s bony anatomy and the radiation beam edges. Once simulation is completed, the skin of the patient is marked to indicate the intended position of each radiation field as well as points illuminated by a set of three orthogonal laser beams whose intersection coincides with the isocentre of the simulator. An identical set of lasers serves the same purpose in the radiotherapy treatment room. In principle, aligning the skin marks with the laser beams in the treatment room should ensure proper patient positioning during daily radiotherapy treatments. In order to exploit radiobiological differences between tumorous and healthy tissues, the treatment dose is often given in many fractions, over several weeks. Unfortunately, experience has shown that skin marks are not reliable indicators of tumor position since changes in patient anatomy occur throughout the long treatment period.

1.2 Rationale for portal imaging

As stated above, the goal of external beam radiation therapy is to deliver a prescribed dose of radiation to a well-defined treatment volume. Small deviations from this goal, due to inaccuracies in dose prescription, planning, and calibration, and precision of dose delivery, may affect treatment outcome. Radiation oncologists are trained to know the appropriate dose to prescribe, within narrow limits. Dose calibration protocols are now sufficiently formalized to ensure accuracy and precision of ±2%, and computer technology has made accurate treatment planning systems common to all radiotherapy departments. However, the accuracy of daily dose delivery (i.e., positioning of the patient so
that the treatment beam irradiates only the prescribed volume) is most vulnerable to human error.

Unfortunately, inaccuracies in daily dose delivery occur frequently. In a study involving sixteen treatment sites, Byhardt et al.\textsuperscript{7} have reported a rate of occurrence of field placement errors over 5 mm ranging between 27% (pelvis) and 7% (cranium), with an overall mean of 15% (10% for errors larger than 1 cm). They have observed that translational shifts of the treatment field from the planned field are the dominant cause of errors, and that patient positioning errors range from 5 to 40 mm, with a mean of 11 mm. In another study, Rabinowitz et al.\textsuperscript{8} have measured inter-treatment and simulator-to-treatment variations for six treatment sites. They have observed a field placement error rate over 5 mm of 62% (23% for errors larger than 1 cm), with mean discrepancies between portal and simulator films ranging between 3.5 mm in head and neck fields to 9.2 mm in thoracic fields, with an overall mean discrepancy of 7.7 mm. Also, they have found an average standard deviation of 3 mm in the day-to-day position of the field margins and shielding blocks, and they have observed that simulator-to-treatment discrepancies are significantly larger than treatment-to-treatment discrepancies at most sites. In 1989, Kihlén and Rudén\textsuperscript{9} found, for ten treatment sites, average standard deviations ranging between 3 and 6 mm (with an overall average of 3.5 mm), despite using laser alignment and good patient fixation. More recently, in a study involving eight treatment sites, Herman et al. have observed that the proportion of fields requiring correction varied between 4% and 26% (for abdominal and pelvic fields, respectively), with an overall mean of 12%. When comparing this last figure with the 15% measured by Byhardt et al., it appears that the overall proportion of radiation therapy fields requiring correction remains high, despite improved clinical practices over the course of several years (i.e., 1978 to 1994).
Clinical studies have shown that the outcome of radiation therapy depends strongly on the accurate administration of the prescribed dose to the diseased volume.\textsuperscript{8,10-16} Unfortunately, many factors (e.g., patient motion, field location, size or shape, beam direction, machine output, ...) cause deviations of the actual treatment from the treatment plan.\textsuperscript{6,14} Goitein and Busse\textsuperscript{17}, Brahme,\textsuperscript{18} as well as Boyer and Schultheiss\textsuperscript{19} have determined that uncertainties in treatment delivery\textsuperscript{6} may result in under-dosage of the tumor and over-dosage of normal tissue, which can decrease considerably the tumor control probability and thus prevent radiation therapy from achieving its goal. For this reason, the overall dosimetric uncertainty should be kept below 5\%\textsuperscript{18}. Uncertainties in the calibration, dose monitoring system, and stability of medical linear accelerators already account for a dosimetric uncertainty of 4.2\%\textsuperscript{20}. To maintain the overall dosimetric accuracy within 5\%, the treatment must be executed with extremely high accuracy, thus leaving little room for errors in the position of the patient relative to the treatment beam. Unfortunately, the uncertainty in the position of the patient is the main contributor to the overall treatment uncertainty.\textsuperscript{14} Several studies have been published regarding the frequency of occurrence, magnitude, and impact of these errors.

Patient positioning errors alter treatment outcome since these errors may cause tumor underdosage and normal tissue overdosage. Suit has shown theoretically that a reduction of tumor dose can reduce the probability of patient survival since tumor underdosage reduces the probability of local tumor control.\textsuperscript{21} Clinical observations have correlated patient positioning errors with (i) the recurrence and relapse rates of Hodgkin's lymphoma,\textsuperscript{12,22,23} (ii) with the failure of prostate cancer treatment,\textsuperscript{16} (iii) worse survival for lung cancer patients,\textsuperscript{10} (iv) increased lung damage,\textsuperscript{13} and (v) increased complications for Hodgkin's lymphoma patients.\textsuperscript{23}
Nowadays, there is an increasing demand to improve local treatment outcome. Such an improvement can be achieved if the radiation dose distribution conforms closely to the PTV while avoiding as much of the surrounding normal tissues as possible. This conformal approach has become possible since the introduction of novel technologies, such as computed tomography, three-dimensional treatment planning programs, and computer-controlled multi-leaf collimators. Currently, treatments are frequently constrained by normal tissue tolerances rather than by the tumor dose levels required to achieve local control. In principle, a reduction of the size of the margin of normal tissue allowed in the PTV should lead to an increased patient tolerance to radiotherapy and a decrease of the probability of complications. Thus, conformal therapy presents an opportunity to increase tumor dose levels beyond those used in conventional therapy, leading potentially to increases in the probability of tumor control and survival.\textsuperscript{21,24} Furthermore, conformal therapy allows improved treatment of situations where the clinical target volume (CTV) lies close to a critical structure, such as the spinal cord, the brain stem, the eye, or the bladder. Clearly, both the vicinity of a critical organ and the reduction of the margin around the CTV require an even more stringent level of accuracy in the delivery of daily radiotherapy treatments. In turn, this need for accuracy requires more intensive patient set-up verification procedures.

One way to verify how accurately the irradiated volume corresponds to the PTV defined by the radiation oncologist is to take radiographs of the patient using the treatment beam. Such radiographs are usually called portal, therapy, or localization images. The portal image can be compared subsequently with the previously-obtained simulator film to verify that the actual position of the patient, relative to the treatment beam, conforms to the intended one. In addition,
portal images can be used to estimate patient dose, as well as to ensure that shielding blocks are in the required positions.

Frequent monitoring of patient position allows intervention and therefore a decrease of the magnitude of field shape and placement errors. In two studies, Marks et al. have shown that the frequent acquisition of portal films (and appropriate correction of setup errors, if any) is one of the causes leading to an observed decrease in the probability of local recurrence of Hodgkin’s lymphoma. In 1974, they have observed that frequent use of portal images reduces the occurrence of field placement errors from 55% to 29% for Hodgkin’s lymphoma patients and, in 1976, they observed that frequent verification of complex fields reduces the field placement error frequency from 36% to 15%. Furthermore, Leong and Shimm have shown that a patient can be positioned, consistently and accurately (less than 1 mm uncertainty), using portal films. Because portal imaging can detect and help correct patient positioning errors, other studies have suggested that patient positioning be checked daily.

Despite the obvious advantages associated with frequent verification of patient position in radiation therapy, most Canadian clinics do not perform portal imaging on a regular basis, at any site. An informal survey conducted at a panel discussion at ASTRO 1993 showed that, while over 50% of the clinics represented in the audience possess electronic portal imaging systems, which should allow rapid verification and correction of patient positioning errors, only a few clinics use the systems daily, and only a single clinic uses the systems on every patient, every day. Despite several improvements over film, the image quality obtained with commercial, electronic portal imaging systems still require improvement, and hence the sparse use of these systems. Significant improvements in image quality can result in increasing use of portal imaging, and
eventually lead to improved accuracy in external beam radiation therapy, and, hopefully, to increased patient survival.

1.3 Evolution of portal imaging technologies

Traditional portal imaging uses film to record an image of the patient using the radiotherapy beam. By selecting appropriate film speed and sensitivity, one can verify the position of the patient at the beginning of the daily treatment or throughout the entire treatment duration. Clinical practice has recognized that placing metal plates in direct contact with film improves the quality of the portal films.\textsuperscript{37} Placing a metal plate between the patient and the film reduces the contribution of low-energy, scattered radiation to the portal image and, if the plate is sufficiently thick, provides maximal electron buildup and thus gives maximum exposure to film.\textsuperscript{38} Placing the metal plate behind the film (i.e., back plate) also increases film exposure because of electrons scattered from the plate. However, using a back plate reduces spatial resolution\textsuperscript{39} and image contrast because of the increased contribution of scattered particles to the image.\textsuperscript{40} Many researchers have attempted to find the optimal metal plate composition and thickness for portal film.

While portal film is still the most widely-used method for patient positioning verification, its use is limited. Because the acquisition of a portal film involves stopping the treatment for film retrieval and development, frequent use of portal film involves costs in effort, time, and disturbance to the patient. Additional delays involved with film development prevent portal films from being used for interactive alignment of the patient with the radiotherapy beam. Furthermore, portal films may have poor image quality because of incorrect exposure and because of the fixed display contrast of film, which prevents the opti-
mal display of low contrast structures. Thus, film is not well-suited for portal imaging.

Electronic portal imaging systems have been introduced to overcome the limitations of portal film.\textsuperscript{41-43} Already, systems based on television cameras and ion chamber matrices have been introduced commercially. With these commercial systems, portal images can be acquired, processed, and displayed optimally on a computer monitor, within 5 s, using only a fraction of the therapy dose. Thus, electronic portal imaging systems can be used to detect and correct patient positioning errors without the delays involved with portal film.

The first imaging system which used a television camera to monitor the position of the patient during radiation therapy has been introduced in 1958.\textsuperscript{44,45} The basic design for such systems is relatively simple, as shown in Fig. 1.1.\textsuperscript{46-49} The x-ray detector of video-based systems consists of a metal plate bonded to a gadolinium oxysulfide (Gd$_2$O$_2$S:Tb) screen. This detector is viewed by a lens and television camera combination via a 45° mirror to prevent damage to the camera from the direct radiation beam. In current implementations, the video signal generated by the camera is digitized by a frame grabber installed in a computer located outside of the treatment room. The digitized image can then be stored on the computer, processed, and displayed on the computer monitor.

As shown in Fig. 1.2, the main problem associated with video-based systems is that only a small fraction (between 0.1% and 0.01%) of the light emitted by the x-ray detector reaches the lens of the television camera. This large loss of information-carrying quanta reduces portal image quality since the noise generated by the electronic components of a video-based system may overwhelm the small video signal if only a small quantity of light reaches the television camera. Furthermore, Munro et al. have shown that the spatial reso-
Figure 1.1: Schematic of television camera-based portal imaging systems. Under x-ray irradiation, the light emitted by the phosphor screen is viewed by a television camera using a mirror and a wide-aperture lens.

Resolution obtained with video-based systems is much worse than that obtained with portal films. Many approaches have been proposed to eliminate these problems and optimize the design of video-based systems.

The first attempt at optimizing the design of video-based systems has proposed an increase in the thickness of the phosphor screen, an increase in the lens aperture, as well as an optimized television camera read-out sequence, resulting in dramatic improvements in image quality. First, optimizing the camera read-out sequence minimizes the contribution of electronic noise to the final image. Secondly, increasing the thickness of the phosphor screen increases the light output of the x-ray detector, thus minimizing the loss of information-carrying quanta. This loss is further reduced by the use of a large-aperture lens.
Figure 1.2: Schematic diagram demonstrating the light collection efficiency of portal imaging systems based on a television camera. The narrow cone shows how only a small fraction of the light emitted from the x-ray detector reaches the camera.

Efforts have been made to improve further the efficiency of video-based systems. Researchers have studied the trade-offs in light output, spatial resolution, and noise when the thickness of the phosphor screen is varied.\textsuperscript{50-55} Alternative x-ray detectors, such as scintillating glasses and grooved phosphor screens,\textsuperscript{56,57} and cameras which are more sensitive to light,\textsuperscript{58,59} have also been proposed for portal imaging. More recently, Jaffray et al. have shown that striking improvements in image quality are possible when using a CCD (charge-coupled device) camera cooled to -20°C to eliminate electronic noise.\textsuperscript{60,61} Researchers have also studied the influence of other physical factors which could affect image quality, such as x rays scattered from the patient\textsuperscript{62} and the finite size of the focal spot of the linear accelerator x-ray source.\textsuperscript{63,64}

The second disadvantage of video-based systems is their bulkiness, which can hinder patient setup and may lead to an increase of field placement errors. Furthermore, these heavy, bulky devices may not be sufficiently rigid, resulting in deviations of the position of the x-ray detector relative to the central axis of the beam as the gantry of the linear accelerator is rotated.\textsuperscript{42}
To overcome the inconvenience of the weight and bulk of video-based systems, researchers have proposed portal imaging devices based on two-dimensional matrices of radiation detectors. One such device, based on an matrix of liquid ionization chambers, is already commercially-available. Its compact size (outer dimensions of 52.5 cm x 52.5 cm x 4 cm and a weight of 7 kg) makes the ion chamber matrix easier to attach rigidly to a linear accelerator gantry. Another advantage is that the images obtained with this device are not affected by the diffusion of optical quanta in the x-ray detector and by geometric distortions encountered in video-based systems because of the lens of the television camera (e.g., spherical aberrations, vignetting, pin cushion and barrel distortions). This system can be used to measure the transmitted dose rate during radiation therapy.

The main limitation of this liquid ion chamber matrix is low quantum utilization. This is because a large fraction of the ions created in the ionization chambers are not collected and therefore do not contribute to the portal image. For this reason, the matrix ionization chamber requires higher doses, and thus longer exposure times (~5 s), to form images than video-based systems. This long acquisition time may hinder both the immediate correction of patient positioning errors and the verification of dynamic procedures, such as arc treatments and variable beam colimation.

Prototype matrix systems based on solid-state devices (i.e., amorphous silicon arrays and amorphous selenium plates) have been proposed by some researchers. Antonuk et al. have proposed an array of amorphous silicon photodiodes for portal imaging. This array, which is extensively described in the literature, is highly resistant to radiation damage. Therefore, the amorphous silicon array can be coupled directly to an x-ray detector of design
similar to that of video-based systems, leading to a light collection efficiency which is orders of magnitude higher than that of video-based systems. Furthermore, the absence of lenses results in the absence of off-axis losses in spatial resolution. Thus, the amorphous silicon array allows a portal imaging system design where the compactness of the ion chamber matrix is combined with the higher efficiency of x-ray detectors encountered with video-based systems.

The main problem encountered with this device is the noise generated by the electronic components of the system. Unfortunately, this noise increases as the size of the array increases and as the length of the cable between the amorphous silicon array and the electronic preamplifiers increases. This last problem is exacerbated by the fact that, unlike the amorphous silicon array, the external electronic components are subject to radiation damage, and thus must be located as far from the radiation beam as possible. Although efforts have been made to eliminate these noise problems, the time of commercial introduction of amorphous silicon arrays is uncertain.

1.4 Imaging system optimization

As seen in Sec. 1.3, many approaches have been proposed to verify the position of the patient with respect to the radiation therapy treatment beam. Which is best suited for the task? Obviously, the approach to choose is the one which allows error-free detection of structures of interest within the patient. Perhaps existing system designs can be improved by the selection of components better-suited to the imaging task and by optimizing their usage, so that they can offer image quality comparable to that obtained with existing prototype systems. In this case, other considerations, such as cost, practicality, and availability, become the relevant parameters. Nevertheless, the image quality obtained with these various system designs must be compared quantitatively.
1.4.1 Detective quantum efficiency and signal-to-noise ratio

The ability to visualize a structure depends, among other factors, on contrast, which is the difference in intensity between different areas in an image. Differing degrees of grayness, or contrast, allow an observer to interpret the information contained in an x-ray image. Mathematically, contrast is defined as the ratio of the intensity variation between two regions of interest to the average background intensity. However, random intensity variations, which are governed by the statistical properties of the incoming x-ray beam and those of the imaging system, may obscure intensity variations caused by the object we wish to visualize (i.e., signal). Thus, a more basic measure of image quality is the signal-to-noise ratio (SNR), which is defined mathematically as the ratio of the desired intensity variations to random intensity variations.

The simple formalism of Motz and Danos\(^6^9\) can be used to illustrate the concepts of contrast and SNR and explain the respective properties of these quantities. Figure 1.3 depicts a simple radiographic imaging task - the detection of a piece of bone of thickness \(x\) embedded in a homogeneous slab of tissue of thickness \(T\).

Using the above definition and Fig. 1.3, contrast is mathematically expressed as\(^6^9\)

\[
C = \frac{\text{signal}}{\text{mean signal}} = \frac{\phi_1 - \phi_2}{(\phi_1 + \phi_2 + 2\phi_s)/2},
\]

where \(\phi_1, \phi_2,\) and \(\phi_s\) are the primary and scatter fluences reaching the x-ray detector (Fig. 1.3). Since these fluences are related to the thickness and x-ray attenuation coefficients of the materials shown in Fig. 1.3, contrast can be also expressed as\(^6^9\)

\[
C = \frac{2(1 - e^{-\lambda})}{1 + e^{-\lambda} + \frac{2SF}{1 - SF}},
\]

(2)
where $SF$ is the scatter fraction (i.e., the ratio of the scatter fluence to the total fluence) and $\Delta$ is the difference in attenuation between the piece of bone and the background (i.e., $\Delta = x |\mu_2 - \mu_1|$), and $\mu_1$ and $\mu_2$ are the x-ray attenuation coefficients for tissue and bone, respectively. Thus, contrast increases as the difference in attenuation between bone and tissue increases, and decreases as the scatter fraction increases. For the x-ray energies used in diagnostic radiology, the scatter fraction is high, but the difference in attenuation between bone and soft tissue is large. On the other hand, for the high energy x rays used in radiation therapy, the scatter fraction is small, but the difference in attenuation between bone and soft tissue is disappointingly small. Thus, contrast is 10 to 20 times less at radiotherapy energies than at diagnostic energies. Note that
contrast is independent of the number of x rays incident on the object being imaged.

As described above, the SNR relates random fluence variations (i.e., noise) with fluence variations caused by the presence of an object. It can be shown that the number of x rays recorded by an x-ray detector is a random process described by Poisson statistics. Therefore, the variance in the number of detected x-rays is equal to the mean number of detected x-rays. Assuming Poisson statistics, and using Fig. 1.3, the SNR can be expressed as

\[
SNR = \frac{\text{signal}}{\text{noise}} = \frac{\phi_1 - \phi_2}{\sqrt{\phi_1 + \phi_2 + 2\phi_s}} = \frac{C}{2} \sqrt{\phi_1 + \phi_2 + 2\phi_s}.
\]

Equation (3) shows that the signal-to-noise ratio depends directly on contrast, and thus depends also on differences in attenuation coefficients and on scatter fluence. Rewriting Eq. (3) in terms of x-ray attenuation differences (i.e., \(\Delta\)), we obtain

\[
SNR = \frac{C}{2} \left[ A \phi_0 \eta e^{-\mu_T} \left(1 + e^{-\lambda} + \frac{2SF}{1-SF} \right) \right]^{\frac{1}{2}},
\]

where \(A\) is the area of the object or structure of interest, \(\phi_0\) is the incident x-ray fluence, \(T\) is the thickness of the tissue being imaged (Fig. 1.3), and \(\eta\) is the detection efficiency of the x-ray detector. Equation (4) shows that, unlike contrast, the SNR depends on the square root of the number of x-rays incident on the patient. One implication from this result is that any structure can be imaged with a SNR sufficiently high, provided that the object is irradiated with a sufficiently large number of x rays. Note that the results obtained with the Motz and Danos approach do not take into account losses in spatial resolution and sources of noise other than primary x-ray quantum noise. Thus, any complete characterization of an imaging system requires the determination of the spatial frequency response and the signal-to-noise characteristics of the system.
Unfortunately, when verifying the position of the patient during a course of radiation therapy, one cannot irradiate patients indefinitely to reveal low-contrast anatomical structures in the portal image, if only to not exceed the prescribed radiation dose. Ideally, portal imaging should allow the interactive correction of the position of the patient throughout the course of a treatment. If the fraction of the therapy dose which is allocated for portal imaging is large, then a large fraction of the treatment will be delivered, possibly erroneously, before patient positioning can be corrected. Thus, similar to diagnostic radiology, the radiation dose required to form a high-quality portal image should be minimized.

The parameter of choice to compare x-ray imaging systems is the detective quantum efficiency (DQE). The DQE defines the SNR transfer characteristics of an imaging system in terms of spatial frequency $\omega$, and is defined as

$$DQE(\omega) = \left[ \frac{SNR_{\text{out}}(\omega)}{SNR_{\text{in}}(\omega)} \right]^2,$$

where $SNR_{\text{out}}(\omega)$ is the SNR of the image produced by the imaging system and $SNR_{\text{in}}(\omega)$ is the SNR associated with the x-ray field impinging on the x-ray detector. An ideal system would detect perfectly all of the signal and noise contained in the incident x-ray field without adding noise to the image, leading to a DQE of 1. However, real imaging systems have DQEs lower than 1 since they add noise to the image and do not detect all incoming x-rays. Therefore, the DQE measures how efficiently an imaging system makes use of the information contained in the incoming radiation field by simultaneously accounting for signal losses and noise increases occurring in the detection process.
1.4.2 Quantum accounting diagram

In most radiographic systems, x rays are converted to secondary quanta, such as optical photons or electrical charge-carriers, in multiple cascaded stages. The number of quanta leaving one stage becomes the effective input to the following stage. Primary input quanta are converted to secondary quanta through one or more stages before contributing to the final image. Any radiographic system extracts information from the x-ray image in the first stage only (i.e., the detection of individual x rays). All later "image-forming stages may result in the loss of information, but never in a gain. Even if subsequent stages have large amplification factors, the information statistically coded in the radiographic image is still limited to the number of x rays absorbed in the x-ray detector.

It is important to maintain high amplification factors at each image-forming stage. This is not to add information to the image, but to prevent the loss of information. The stage with the fewest quanta is called the quantum sink. The quantum sink defines the limiting value of the image SNR to be, at best, no greater than the square root of $\phi_{\text{min}}$. As shown by Eq. (4), SNR depends on the square-root on the number of information-carrying quanta incident on a stage. Because information-carrying quanta are lost irrecoverably in the quantum sink, amplification stages following a quantum sink cannot improve the SNR. Therefore, a well-designed imaging system has a quantum sink located at the very first image-forming stage (i.e., a primary quantum sink), so that the information encoded in the image depends only on the number of x-rays detected by the imaging system. Such a system is called "x-ray quantum limited". If the quantum sink corresponds to any other stage (i.e., secondary quantum sink), further losses of information-carrying quanta occur in the imaging system due to sub-optimal system design, and image quality is reduced.
For over 47 years, system designers have determined the location of the quantum sink by evaluating the average number of quanta at each stage as the product of all preceding system gain factors for each stage.\textsuperscript{70,71} The results are often displayed graphically in a diagram where the average number of quanta (per pixel or per incident primary quantum) is plotted as a function of stage number. Such diagrams are sometimes called "nomograms". The quantum sink corresponds to the lowest point (i.e., fewest quanta) on the graph. Using this graph, designers can select specific factors, such as x-ray detection efficiency, amplification factors or coupling efficiencies to prevent or remove secondary quantum sinks and optimize the design of an imaging system. However, this approach is limited to the extent that it neglects the spatial spreading of secondary quanta which always occurs in real imaging systems (resulting in a "zero-spatial-frequency" analysis), and it is an approximate analysis that neglects the noise-enhancing effect of non-deterministic gain stages on image noise. Most (if not all) realistic gain stages are non-deterministic, implying that some degree of statistical uncertainty exists in the actual gain value.

Both limitations of the zero-spatial-frequency analysis have been removed recently by the introduction of the concept of a spatial-frequency-dependent quantum sink.\textsuperscript{72} In this revised approach, each image-forming stage of an imaging system is either a gain stage or a blur stage, but not both (see Fig. 1.4). The introduction of blur stages in the zero-spatial-frequency analysis allows the calculation of an effective number of quanta at each stage for any spatial frequency. Plots of the effective number of quanta, for given spatial frequencies, as a function of stage number gives the "quantum accounting diagram" (QAD), which is calculated by the product of all gains and squared MTF values for up to and including the stage of interest. Fig. 1.5 shows a sample QAD.
Figure 1.4: Schematic diagram representing elementary gain and spread stages for quanta entering a single pixel.

A direct relationship has been shown to exist between this QAD analysis and the more comprehensive DQE analysis, thereby linking these previously unrelated approaches. The relationship between the DQE and the QAD is based on the properties of signal and noise propagation, as a function of spatial frequency, through the image-forming stages of a system. Based on equations derived by Rabbani et al., Cunningham et al. have shown that the DQE for an imaging system consisting of a number $M$ of image-forming stages is

$$DQE_{i,M}(\omega) = \frac{1}{1 + \sum_{i=1}^{M} \left( \frac{1 + \varepsilon_{\phi_i} |T_i(\omega)|^2 + S_a(\omega)/\Phi_i}{P_i(\omega)} \right)}$$

(6)

where $\varepsilon_{\phi_i}$ is the Poisson excess [i.e., $\varepsilon_{\phi_i} = (\sigma^2/\bar{g}_i) - 1$] associated with an average gain $\bar{g}_i$, $T_i(\omega)$ is the MTF for stage $i$, $S_a(\omega)/\Phi_i$ is the relative additive noise in the $i^{th}$ stage and $P_i(\omega)$ is the product of the gains and squared MTFs up to and including the $i^{th}$ stage, given by

$$P_i(\omega) = \prod_{j=1}^{i} \bar{g}_j |T_j(\omega)|^2.$$  

(7)
Figure 1.5: Sample quantum accounting diagram for zero (solid) and a non-zero (dashed) spatial frequencies. In a gain stage, the QAD lines corresponding to different spatial frequencies are parallel. In a spread stage, the QAD lines diverge.

Equation (7) describes the product of the mean gain and squared MTF of every stage up to and including the $i^{th}$ stage and can be interpreted as an effective number of quanta (x rays, light quanta, or photoelectrons) which propagate the image signal through the stages of the system, at each spatial frequency. A plot of $P_i(\omega)$, as a function of stage number $i$, yields the QAD line for spatial frequency $\omega$. Therefore, the QAD is an essential component of the theoretical DQE of an imaging system.

1.5 Summary

Clinical studies have shown that, despite improved immobilization and clinical practice over the years, errors in the positioning of the patient relative to the radiotherapy treatment beam still occur frequently. These errors may jeopardize the success of radiotherapy and influence treatment outcome. Fortunately, frequent portal imaging has been shown to reduce the occurrence of patient positioning errors. Despite the advantages of portal imaging, few cancer clinics use portal imaging regularly, in part because of the disappointingly low image quality obtained with portal imaging. A number of studies have ad-
dressed the fundamental physics of portal imaging. Investigators, including ourselves (see Chapter 2), have been trying out novel imaging technologies in the hopes of improved portal image quality. This empirical approach has lead to some success, but also to many disappointments since the weak points in a system design are not properly identified. Thus, many of the system designs presented in Sec. 1.3 are not optimal, leading to disappointingly low image quality, underusage of portal imaging technology, and, ultimately, to a probable decrease in tumor control and/or patient survival rates. This thesis presents three theoretical tools that can be used to optimize the design and usage of medical imaging systems. These tools have been used to improve the quality of images obtained with video-based portal imaging systems, as well as compare the imaging performance of video-based systems with other system designs. The tools described in this thesis can point to system-dependent limitations on portal image quality. If possible, these limitations should be removed so that the only physical limitations to portal image quality reside in the number of x rays absorbed by the detector of such systems.

1.6 Thesis overview

The goal of this thesis research is to optimize the use and design of a video-based portal imaging system by integrating many of the various fundamental factors that limit the quality of portal images. In this thesis, we present fully detailed analyses which integrate most of the factors affecting image quality, including the size of the x-ray source, x-ray absorption noise, as well as all of the physical parameters associated with each of the components of a video-based portal imaging system. These analyses have been applied to optimize the design and usage of this video-based system as well as to compare the image quality obtained with this system with that obtained with other portal imag-
ing devices. The material is presented as follows: (a) optimization of the radiographic magnification for existing video-based systems and portal films; (b) empirical optimization of the x-ray detector of a video-based system; (c) detailed analysis, using a quantum accounting diagram (QAD), of a video-based portal imaging system; and, (d) demonstration of an approach based on QAD's for the optimization of the design of linear imaging systems.

Chapter 2 presents an approach, based on the interplay between the size of the x-ray source and the noise and resolution characteristics of an imaging system, to compute the radiographic magnification that minimizes the combined blurring due to the imaging system and the focal spot of the x-ray source. Thus, by using the optimal magnification, image quality can be improved without modifying the design of existing imaging systems. Chapter 3 describes some of the difficulties encountered when empirical approaches are used to improve the design of an existing imaging system. The work described in Chapter 3 shows that selecting a system component on the basis of a single desirable parameter, without considering other relevant factors, may lead to disappointing results. This prompted a detailed analysis of the video-based portal imaging system, using the quantum accounting diagram and the theoretical derivation of the detective quantum efficiency of the system (Chapter 4). All of the gain, blur, and electronic noise stages of a video-based portal imaging system were measured or calculated, and the impact of each stage on image quality was determined. Chapter 5 presents an approach for the optimization of imaging system design based on the analysis presented in Chapter 4. This approach, which involves the physical parameters involved with the human visual system and the computation of a single figure of merit, was applied to compare small and radical changes to the design of an existing portal imaging system.
Finally, Chapter 6 summarizes our results and presents possible avenues for further work in the field of portal imaging.

Chapter 2 was published in Med. Phys. 21 (9), pp. 1435-1445 (1994), under the title “Optimal radiographic magnification for portal imaging” by J.-P. Bissonnette, D. A. Jaffray, A. Fenster, and P. Munro. Chapter 3 was published in Med. Phys. 23 (3), pp. 401-406 (1996), under the title “Evaluation of a high-density scintillating glass for portal imaging” by J.-P. Bissonnette and P. Munro. Chapter 4 has been accepted for publication in Medical Physics under the title “A quantum accounting and detective quantum efficiency analysis for video-based portal imaging” by J.-P. Bissonnette, I. A. Cunningham, D. A. Jaffray, A. Fenster, and P. Munro. Chapter 5 was submitted to Medical Physics under the title “Optimal phosphor thickness for portal imaging” by J.-P. Bissonnette, I. A. Cunningham, and P. Munro. All of the authors participated in the development and planning of the work presented in this thesis. All of the experimental and theoretical work was performed by J.-P. Bissonnette, under the supervision of P. Munro.
References


36 P.B. Dunscombe, The future of portal imaging, presented at the 41th Annual General Meeting of the Canadian Organization of Medical Physicists and the Canadian College of Physicians in Medicine, Montréal, June 1995.


Chapter 2: OPTIMAL RADIOGRAPHIC MAGNIFICATION FOR PORTAL IMAGING

Abstract

We have been using two approaches to estimate the optimal radiographic magnification for a T.V. camera-based portal imaging system and for portal films. The first approach optimizes signal transfer while the second optimizes signal-to-noise ratio (SNR) transfer. In order to perform these optimization calculations, the physical characteristics of the imaging system [MTF and NPS] as well as the sizes of the radiation sources of our medical linear accelerators have been measured. Using these data, the optimal magnification considering signal transfer alone ($M_{\text{signal}}$) has been calculated to range between 2.0 and 2.3 for the T.V. camera-based imaging system and is about 1.0 for portal films. Conversely, the optimal magnification considering SNR transfer ($M_{\text{SNR}}$) has been calculated to range between 1.5 and 1.7 for the T.V. camera-based imaging system and is about 1.0 for portal films. The results suggest that most portal imaging systems are operated close to their optimal radiographic magnification.

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1 This chapter was published in Medical Physics, 21 (9) 1435-1445 (1994). The title of the article is "Optimal radiographic magnification for portal imaging" by J.-P. Bissonnette, D. A. Jaffray, A. Fenster, and P. Munro.
2.1 Introduction

Improvements in medical linear accelerators and in treatment planning capabilities have increased our ability to plan and deliver complex radiation treatments. An accuracy of 5% is required for the delivered dose in the target volume.\textsuperscript{1,2} While treatment planning programs generally can predict x-ray dose distributions within 3-5%,\textsuperscript{3,4} the accuracy of the actual treatment is reduced by systematic and random errors in the placement of the patient with respect to the therapy beam (field placement errors). These errors occur frequently and can influence the outcome of a treatment.\textsuperscript{5-10} Fortunately, frequent monitoring of patient positioning can improve the geometric accuracy of the treatment.\textsuperscript{5,11} On-line portal imaging provides a quick and accurate assessment of the geometric precision of radiotherapy treatments, and may improve treatment outcome.\textsuperscript{11}

Many portal imaging systems are being developed to monitor patient position routinely.\textsuperscript{12-20} The development of these systems have made studies of the fundamental limitations of imaging with high energy x rays necessary. Not only must the performance of the imaging systems themselves be assessed, but the influence of other factors, such as the size, shape, and energy of the radiation source must also be determined.

In transmission radiography, spatial resolution is reduced due to blurring caused by both the radiation detector and the finite size of the focal spot. Figure 2.1 shows the effect of radiographic magnification on image quality in transmission radiography. A large magnification will minimize the blurring caused by the detector because a magnified image is projected at the image plane. However, a large magnification will maximize the blurring introduced by the finite size of the focal spot. When magnification is close to unity, the focal spot will
Figure 2.1: The effect of radiographic magnification on blur caused by the focal spot and by the radiation detector. Large magnifications minimize blur due to the radiation detector but emphasize focal spot blur. Conversely, small magnifications minimize focal spot blur but emphasize blur due to the radiation detector.

have minimal effect on spatial resolution but the detector blur will be maximized. Clearly, an optimal magnification exists where the combined blurring due to the detector and the focal spot will be minimized.

Droege and Cytacki used measurements of metal plate/film modulation transfer functions [MTF] in combination with assumptions about the x-ray source size and shape to estimate the effect of magnification on spatial resolution of portal films. Swindell et al. calculated theoretically the spatial resolution of a crystal scintillator imaging system and used these data in combination with assumptions about the size and shape of the x-ray source to estimate the optimal magnification. In both cases, only the effects on spatial resolution (i.e. signal transfer) were considered. Meertens et al. also have considered the effects of radiographic magnification on the line spread functions [LSF] measured from digitized portal films.

In this chapter, we calculate the optimal magnification for a T.V. camera-based portal imaging system and for portal films using two approaches. One approach calculates the magnification which optimizes signal transfer (i.e.
where the equivalent passband$^{25,26}$ is maximized), while the other approach determines the magnification which optimizes signal-to-noise ratio (SNR) transfer. These calculations are limited to the interplay between the focal spot and imaging system and do not model the entire radiographic situation. (Note that in our terminology the "imaging system" refers to the radiation detector, and not to the entire imaging chain.) Therefore, factors, such as x-ray scatter, which could influence the optimal magnification have not been considered in this paper. We have used the measurements of Jaffray et al. to determine the MTFs of linear accelerator focal spots since their measurements show that isotropic, Gaussian-shaped focal spots are the exception rather than the rule for medical linear accelerators.$^{27-28}$ Although the results presented in this paper are specific to our combination of portal imaging equipment and accelerators, these results have general application because of similarities in the characteristics (i.e. x-ray focal spot size) of linear accelerators and similarities in characteristics between our portal imaging system and those which are widely available commercially.

2.2 Theory

2.2.1. Signal transfer

Two approaches can be used to find the optimal radiographic magnification. The first approach determines the magnification which maximizes signal transfer,$^{25,26,29}$ where one seeks a magnification that takes maximum advantage of the resolution capability of the imaging system while avoiding an undesirable degree of geometric unsharpness caused by the focal spot. In such an optimization, we consider only the MTF of the focal spot [$\text{MTF}_{fs}$] and the MTF of
the imaging system \([\text{MTF}_{\text{sys}}]\). The MTF of the entire imaging chain (i.e., imaging system + x-ray source) \([\text{MTF}_{\text{tot}}]\) can be given by\(^{25,29}\)

\[
\text{MTF}_{\text{tot}}(f') = \text{MTF}_{\text{Is}} \left( f' \frac{[M-1]}{M} \right) \times \text{MTF}_{\text{sys}} \left( \frac{f'}{M} \right)
\]

(1),

where: \(M\) is the radiographic magnification; and, \(f'\) represents spatial frequencies referenced to the object plane. A common method used to calculate the optimal magnification is to assume that the MTFs have a Gaussian shape and to calculate the optimal magnification using the full-width at half-maximum of the MTFs, as described by Barrett and Swindell.\(^{29}\) An alternative approach is to maximize the area under the squared MTF of the imaging chain. The area under the squared two-dimensional MTF (i.e., \(\int \int \text{MTF}^2_{\text{tot}} \left( f_x, f_y \right) df_x df_y\), or \(2\pi \int \text{MTF}^2_{\text{tot}} (f) f df\) if radial symmetry is assumed), which has been defined as the equivalent passband, has been correlated with subjective impression of sharpness.\(^{25}\) Therefore, the magnification which optimizes signal transfer, \(M_{\text{signal}}\), occurs where a plot of the equivalent passband versus magnification reaches a maximum. Note that this approach only considers image sharpness and does not account for the influence of scattered radiation on the optimal magnification.

Thus, to find the magnification, \(M_{\text{signal}}\), that optimizes signal transfer (i.e., maximizes the square of the area under \(\text{MTF}_{\text{tot}}\)), we must measure the MTF of the accelerator focal spot and the MTF of the imaging system.

2.2.2 SNR transfer

The second approach optimizes the magnification by maximizing signal-to-noise ratio transfer. Sandrik and Wagner\(^{26}\) have derived an expression for the SNR of the entire imaging chain (i.e., imaging system + x-ray source) in terms of: (i) the system MTF; (ii) the system noise power spectrum [NPS]; and,
(iii) the focal spot MTF. The square of the SNR per unit photon reaching the object is given by:\(^{26}\)

\[
\frac{SNR^2}{qM^2} = \int \int \frac{MTF_{sys}^2\left(\frac{f'_x}{M}, \frac{f'_y}{M}\right)}{NPS\left(\frac{f'_x}{M}, \frac{f'_y}{M}\right)} MTF_{is}^2\left(\frac{f'_x[M-1]}{M}, \frac{f'_y[M-1]}{M}\right) df'_x df'_y \tag{2}
\]

where: \(q\) is the photon fluence used to form an image at \(M=1\); \(NPS\) is the noise power spectrum of the imaging system; and, \(MTF_{is}, MTF_{sys}\) and \(f'\) have been defined in Eq.(1). As discussed in Ref. 26, Eq. (2) assumes a unit-contrast object such that its spatial-frequency representation is constant over the frequency domain where the quantity \(MTF_{sys}^2 MTF_{is}^2/NPS\) is non-negligible. In other words, Eq. (2) is object-independent. Generally, the \(qM^2\) term is used to obtain the \(SNR^2\) per unit object fluence when the fluence at the image receptor is maintained. If the MTFs and the NPS are circularly symmetric, Eq. (2) then becomes:\(^{26}\)

\[
\frac{SNR^2}{qM^2} = 2\pi \int \frac{MTF_{sys}^2\left(\frac{f'}{M}\right)}{NPS\left(\frac{f'}{M}\right)} MTF_{is}^2\left(\frac{f'[M-1]}{M}\right) f' df'	ag{3}
\]

The optimal magnification calculated using the SNR transfer method (\(M_{SNR}\)) can be determined from the maximum of a plot of \(SNR^2/(qM^2)\) as a function of magnification \(M\). Note that this calculation is limited to the interplay between the resolution and noise characteristics of the focal spot and the imaging system, and does not model the entire radiographic process (e.g., does not consider the influence of x-ray scatter on the optimal magnification).

Thus, to find the magnification, \(M_{SNR}\), that optimizes SNR transfer by use of Eqs. (2) or (3), we must measure the MTF of the accelerator focal spot, the MTF of the imaging system, and the NPS of the imaging system.
2.3 Materials and methods

2.3.1 MTF

To calculate the optimal magnification for portal films and the portal imaging system, using either the signal transfer method or the signal-to-noise transfer method, the system MTF (i.e., the MTF of the entire imaging system, including the copper plate/phosphor screen, lens, T.V. camera, and frame grabber), the portal film MTF (i.e., the MTF of film in contact with a metal plate), and the focal spot MTF (i.e., the MTF of the x-ray source for a given magnification) of our Varian 2100c linear accelerator must be known. Also, for reasons explained in section 2.4.1, the detector MTF (i.e., the MTF of the copper plate/phosphor screen alone) must also be known. The portal film MTFs and the focal spot MTFs were obtained from the literature\textsuperscript{24,27,28} while the system and detector MTFs were measured using the techniques described below.

Both the system and detector MTF were measured using methods described in detail elsewhere\textsuperscript{30-32}. Two steel blocks, 5 cm thick, 10 cm wide, and 60 cm long, were clamped together to collimate the radiation beam to a narrow slit. Steel shims clamped between the edges of the blocks defined the width of the slit used to form the beam. The slit width was measured using a 50 kV\textsubscript{p} x-ray beam and direct-exposure film. Because of the different resolution characteristics of portal films and of the portal imaging systems, two different slit widths were used. The steel blocks were mounted on a heavy duty translation stage (Daedal Inc., Harrison City, Pa). With the radiation beam of the linear accelerator directed horizontally (i.e. gantry angle of 90°), translating the steel blocks horizontally aligned the slit with the x-ray focal spot of the accelerator.

The MTF of the copper plate/phosphor screen detector of our T.V. camera-based imaging system, which we will subsequently refer to as "detector
MTF* was measured by exposing Kodak EM-1, single emulsion film placed in contact with the copper plate/phosphor screen detector to an 88 µm wide slit of x-rays. The films were processed in a Kodak M6B X-Omat film processor. These "slit images" were then digitized using an Optronics International Inc. Model P-1000 drum scanner (Optronics International Inc., Chelmsford, Ma). Six different films, each exposed to a different x-ray dose (ranging between 10-300 monitor unit* irradiations from the linear accelerator), were used to determine the tails of the LSF accurately. The data obtained from these films were linearized using the sensitometric curve of the film, averaged to reduce the uncertainty in the estimate of the LSF (i.e. between 800 and 1000 rows of pixels were averaged), pieced together, and processed using standard techniques developed for the measurement of MTF's in diagnostic radiology. The tails of the LSF were extrapolated using a least-squares exponential fit, and the detector MTF, obtained from the Fourier transform of the normalized LSF, was corrected for the effects of the finite width of the radiation slit.

The system MTFs were obtained from images produced by the T.V. camera-based imaging system irradiated using a 405 µm slit of radiation. [Measurements made with a narrower (130 µm) beam showed that the system MTF was negligible for spatial frequencies higher than 1.5 mm⁻¹.] Two sets of images were acquired to account for spatial nonuniformities that arise because of: (i) the dark current generated in the target of the T.V. camera; (ii) an imperfect shading correction;* and, (iii) nonuniformities in the phosphor layer. One set of images was acquired with the slit beam of x rays irradiating the imaging system

* 1 monitor unit = 1 cGy dose to the isocentre for a 10 x 10 cm² field at a depth of 5 cm in water.

* Local variations in photosensor properties, caused by off-axis vignetting of a lens and/or by non-normal landing of the electron beam, appear as darker or lighter areas in an image. This is called "shading". Shading is corrected by means of a special video amplifier whose amplification changes depending upon which part of the photosensor is being interrogated by the electron beam. Sometimes, this shading correction cannot compensate completely for the non-uniformities.
(slit images), and another set of images was acquired with the steel blocks positioned so that the x rays could not reach the imaging system (background images). The slit and background images were acquired with exposure times similar to those used to acquire clinical images (~1s) so that the measured system LSFs are representative of the clinical situation. The slit and background images were digitized by an 3-bit video frame-grabber and were acquired at least 10 seconds apart to minimize the effects of T.V. camera lag. Since the portal imaging system is linear, the digitized background and slit images could be used without corrections for linearity. The slit images were summed to reduce noise and thereby increase the confidence in the estimate of the tails of the LSF. The difference between the summed slit images and the summed background images resulted in the "LSF image". LSF images were acquired with the slit oriented parallel and perpendicular to the scan lines of the T.V. camera, and with 6 and 18 MV x-ray beams.

To determine the MTF for spatial frequencies higher than the Nyquist frequency determined by the spacing of the pixels, the angulated slit technique was used. The slit was projected at a slight angle with respect to the sampling grid to decrease the sampling interval for this experiment. The angulated slit resulted in a shift of the sampling relative to the LSF (see Fig. 2.2), which varied between 25 μm and 52 μm, depending on the angle of the slit relative to the sampling grid. The reordering of successive rows (sampling parallel to the scan lines) or columns (sampling perpendicular to the scan lines) of pixels resulted in an LSF with effective sampling intervals much smaller than the pixel size.

As described in section 2.4.1, one of the limitations of the MTF measurement is the accurate estimation of the tails of the LSF of a T.V. camera-based imaging system. The tails of the LSF have been extrapolated from the central portion of the measured LSFs using a least-squares exponential fit. However,
this extrapolation results in a systematic overestimate of the system MTF. The tails of the LSF are expected to decrease more slowly than the central portion of the LSF due to scattered photons and/or bremsstrahlung photons generated inside the detector itself.\textsuperscript{24} To correct this overestimation, the system MTFs were renormalized to equal that of the detector MTF at a spatial frequency of 0.1mm\textsuperscript{-1}, reducing the system MTFs by approximately 10%. (The detector MTF, which was measured with film, could be determined accurately at a spatial frequency of 0.1mm\textsuperscript{-1}.) The system MTF, which was obtained from the Fourier transform of the LSF, was corrected for the effects of the finite width of the radiation slit.\textsuperscript{31,33} Following the convention of Refs. 35 and 36, the system MTFs were not corrected for the effects of the pixel aperture (which was 0.67 mm $\times$ 0.67 mm).

Therefore, these calculations yielded the presampling MTF of the T.V. camera-based portal imaging system. The uncertainty in the MTF was assessed from repeated independent measurements. The measurements were repeated three times with the 6 MV x-ray beam, and twice with the 18 MV x-ray beam.

\textbf{Figure 2.2:} Illustration of the tilted slit technique used to measure the MTF beyond the Nyquist frequency.\textsuperscript{34} Each row of pixels samples the LSF with a slightly shifted sampling interval. Reordering the samples from many rows yields a highly sampled LSF.
2.3.2 NPS

To calculate the optimal radiographic magnification using the signal-to-noise transfer method, the NPS for the T.V. camera-based imaging system and for portal films must be known. The NPS for portal films (ortho-M film in contact with a tungsten plate), when irradiated by 6 and 18 MV x-ray beams, were obtained from Ref. 24. The following describes the measurement of the NPS of the T.V. camera-based imaging system.

The imaging system was irradiated with a uniform beam, producing a large number of "uniform illumination images". These images were obtained with one monitor unit irradiations from a Clinac 2100c linear accelerator which were generated at a dose rate of 240 monitor units per minute. The accumulation time was equal to the time taken by the linear accelerator to deliver a 1 monitor unit irradiation (0.3-0.5 s, typically), which is identical to the accumulation time used clinically. Structural variations in these images due to: (i) lens vignetting; (ii) imperfections in the phosphor screen; (iii) x-ray beam nonuniformities (i.e., flatness and symmetry); and, (iv) imperfect camera shading corrections; were eliminated by subtracting two sets of "uniform illumination images" to yield a set of "noise images". These structural variations were eliminated because they were larger than the fluctuations caused by random processes (e.g., photon counting statistics, preamplifier noise ...). A large number (~55) of noise images were used to reduce the uncertainty in the estimation of the NPS. The central portion (256×256 pixels) of the noise images was used to calculate directly the two-dimensional NPS,\(^3\) as required by Eq. (2). The NPS was measured three times, over a period of 10 months. Since the subtraction increases the variance by a factor of 2, the NPS have been corrected accordingly.

The noise in images obtained with our video-based imaging system has five sources: (i) random fluctuations in the detection of x-rays (i.e., x-ray quan-
tum noise); (ii) random fluctuations in the generation of secondary quanta (i.e., optical light and photo-electrons in the T.V. camera); (iii) thermal noise from the input resistor of the T.V. camera (i.e., Johnson noise); (iv), noise generated by the preamplifier of the T.V. camera; and (v), noise from the video frame grabber. To verify the relative importance of x-ray quantum noise, the NPS of the imaging system was measured with the lens of the T.V. camera not focused on the plane of the x-ray detector (i.e., the metal plate/phosphor screen). When the lens of the camera is not in focus, higher spatial frequency components of both the signal and the (x-ray quantum) noise generated in the x-ray detector are not transferred by the imaging system. Therefore, noise contributions from the x-ray detector (i.e., x-ray quantum noise) will be eliminated, and only shot noise, Johnson noise, preamplifier noise, and digitization noise will remain. Except for the defocusing of the lens of T.V. camera, the calculation of the NPS was made identically to that described previously. Only one measurement of the NPS was performed with the lens of the camera not in focus for each x-ray beam.

2.3.3 Optimal magnifications

Using the MTF and NPS results from previous studies\textsuperscript{24} for portal films, previously measured source size data,\textsuperscript{27,28} and the MTFs and NPS results for the T.V. camera-based imaging system presented in this paper, the optimal magnifications were calculated as follows. For portal films, radial symmetry was assumed, and the optimal magnifications were calculated using the equivalent passband (i.e., the area under $MTF^2$) and Eq. (3). The data for the ortho-M tungsten plate detector described in Ref. 24 was chosen to represent typical portal films. For the portal imaging system, the optimal magnifications were calculated using the two-dimensional equivalent passband and Eq. (2). Since the MTF's of the T.V. camera-based imaging system were measured only in the di-
rections parallel and perpendicular to the camera scan lines, the two-
dimensional MTF was estimated by a bilinear interpolation of the MTF data
measured along the orthogonal axes.

The equivalent passband was calculated for magnifications ranging be-
tween 1.0 and 3.0. The optimal magnification, $M_{\text{signal}}$, occurs where the
equivalent passband curves reach a maximum. The quantity $SNR^2/(qM^2)$ was
calculated for the same range of magnifications. The photon fluence at unit
magnification, $q$, was estimated using the fluence-to-dose conversion tables of
Rogers. However, since the photon fluence is constant, it will not affect the
shape of the optimization curves and, therefore, the determination of the optimal
magnification. The maxima in the SNR transfer curves indicate the optimal
magnifications ($M_{\text{SNR}}$) for signal-to-noise transfer. The computations were per-
formed on a Macintosh IIx computer (Apple Computer, Inc., Cupertino, Ca) and
a Sun Sparcstation 2 (Sun Microsystems Inc., Sunnyvale, Ca).

2.3.4 Visual effect

The system MTFs used in our optimization calculations were measured
using techniques which minimize the effects of the sampling interval (i.e., the
pixel size). However, when acquiring images at different magnifications, both
the radiographic magnification and the pixel size, at the object plane, will
change. Both of these effects may influence the visual appearance of the ob-
ject. Therefore, to isolate the visual effects of radiographic magnification from
those of pixel size, the following image acquisition technique was adopted.

Images of a 3 mm thick star test pattern were acquired at radiographic
magnifications of 1.0 and 2.0 (a radiographic magnification of 2.0 corresponds
to the optimal magnification for signal transfer). Both images were acquired
using identical one monitor unit irradiations from a 6 MV x-ray beam, and the
radiographic magnification of the image was changed by moving the object closer to the source while leaving the position of the imaging system fixed. Therefore, the dose to the x-ray detector remained constant. However, when the first image was acquired (at $M=1.0$), the camera was operated in $1024\times1024$ acquisition mode while the camera was operated in $512\times512$ acquisition mode when the image was acquired at a magnification of 2.0. It should be noted that we used a non-standard T.V. camera where each video line was acquired in 63.5 $\mu$s, regardless of whether the camera was operating in $1024\times1024$ or $512\times512$ acquisition modes. Furthermore, the video digitizer sampled each video line 1024 times in both $1024\times1024$ and $512\times512$ acquisition modes. In the $512\times512$ mode, two sampled pixels were averaged together before display. The digitization rate was constant for both acquisition modes. Therefore, any visual changes in the appearance of the star pattern could be attributed entirely to the effects of radiographic magnification, and not to differences in pixel size at the object plane or to differences in the operating characteristics of the T.V. camera or video digitizer for the two acquisition modes.

2.4 Results

2.4.1 MTF

Care must be taken when measuring the MTFs of a T.V. camera-based system because a number of factors affect the MTF. These factors include: (i) the bandwidth of the T.V. camera preamplifier; (ii) lateral charge migration in the target of the T.V. camera; and, (iii) release of trapped charges by the lead oxide target of the T.V. camera. Care must be taken to ensure that these factors have the same influence on the MTF measurements as they do on the clinical images.
Figure 2.3 shows LSF’s of the system, measured with the slit oriented parallel and perpendicular to the camera scan lines, before and after adjustment of the bandwidth of the camera preamplifier. Figure 2.3a shows that adjustments of the bandwidth of the preamplifier has little effect on the shape of the LSF measured when the radiation slit is oriented horizontally so that it is parallel to the scan lines of the T.V. camera (i.e., the LSF is measured perpendicularly to the horizontal scan of the T.V. camera). Fig. 2.3b shows that when the slit is oriented perpendicular to the scan lines of the T.V. camera (i.e., the LSF are measured parallel to the horizontal scan of the T.V. camera), the adjustments of the bandwidth of the preamplifier change the shape of the LSF drastically. Since pixels are sampled at a high rate in the horizontal direction (~MHz) but at a much lower rate (~kHz) in the vertical direction, changes in the bandwidth of the preamplifier only affect pixels acquired in the horizontal direction.

The tails of the LSF are difficult to measure accurately when using T.V. camera-based imaging systems. The LSF’s are measured using film by making many different exposures and by piecing together portions of the LSF from many films. The different exposures (on different films) ensure that both the centre and the tails of the LSF are in the linear region of the D versus logE curve. However, such an approach is not possible when using a T.V. camera. Saturation of the video amplifier leads to unpredictable results. Furthermore, dark current, lateral charge migration and the release of trapped charges in the target of the T.V. camera are all time-dependent phenomena, and thus images acquired using different exposure times cannot be pieced together as is done with film.
Figure 2.3: Effect of camera bandwidth on LSF measurement for a 6 MV beam. These LSF's have been measured with the radiation (a) slit parallel and (b) perpendicular to the camera scan lines. The solid lines represent the LSF obtained with the camera bandwidth recommended by the manufacturer. A poor adjustment of the bandwidth causes the frequency-response of the T.V. camera to be non-uniform, and frequency-dependent phase shifts may occur, resulting in ringing and streaks in the video image. Note that modifying the camera bandwidth has little effect on the shape of the LSF with the slit parallel to the camera scan lines.

To demonstrate how accumulation time (i.e. exposure time) changes the resulting video signal, Fig. 2.4 shows the LSF measured using delayed readout of the T.V. camera. These LSF's have been obtained from identical 6 MV x-ray
irradiations, but delays (from 0 to 60 s) have been introduced between the end of the x-ray irradiations and camera readout. Figure 2.4 shows that the width of the LSF increases as the delay between irradiation and readout increases. We attribute this change in the width of the LSF to be due to lateral charge migration. In addition, the area under the LSF also increases with accumulation time. Since the LSFs were obtained from the subtraction of slit images and background images, the contribution of the dark current to the LSF's should have been eliminated. Therefore, we attribute this increase in the area under the LSF to be due to the release of trapped charges$^{39,40}$ within the target of the camera. In the lead oxide target of the camera tube, impurities cause a large number of charges (i.e., electrons and holes) to be trapped between the valence and conduction bands. Trapped charges can be released through either photo-excitation or thermal excitation. The electric field inside the target of the T.V. camera then causes these released charges to migrate towards the electrodes of the target, thus preventing the recombination of electrons and holes.$^{39}$ The time-dependent increase in the area under the LSF shown in Fig. 2.3 can be attributed to charges released through thermal excitation since the LSF images were acquired with identical x-ray irradiations. Consequently, the shape of a LSF measured with a 1 s irradiation will be different from that of a LSF measured with a 10 s irradiation. Since the shape of the LSF depends on accumulation time, we cannot piece together a LSF from multiple exposures to obtain an accurate estimate of the tails of the LSF, as is done with screen/film.$^{30}$ Therefore, the approach described in section 2.3.1 was used to estimate the tails of the LSF and normalize the system MTF.
Figure 2.4: Effect of delayed camera readout on LSF for a 6 MV beam with a slit beam of x rays perpendicular to the TV camera scan lines. As the delay increases, the area under the LSF and the width of the LSF increase. This phenomenon cannot be attributed to dark current since subtraction of background images eliminates the offset caused by dark current. This extra signal may correspond to the release of trapped charges and the lateral migration of charges. Similar behavior is observed for a slit parallel to the camera scan lines and for an 18 MV x-ray beam.

Figure 2.5 shows the MTF measured for the imaging system [MTF\textsubscript{sys}] (i.e., for the copper plate/phosphor screen detector + mirror + lens + T.V. camera assembly) and the detector MTF (i.e. the copper plate/phosphor screen detector alone) for 6 MV and 18 MV beams. The error bars represent the maximal deviation from the average MTF, which was obtained from two (18 MV) to three (6 MV) independent measurements. Since the detector MTFs were measured only once, no error bars are presented for these data. The frequencies at which the MTF's are 0.5 (f\textsubscript{0.5}) and 0.1 (f\textsubscript{0.1}) are listed in Table 2.1.

Comparison of the detector and system MTFs show the extent to which the optical and electronic components of the imaging system degrade spatial resolution. Figure 2.5 shows that the system MTF measured in the horizontal direction are greater than those measured in the vertical direction. Differences in MTF depending upon the orientation have been reported previously for video
Figure 2.5: The detector (solid) and the system MTF's when irradiated by 6 and 18 MV x-ray beams. The system MTF's have been renormalized so that they are equal to the detector MTF's at a spatial frequency of 0.1 mm⁻¹. The error bars represent the maximum deviation of individual measurements from the mean. Since the detector MTF's were measured only once, no error bars are shown.

However, in those reports the MTF's in the vertical direction are greater. The reasons for the dependence of the MTF on the orientation of the slit are not clear.
<table>
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<tr>
<th></th>
<th>6 MV</th>
<th>18 MV</th>
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</thead>
<tbody>
<tr>
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<td>$f_{0.5}$ (mm$^{-1}$)</td>
<td>$f_{0.1}$ (mm$^{-1}$)</td>
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Table 2.1: Summary of MTF measurements. The frequencies at which the MTFs are 0.5 ($f_{0.5}$) and 0.1 ($f_{0.1}$) are shown for the measurements performed with the 6 MV and 18 MV x-ray beams. The source MTF’s for our Varian 2100c linear accelerator were obtained from Jaffray et al.\textsuperscript{27,28} and the tungsten plate/film data were obtained from Munro et al.\textsuperscript{24}

2.4.2 NPS

Figure 2.6 shows slices through the two-dimensional NPS measured for our T.V. camera-based portal imaging system. The slices have been chosen so that the NPS in the directions parallel to the camera scan lines [NPS(0,$f$)] and perpendicular to the camera scan lines [NPS($f$,0)] are shown. The NPS measured with the lens of the camera in focus have been smoothed for this plot in order to provide a clearer contrast with the NPS measured with the lens not in focus. Repeated measurements of the NPS shows an uncertainty ranging between 23% and 68% while the uncertainty in a single measurement ranged between 12% and 18%. Figure 2.6 shows that the NPS obtained with the lens in focus (thick lines) is essentially identical to the NPS measured when the lens is not in focus (thin lines). As described in section 2.3.2, the only difference between these curves is the presence or absence of x-ray quantum noise. Since both NPS curves are identical, x-ray quantum noise must be a small contribution to the overall NPS. In addition, Fig. 2.6 shows that the 2-dimensional NPS is not flat since the NPS in the direction perpendicular to the camera scan lines is higher than the NPS in the direction parallel to the camera scan lines. Figure 2.6 also shows that the NPS measured for the 18 MV beam are similar to those measured for the 6 MV x-ray beam.
Figure 2.6: Slices through the two-dimensional noise power spectra for the imaging system for the 6 MV and 18 MV x-ray beams. The NPS in the orthogonal directions parallel (solid) and perpendicular (dotted) to the scan lines of the TV camera are shown. The NPS have been measured with the lens of the TV camera focused (thick lines) and unfocussed (thin lines). The NPS obtained with the focused lens are the average of three measurements and have been filtered using a 3x3 median filter to distinguish them visually from the NPS results using the defocused lens. The sampling interval is 0.67 mm.

Since x-ray quantum noise is a negligible contributor to the NPS, the recorded noise may be produced by: (i) shot noise (i.e., photo-electron quantum noise); (ii) Johnson noise; (iii), camera preamplifier noise; or (iv) digitization noise. To verify if shot noise contributes significantly to the system noise, we
have determined the noise in the video signal for different x-ray intensities. A plot of noise versus signal from a shot-noise limited system would be expected to have a slope of 0.5 on a log-log plot since shot noise obeys Poisson statistics (i.e., the noise is proportional to the square-root of the signal in a Poisson process).\textsuperscript{41} Figure 2.7 shows a log-log plot of the noise in the video signal, represented by the standard deviation in the 8×8 central pixels of noise images, as a function of the mean video signal obtained with various doses from the 6 MV x-ray beam (represented by the mean pixel value in this central area). It shows that the noise in the video signal remains constant for all video signals (i.e., there are no signal-dependent noise contributions). The results in Fig. 2.7 suggest that noise added by the T.V. camera electronics (i.e., Johnson noise and camera preamplifier noise) or possibly noise added by the video digitizer are the dominant noise sources for this portal imaging system.

These results are important for another reason. There is an assumption inherent in Eqs. (2) and (3) that the NPS is independent of exposure. However,
for some imaging systems this is not case. For these systems, any estimate of
the optimal magnification would require that the NPS be measured under a va-
riety of exposure conditions to determine how the NPS will change with magni-
fication (i.e., exposure). The results shown in Fig. 2.7 show that the shape of the
NPS for our portal imaging system is independent of image magnification and
exposure. Hence, only one measurement of the NPS is sufficient to determine
the shape of the NPS for all conditions that our portal imaging system is likely to
encounter.

2.4.3 Optimal magnifications: portal films

Figure 2.8 shows the result of our optimization calculations for portal films
for 6 MV (solid curves) and 18 MV (dashed curves) x-ray beams. Figure 2.8a
was obtained using the signal transfer method [the area under the squared
overall MTF given by Eq.(1)] while Fig. 2.8b was obtained with the SNR transfer
method [Eq.(3)]. Using the error propagation technique of Bevington and
estimated uncertainties of ±3% in each data point for the portal film MTFs (as re-
ported in Ref. 24), uncertainties of ±2.5% in each data point for the focal spot
MTFs (as reported in Ref. 28) and uncertainties of ±6% in each data point for
the portal film NPS (as reported in Ref. 24), we estimate the uncertainty in the
optimization curves (i.e., the integral of these data) to be ±0.02% and ±0.65%
for signal and SNR transfer, respectively. All curves display a well-defined
maximum at a radiographic magnification close to unity. The low values for the
optimal magnification obtained with the signal transfer approach \( M_{\text{signal}} \) and
the SNR transfer approach \( M_{\text{SNR}} \) (See Fig. 2.8 and Table 2.2) are due to the
high spatial resolution (i.e., large system MTF) of portal films with respect to the
MTF of the focal spot. Since the size of the focal spot dominates the overall
resolution, a small magnification is needed to limit the blur introduced by the fo-
Figure 2.8: Signal transfer (a) and signal-to-noise transfer (b) as a function of radiographic magnification for portal films (a tungsten plate/Ortho-M film system) irradiated with a 6 MV (solid) and an 18 MV (dashed) x-ray beams. These curves were obtained for a Varian 2100c medical linear accelerator.

cal spot. This result, which is widely recognized by clinical experience, shows that the highest quality portal films are obtained when the film is in close contact with the patient. In addition, these low optimal magnifications confirm the observations of Droege and Cytacki.21
2.4.4 Optimal magnifications: T.V. camera-based imaging system

The optimization curves obtained with the measurements performed with the T.V. camera-based portal imaging system are shown in Fig. 2.9. Using the error propagation technique of Bevington\textsuperscript{42} and an uncertainty of ±12\% in each data point for the system MTFs, an uncertainty of ±2.5\% in each data point for the focal spot MTFs (as reported in Ref. 28), and assuming a worst-case uncertainty of ±50\% in each data point for the system NPS, we estimate the uncertainty in the optimization curves (i.e., the integral of these data) to be no more than ±0.3\% for the signal transfer curves of Fig. 2.9a, and less than ±0.7\% for the SNR transfer curves of Fig. 2.9b. Each optimization curve shows a broad peak. The maximum of each curve indicates the optimal magnification.

Figure 2.9a shows that the image magnifications which optimize signal transfer, $M_{\text{signal}}$, range between 2.0 and 2.3 (see Table 2.2). Our optimal magnifications are somewhat smaller than the optimal magnification of 2.17 reported by Swindell and others\textsuperscript{22} since the full-width at half-maximum (i.e. $2\times f_{0.5}$) of our focal spots and of our system LSF differ greatly from the values that were used in their calculations. Our values for $M_{\text{signal}}$ imply large patient-to-detector distances of between 1.0 m and 1.3 m for a typical source-to-patient distance of 1.0 m. Such large patient-to-detector distances are impractical in the clinical environment because of the bulk of the optical path required for T.V. camera-based portal imaging systems. Furthermore, the x-ray detector would have to be 70 cm on a side to maintain a field of view (FOV) of $35\times35\text{cm}^2$ for an object located 100 cm from the x-ray focal spot.

The quantity $\text{SNR}^2/(q\beta P)$, calculated for a range of magnifications [Eq. (2)], is shown in Fig. 2.9b. The maxima in the SNR transfer curves, which correspond to the optimal magnification, $M_{\text{SNR}}$, occur between $M=1.5$ and $M=1.7$ for our T.V. camera-based imaging system, depending on x-ray beam energy (See
Figure 2.9: Signal transfer (a) and signal-to-noise transfer (b) as a function of radiographic magnification for our T.V. camera-based imaging system. The optimization curves are shown for measurements performed with a 6 MV x-ray beam (solid) and an 18 MV x-ray beam (dotted). These curves were obtained for a Varian 2100c medical linear accelerator.

Fig. 2.9b, and Table 2.2). The difference in optimal magnification is due primarily to differences in the system MTF's for the 6 MV and 18 MV x-ray beams.
<table>
<thead>
<tr>
<th></th>
<th>6 MV M_{MAX}</th>
<th>6 MV M_{MIN}</th>
<th>18 MV M_{MAX}</th>
<th>18 MV M_{MIN}</th>
</tr>
</thead>
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</tr>
<tr>
<td>Tungsten plate</td>
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<td>1.01</td>
<td>1.01</td>
<td>1.01</td>
</tr>
</tbody>
</table>

Table 2.2: Optimal magnifications for our video-based portal imaging system and for portal films.

2.4.5 Visual effect

The two images, acquired as described in section 2.3.4, are shown in Fig. 2.10. Fig. 2.10a has been acquired at unit magnification using the 1024×1024 acquisition mode while Fig. 2.10b has been acquired at a magnification of 2 using the 512×512 acquisition mode. Thus, the pixel size, at the object plane, is identical in the two images. To eliminate gradual changes in the average brightness of the image caused by lens vignetting, the display of the images has been optimized by use of a display equalization algorithm. Nevertheless, the two images have been displayed using identical window and level settings. A visual comparison of Figs. 10a and 10b reveals that the test pattern is sharper and much better resolved when the larger magnification is used to form the image. Clearly, operating the T.V. camera-based imaging system at the optimal magnification results in a visual improvement in image quality.

2.5 Discussion

While the goal of our study was to determine the optimal magnification for portal imaging systems, our measurements revealed a number of other important issues. Figure 2.3b shows that the LSF is asymmetric, even when the bandwidth of the preamplifier is set to the factory-recommended settings. Since
the response of the preamplifier influences the shape of the LSF, the spatial resolution measured in the direction parallel to the camera scan lines will be influenced not only by the physical processes which lead to signal spread in the imaging system (e.g., light spread in the phosphor screen; spherical aberrations in the lens; charge migration in the target of the T.V. camera; electron beam size of the T.V. camera...), but also by the design and adjustment of the camera electronics. Poor adjustment of the camera electronics can result in asymmetric LSFs (which have been reported previously\cite{36}) and large negative-going-tails in the LSF. The interpretation of MTFs that have been calculated from asymmetric LSFs is difficult,\cite{44} and the handling of LSF with negative-going tails is undefined. Therefore, for intercomparison purposes, we recommend that the MTF of camera-based imaging systems should be measured with the slit of x rays parallel to the camera scan lines since our measurements show that the bandwidth of the camera preamplifier does not result in asymmetric LSFs or these with negative-going tails. Furthermore, to limit the effects of time-dependent phenomena which arise from the non-standard operation of a T.V. camera (i.e., dark current, charge migration and release of trapped charges), we recommend that the measurement of the MTF of a T.V. camera-based imaging system be performed such that the images are acquired with irradiation times similar to the intended clinical exposure time.

The optimization curves of Fig. 2.9 show that, in the case of the T.V. camera-based portal imaging system, the optimal magnifications obtained from SNR transfer are lower than those determined by signal transfer. These lower values are due to the fact that the SNR transfer optimization accounts for the transfer of noise in the imaging system.
Figure 2.10: Images of a star pattern (spatial resolution test tool) acquired with our T.V. camera-based imaging system using magnifications of (A) 1.0 and (B) 2.0. Both images were obtained with 1 monitor unit irradiations from a 6MV x-ray beam with the position of the object being used to adjust magnification. The operating mode of the T.V. camera was adjusted so that the size of the pixels at the object plane was identical in both images. Both images have been displayed using identical display contrasts, and non-uniform brightness in the images (caused by lens vignetting) has been eliminated by the use of display equalization.43

Figure 2.9 shows that the optimal magnification of the portal imaging system ranges from 1.5-2.3, depending upon whether signal or signal-to-noise transfer is being maximized. In contrast, Fig. 2.8 shows that the optimal magnification for portal films is 1.0. These results suggest that portal films and T.V. camera-based portal imaging systems should be used differently. While portal films should be placed as close to the patient as is practically possible, the T.V. camera-based portal imaging systems should be kept at some distance (50-130cm) from the isocentre. Fortunately, the exact placement of the T.V. camera-based portal imaging systems is not critical. Figure 2.9 shows that the peaks of the optimization curves are rather broad. Therefore, a small deviation from the optimal magnification has a small effect on portal image quality. For example,
the operation of our portal imaging system at a magnification of 1.3 results in a modest decrease in $\frac{SNR^2}{(qM^2)}$ (11%) compared to that at the $M_{SNR}$; thus, our T.V. camera-based imaging system can be operated at a magnification of 1.3 with a relatively small penalty in terms of image quality. The broad peaks of the SNR transfer curves also show that thick objects (i.e., patients) are imaged well. When a patient is imaged, the anatomic structures located closer to the x-ray focal spot have a larger magnification than the structures located closer to the imaging system. For example, for an imaging system placed 130 cm away from the x-ray source, and a 25 cm thick patient (positioned so that the centre of the patient coincides with the isocentre of the linear accelerator), the anatomical structures within the patient will be imaged with a magnification ranging between 1.15 and 1.49. This range of magnifications corresponds to changes in $\frac{SNR^2}{(qM^2)}$ of less than 22% with respect to that at the $M_{SNR}$ for the 6 MV beam, and less than 10% with respect to that at the $M_{SNR}$ for the 18 MV beam.

One consideration is the field of view (FOV) of the portal imaging system at the isocentre. Increasing the radiographic magnification decreases the FOV and thus necessitates a bulkier imaging system. For example, assuming that we require a FOV of 35×35 cm$^2$ at the isocentre$^{45}$ while operating at a magnification of 1.5, the area of the copper plate/ phosphor screen detector would have to be 53×53 cm$^2$. The additional bulk of a video-based portal imaging system designed for this magnification would probably decrease the practicality of such a system (i.e., increasing the FOV by increasing the area of the x-ray detector reduces the optical coupling efficiency and thus the image SNR), even though it would be operating optimally. Therefore, the developers of portal imaging systems must strike a balance between practical considerations and the improved image quality obtained by optimizing magnification. Nevertheless, Fig. 2.10 shows that improvements in image quality when using magnification can be
visually striking. Thus, the optimal magnification of a given imaging system and
accelerator combination should be known and the system should be operated
at this magnification if there are no practical difficulties in doing so.

The results of our optimization calculations ignore the effect of scattered
radiation on image quality. It is well-known that x-ray scatter decreases con-
trast, and that the amount of scattered radiation which is intercepted by an x-ray
detector decreases as the distance between the object and the x-ray detector
decreases (i.e., magnification increases). Thus, one expects that any estimate
of optimal magnification that includes the effects of x-ray scatter would result in
optimal magnifications which are larger than those reported in this paper. Fortu-
nately, scatter is a small problem at megavoltage energies. Scatter fractions
have been reported in the literature for the 6 MV x-ray beam of a Varian 2100c
linear accelerator.\textsuperscript{46} For a 30×30 cm\textsuperscript{2} x-ray field at the isocentre impinging on
a 17 cm thick slab of poly-methyl-methacrylate, the scatter fraction is 0.34 in a
contact geometry (M = 1), and for the optimal magnifications recommended in
this paper for the T.V. camera-based portal imaging system, the scatter fractions
are less than 0.15.\textsuperscript{46} Since the scatter fractions at megavoltage energies are
small, the effect of scatter on the image quality obtained with T.V. camera-based
portal imaging systems ranges from negligible to moderate.\textsuperscript{46} Therefore, a
more rigorous optimization theory, which would account for the effect of scat-
tered radiation on image quality, would predict optimal magnifications which are
only slightly higher than those reported in this paper.

Improving the signal and noise transfer characteristics of an imaging
system may modify the optimal magnification. The signal and SNR transfer
theories show that the optimal magnification decreases if the signal and noise
transfer characteristics of an imaging system are improved. Currently, research
groups are investigating the use of solid-state technology to develop new portal
imaging systems. Already, the device based on amorphous silicon detector proposed by Antonuk et al.\textsuperscript{19} shows promise of significant improvements in terms of MTFs and NPS. These imaging systems would be expected to have optimal magnifications closer to unity than those presented in this paper.

2.6 Conclusions

We have measured the MTF and the NPS of our imaging system in order to estimate the optimal radiographic magnifications required for our imaging chain (which includes the x-ray source and the imaging system). The system MTF measurements have demonstrated that the electronic and optical components of our imaging system degrade system resolution, and that these measurements should be performed in such a way that the potentially deleterious effects of the electronic bandwidth, charge migration in the target, and lag of the T.V. camera are minimized. Data accumulation times similar to those used to acquire clinical images should be used to minimize visual effects from charge migration and camera lag. Furthermore, MTFs measured in the direction perpendicular to the camera scan lines will minimize the non-linear effects of the T.V. camera bandwidth. The NPS measurements show that noise from the T.V. camera electronics is a major contributor to the noise in images acquired with our imaging system.

Magnifications which maximize signal transfer ($M_{\text{signal}}$) and signal-to-noise ratio transfer ($M_{\text{SNR}}$) were calculated for both our T.V. camera-based imaging system and portal films. For our T.V. camera-based imaging system, the $M_{\text{signal}}$ vary between 2.0 and 2.3 while the $M_{\text{SNR}}$ range between 1.5 and 1.7. For a portal film system, both the $M_{\text{signal}}$ and the $M_{\text{SNR}}$ are slightly above 1.0. Thus, for portal films, the best image quality will be obtained without magnification (i.e., $M = 1$).
References


Chapter 3: EVALUATION OF A HIGH DENSITY SCINTILLATING GLASS FOR PORTAL IMAGING¹

Abstract

One of the main factors which limits the performance of T.V. camera-based portal imaging systems is the poor light-collection efficiency of the lens and T.V. camera. An x-ray detector which produces more light per incident x ray would help overcome this limitation. We have been evaluating a high density (3.8 g/cm³), thick (12 mm) glass scintillator for its suitability as an x-ray detector for T.V. camera-based portal imaging systems. The light output and spatial resolution of the glass scintillator have been compared to those of a copper plate/phosphor screen detector using radiographic film and the T.V. camera of our portal imaging system. The film measurements show that the light output of the glass scintillator is 82% of that of the copper plate/phosphor screen while the T.V. camera measurements show that this value is 48%. A theoretical model of light transport described in this chapter suggests that this discrepancy is due to refraction at the glass-air interface. Our measurements of the modulation transfer function (MTF) show that the spatial resolution obtained with the glass scintillator is similar to that obtained with the copper plate/phosphor screen. However, the spatial resolution obtained with the glass scintillator decreases as the angle of x-ray incidence increases; this decrease, which is not observed for the copper plate/phosphor screen detector, is due to the large thickness of the glass scintil-

¹ This chapter was published in Medical Physics, 23 (3): 401-406 (1996). The title of the article is "Evaluation of a high density scintillating glass for portal imaging" by J.-P. Bissonnette and P. Munro.
lator. Due to the limited light output and the variable spatial resolution, the transparent glass scintillator, in its current form, is not suitable for portal imaging.
3.1 Introduction

A wide variety of on-line portal imaging systems are being developed to verify radiation field placement in radiation therapy.\(^\text{1-4}\) It has been shown that, for T.V. camera-based portal imaging systems, the low amount of light reaching the T.V. camera limits the performance of these imaging systems.\(^\text{5}\) This limitation can be overcome by: (i) increasing the light collection efficiency of the system's optical chain; or, (ii) increasing the light output of the x-ray detector which is used in the portal imaging system. Current T.V. camera-based portal imaging systems use a metal plate/phosphor screen combination as the x-ray detector.\(^\text{6-18}\)

Recently, alternative designs of the metal plate/phosphor screen-based detectors have been proposed,\(^\text{19}\) and new x-ray detectors have become available commercially for industrial radiography.\(^\text{2,20}\) A high density (3.8 g/cm\(^3\)) glass scintillator (Industrial Quality Inc., Gaithersburg, Md) has been developed for imaging with high energy x rays.\(^\text{2,20}\) This glass scintillator has been promoted as a high resolution, high quantum efficiency alternative to phosphor scintillators.\(^\text{2,20}\) In this chapter, we compare the light emission and spatial resolution characteristics of the glass scintillator with those of a copper plate/phosphor screen combination.\(^\text{13}\) In addition, we present theoretical calculations demonstrating some of the differences in light transport between the two x-ray detectors.

3.2 Background on optical light transport

Light is transported in different ways for the two detectors under study. Since the glass scintillator is transparent, the light produced isotropically within the glass scintillator is subjected to refraction at the exit surface of the glass, as depicted in Fig. 3.1. Refraction modifies the angular distribution of the light exiting the glass scintillator since it causes light exiting the glass to be bent away
Figure 3.1: Transport of optical photons inside the x-ray detectors. In the phosphor, optical photons follow random paths, and the angular distribution of light exiting the phosphor is independent of direction. In the glass scintillator, light rays exiting the glass are bent away from the central axis because of refraction.

from the central axis. In contrast, the scintillating layer of the copper plate/phosphor screen detector is a turbid material. As depicted in Fig. 3.1, the light generated within the phosphor randomly scatters many times before being absorbed in the phosphor material, or being emitted from the phosphor layer (in a random direction). Because of the random trajectories followed by optical photons in the phosphor screen, refraction events which occur in the screen do not significantly affect the angular distribution of light exiting the screen, which is known experimentally to be almost independent of direction. Therefore, the copper plate/phosphor screen combination must be considered a non-refractive light emitter while the glass scintillator is a refractive emitter.

The copper plate/phosphor screen detector can be considered as a continuum of point-like sources distributed over a surface while the glass scintillator
can be considered as a continuum of point-like sources distributed over a volume. However, the description of a two-dimensional or a three-dimensional light source, especially when reflection is involved, is difficult mathematically. We have assumed that, for our purposes, the light sources from the copper plate/phosphor screen detector and the glass scintillator can be modeled as point sources. While this is a simplification, it does allow us to draw some conclusions about the effect of refractive surfaces on the luminous flux collected from x-ray detectors, which is the purpose of these calculations.

For a point-like source, the luminous flux collected over the solid angle subtended by a light sensor, \( \Omega \), when refraction is not involved, can be calculated by integrating the function\(^{21-26} \)

\[
P(z) = \int \frac{k}{4\pi} \cos \theta \, d\Omega = \int \int \frac{k}{4\pi} \frac{z}{(x^2 + y^2 + z^2)^{3/2}} \, dx \, dy. \tag{1}
\]

In Eq. (1), \( k/4\pi \) is the luminous intensity (in lumen/steradian) of the source, \( \theta \) is the angle between the normal of the light sensor and the direction of the incident ray, \( x \) and \( y \) are the two-dimensional coordinates of a light sensor of size \( 2X \times 2Y \), and \( z \) is the distance between the light source and the light sensor.

For a refractive emitter, Eq. (1) cannot be used directly to estimate the amount of light collected at some distance from the exit surface of the detector since more light exits the refractive emitter at large angles with respect to the central axis than would occur with a non-refractive emitter.\(^{27} \) The effects of refraction can be incorporated into Eq. (1) merely by adjusting the limits of integration (\( \pm X \) and \( \pm Y \)) so that the apparent change in size of the light sensor caused by refraction is accounted for. The apparent size of the light sensor when viewed through a glass-air interface can be determined by using a relationship between the angle of emission of a light ray with respect to the perpendicular to the glass-air interface, \( \theta_1 \), and the distance from the central axis where the light ray strikes
the light sensor, \( r \) (see Fig. 3.1). This relationship can be derived by use of Snell's law and trigonometric identities (see Appendix A),

\[
r = z \frac{n \sin \theta_1}{\sqrt{1 - n^2 \sin^2 \theta_1}} + d \tan \theta_1,
\]

(2)

where \( d \) is the distance between the light source and the glass-air interface, \( z \) is the distance between the glass-air interface and the light sensor, and \( n \) is the index of refraction of the glass. One cannot rearrange Eq. (2) to isolate \( \theta_1 \). However, Eq. (2) can be solved numerically so that the maximum angle of emission, \( \theta_{max} \), that light from the glass scintillator can be emitted and still reach a light sensor with dimensions \( 2X \times 2Y \) can be found. For a square light sensor, the corrected limits of integration for Eq. (1) become \( X = Y = d \tan \theta_{max} \).

Note, as mentioned above, that Eqs. (1) and (2) are restricted to infinitesimally small light sources located on the central axis of a light sensor. A comprehensive model would account for the physical size of the source by integrating Eq. (1) over the surface (e.g., copper plate/phosphor screen) or the volume (e.g., thick, transparent crystal) described by the light source.

3.3 Materials and methods

3.3.1. X-ray detectors

Two x-ray detectors have been studied: (i) a copper plate/phosphor screen detector; and, (ii) a new scintillating glass detector. For simplicity, the copper plate/phosphor screen detector will be referred to hereafter as the "phosphor detector". The phosphor detector consists of a 1 mm copper plate to which has been bonded a 360 mg/cm\(^2\) phosphor layer. This thickness is typical of the phosphor screen thicknesses encountered in video-based portal imaging systems.\(^{1-4}\) The phosphor layer consists of small particles (10-25 \( \mu \)m) of Gd\(_2\)O\(_2\)S:Tb
held together by an organic binder. The physical density of Gd$_2$O$_2$S:Tb is 7.4 g/cm$^3$, but the packing density of the phosphor layer is only 50%, yielding a physical density for the phosphor layer of approximately 3.7 g/cm$^3$. Thus, the phosphor layer is approximately 1 mm thick. Because of optical attenuation, only 40% of the light generated inside the phosphor detector escapes from the exit surface [D. Trauernicht, Eastman Kodak (private communication)].

The second detector consists of a 12 mm thick, high-density (3.8 g/cm$^3$) glass which has been doped with Gd$_2$O$_2$S:Tb to yield a transparent scintillator. Light emitted from the surface of the glass is refracted away from the central axis because of the glass-air interface. The fraction by weight of Gd$_2$O$_2$S:Tb inside the glass scintillator is 3%, which results in 137 mg/cm$^2$ of phosphor inside the glass [T. Jones, Industrial Quality Inc. (private communication)].

3.3.2. Light output measurements

The light output of the glass scintillator and that of the phosphor detector were measured on exposure to the 6 MV x-ray beam from a linear accelerator (Varian Clinac 2100c). In the following sections, the term "light output ratio" refers to the ratio of the light output obtained from the glass scintillator to that from the phosphor detector at the same distance from the x-ray detectors along the central axis. Light output ratios were determined for two geometries. In the first geometry, a T.V. camera was used to measure the light output ratio of the two detectors, which were both located at a distance of 150 cm from the T.V. camera. In the second geometry, radiographic film was placed in direct contact with the x-ray detectors to measure the light output ratio at the exit surface of the detectors.

One set of light output ratios was measured by digitizing the video signal from the T.V. camera when the x-ray detectors were exposed to 2 monitor unit ir-
radiations. [For our accelerator, 1 monitor unit delivers a 1 cGy dose to the isocentre for a 10×10cm² field at a depth of 5 cm in water.] The T.V. camera, which was operated in target accumulation mode, has a linear response to changes in accelerator output. Three images were acquired, 10 s apart, for each x-ray detector to reduce the effect of any fluctuations in the x-ray fluence delivered by the linear accelerator. The measurements were performed with a 15×15cm² x-ray field at the plane of the detector. In addition, the illuminated region, as viewed by the camera, was restricted by a 13×13cm² (~200×200 pixels) area by an optical aperture placed between the x-ray detector and the T.V. camera. The “raw” pixel values were obtained by averaging the central 10×10 pixels of the three images. Choosing a small number of pixels minimized variations due to vignetting and shading. A “dark” pixel value was calculated from four 10×10 pixels regions which were located at least 50 pixels (3.3 cm) from the edge of the optical aperture. The average dark pixel values were subtracted from the raw pixel values to yield the net light output values [analog-to-digital converter (ADC) units]. This correction accounted for the effects of beam-discharge lag, dark current, and drift.

To measure the amount of light at the exit plane of the two detectors, measurements were made using Kodak EM-1 radiographic film, with the emulsion side of the film placed in direct contact with the x-ray detector. The detector and film were placed in a light-tight container and exposed using an irradiation of 5 monitor units to yield films with optical densities of approximately 2.0. The films were processed in a Kodak M6B X-Omat film processor and the optical density of the resulting films was measured with a film densitometer (X-Rite, Grandville, MA, model 301). The measured optical densities were corrected for the non-linear response of the film before being used to calculate the light output ratio.
3.2.3 Light transport calculations

To assess the effect of refraction on the light output from an x-ray detector, Eq. (1) was used to calculate the luminous flux at some distance from a point-like source under two situations. [Throughout this chapter, we will refer to the total luminous flux reaching a $1 \times 1 \text{mm}^2$ planar region as "luminous flux" (in lumen).] Equation (1) was first used to calculate the luminous flux ignoring the effects of refraction. Then, Eqs. (1) and (2) were used to calculate the luminous flux from a point source located within a high index of refraction material such as the glass scintillator (i.e., $n=1.629$). As described in Sec. 3.2, the limits of integration of Eq. (1) were modified, by use of Eq. (2), so that the effects of refraction at the glass-air interface were accounted for. The value of $\theta_{\text{max}}$ (i.e., the maximum angle of emission that light from the glass scintillator can be emitted and still reach the $1 \times 1 \text{mm}^2$ planar region) was determined by iterative evaluation of Eq. (2), using a range of values of $\theta_1$ while the other variables remained constant. $\theta_{\text{max}}$ was subsequently used to obtain the appropriate limits of integration for Eq. (1) (i.e., $\propto \propto d \tan \theta_{\text{max}}$). In the two situations described in this paragraph, the point-like sources were assumed to be of equal intensity.

3.3.4 Spatial resolution measurements

When the angle of x-ray incidence on the x-ray detector increases (due to beam divergence), the apparent size of the light source increases and the point-spread function becomes asymmetric, as depicted schematically in Fig. 3.2. When the x-ray detector is thin, as is the case with the phosphor detector, this effect is negligible. However, this effect can become significant in the case of the glass scintillator because of its large thickness. Note that x-ray divergence influences the apparent size of the light source in the radial direction only, and leaves the apparent size of the light source in the tangential direction unchanged.
Figure 3.2: When the glass scintillator is irradiated, light is emitted along the path of the electrons set in motion by incident x rays. If the x-ray beam is divergent, x rays travel laterally, thus producing an increase of the apparent size of the light source. This increase occurs in the radial direction only.

The modulation transfer function (MTF) of the portal imaging system (using the phosphor detector and the glass scintillator) was measured using standard techniques. Briefly, two steel blocks (60 cm thick) were clamped together to collimate the x rays from the linear accelerator to a narrow (0.4 mm wide) x-ray beam. The images obtained with this narrow x-ray beam generated the line-spread functions (LSF) which were Fourier-transformed to yield the MTF’s. The images were acquired at least 10 s apart so that the effects of T.V. camera lag were minimized, and exposure times similar to those for clinical images (~ 1 s) were used to ensure that the time-dependent phenomena associated with the operation of the T.V. camera had similar effects on both the MTF measurement and the clinical images. The MTF’s were measured twice and the results averaged to reduce experimental uncertainties.

Additional MTF measurements were performed with both detectors rotated by 11° with respect to the axis of the narrow x-ray beam (see Fig. 3.3) to examine the effects of oblique x-ray incidence (i.e., beam divergence) on the spatial resolution of the portal imaging system. Choosing this orientation allowed the measurement of the spatial resolution that would be obtained at the edge of a
Figure 3.3: Schematic diagram of the MTF measurement procedure for perpendicularly and obliquely incident x rays. The portal imaging system was rotated 11° with respect to the perpendicular of the x-ray beam axis, simulating the divergence for x rays at the edge of a 40×40 cm² field.

40×40 cm² divergent x-ray beam. Since the slit images were always viewed using the central portion of the T.V. camera, any additional degradation on in spatial resolution caused by the lens and T.V. camera (e.g., coma, curvature of field, electron beam landing, ...) as one moved from the centre to the edge of the field of view was eliminated. Thus, the measurement technique evaluated the effect of x-ray incidence angle on spatial resolution while holding all other parameters constant. The MTF’s measured with obliquely incident x rays were measured only once for each detector.

3.4 Results
3.4.1 Light output measurements

The results of the light output measurements, summarized in Table 3.1, indicate that the light output obtained from the glass scintillator is lower than that obtained from the phosphor detector in a contact geometry and at a distance of 150 cm. In addition, the light output ratio is larger in a contact geometry (0.82) than at a distance of 150 cm (0.45). This change in the light output ratio with respect to distance suggests that the angular distribution of light emitted from the phosphor detector and the glass scintillator are different.

3.4.2 Light transport calculations

Figure 3.4 shows that, in a contact geometry, refraction doesn’t affect the amount of light collected from point-like sources of equal intensities (i.e., the ratio of luminous fluxes is 1.0). However, as the distance increases, refraction reduces the luminous flux reaching the point of optical detection. In the limit (i.e., large distances), the ratio of the luminous flux from a refractive emitter to that from a non-refractive emitter approaches the inverse of the square of the index of refraction (i.e., the ratio of luminous fluxes is 0.38), in agreement with expectations. These calculations suggest that refraction at the glass-air interface is the reason why the light output ratios measured in the previous section change between contact and non-contact geometries. Note that the result of the calculations, shown in Fig. 3.4, are for a point source located on the axis of an optical detector. These calculations will overestimate the effect of refraction on the light

<table>
<thead>
<tr>
<th></th>
<th>Film measurement (relative units)</th>
<th>T.V. camera measurement (ADC units)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Phosphor detector</td>
<td>245±2</td>
<td>203.5±2.2</td>
</tr>
<tr>
<td>Glass scintillator</td>
<td>201±2</td>
<td>92.42±2.4</td>
</tr>
<tr>
<td>Ratio</td>
<td>0.82±0.01</td>
<td>0.45±0.01</td>
</tr>
</tbody>
</table>

Table 3.1: Summary of the measurements of the light output. The film measurements characterized the response of the detectors in a contact geometry, and the T.V. camera measurements characterized the performance at a distance of 150 cm.
output of our glass scintillator (which can be thought of as many point sources distributed over a volume) since the angular distribution of light exiting a refractive emitter is smallest in the forward direction (see Fig. 3.1).

3.4.3 Spatial resolution

Figure 3.5 shows the MTF's measured for the phosphor detector (dashed) and the glass scintillator for normally (solid) and obliquely (dotted) incident x-rays. The precision of the measurements is approximately ±0.03 for the phosphor detector and ±0.08 for the glass scintillator. The MTF's measured for the phosphor detector with normally and with obliquely incident x rays are identical (and have been represented by one dashed line). The MTF of the glass scintillator is slightly better than that of the phosphor detector for perpendicularly incident x rays. However, the MTF of the glass scintillator is degraded when x rays are incident obliquely. As mentioned in Sec. 2.3.4, this decrease in spatial resolution occurs in the radial direction only. Therefore, in Fig. 3.5, the dotted line

![Graph showing the relative luminous flux versus distance from x-ray detector.](image)
Figure 3.5: MTF's measured with the T.V. camera-based portal imaging system for the copper plate/phosphor screen (dashed) and the glass scintillator (solid and dotted). The MTF's have been measured with normally incident x rays and obliquely incident x rays. The MTF's for the copper plate/phosphor screen obtained with normally and with obliquely incident x rays are identical. These measurements were obtained with the 6 MV beam from a Varian 2100c linear accelerator.

represents the worst-case MTF. At the edge of a large field, the MTF obtained with the glass scintillator in the tangential direction should be similar to that obtained with normally incident x rays.

3.5 Discussion

In this chapter, we examine two x-ray detectors proposed for portal imaging: a thick, high-density glass scintillator and a copper plate/phosphor screen combination. The measurements show that the ratio of the light output from the two detectors is larger in a contact geometry than that obtained 150 cm away from the detectors (see Table 3.1). The luminous flux calculations provide a possible explanation. The computations show that refraction at the glass-air interface causes significant losses in luminous flux. The calculations support the hypothesis that refraction at the glass-air interface changes the angular distribu-
tion of light exiting the glass scintillator compared to the phosphor screen, resulting in the decrease in the light output ratios that have been measured experimentally for these two detectors.

The MTF of the current portal imaging system was compared with the MTF obtained when the glass scintillator is substituted for the phosphor detector. The measured MTF's illustrate one of the drawbacks of using the glass scintillator. While the MTF's measured using x rays with perpendicular incidence are almost identical for both detectors (the glass scintillator is slightly better), the MTF measured using x rays with oblique incidence is much worse for the glass scintillator. When the obliquely incident x rays interact with the (very thick) glass scintillator the apparent size of the optical light source increases in the radial direction (see Figs. 3.2 and 3.5). The visual effect of this radial distortion is difficult to estimate since the MTF in the tangential direction is unaffected. Clearly, however, the radial distortion reduces the spatial resolution of the glass scintillator.

Our measurements and calculations have revealed that the glass scintillator is a poor x-ray detector for portal imaging. The glass scintillator will be useful for portal imaging only if its light output is increased and its thickness decreased. However, the glass scintillator may be useful in other imaging situations. For instance, the use of the glass scintillator would be suitable for situations where there is little x-ray divergence (i.e., small field of view or large distance separating the x-ray source and the glass scintillator). Already, medical imaging systems which use transparent scintillators have been implemented successfully for applications involving small fields of view.36,37 The glass scintillator is also suitable for industrial radiography, where one can compensate for the low light output of the glass by increasing the radiation dose used to form an image.
3.6 Conclusions

Our evaluation of the glass scintillator has shown that the amount of light collected from the glass scintillator is less than that of the copper plate/phosphor screen detector. Furthermore, the spatial resolution obtained with the glass scintillator varies with the angle of incidence the x-ray beam; this variation was not observed with the copper plate/phosphor screen detector. Because of the low light output of the glass scintillator and the degradation of spatial resolution associated with the angle of x-ray incidence, the glass scintillator evaluated in this study is not recommended for portal imaging.
Appendix A: Derivation of the apparent size of the light sensor

Consider the refraction event shown in Fig. 3.1. A refracted light ray is intercepted at a distance $r$ from the central axis on the plane located at a distance $z$ from the glass/air interface. $\Delta_1$ is related to the angle of emission $\theta_1$ and the distance between the source and the glass-air interface $d$ by $\Delta_1 = d \tan \theta_1$. Similarly, $\Delta_2$ is related to the exit angle $\theta_2$ and $z$ by $\Delta_2 = z \tan \theta_2 = z \frac{\sin \theta_2}{\sqrt{1 - \sin^2 \theta_2}}$, and

$r$ is thus given by the sum of $\Delta_1$ and $\Delta_2$, resulting in

$$r = \Delta_1 + \Delta_2 = d \tan \theta_1 + z \frac{\sin \theta_2}{\sqrt{1 - \sin^2 \theta_2}}. \tag{A1}$$

However, $\theta_1$ and $\theta_2$ are related by Snell's law. Assuming an index of refraction of 1 for air, the relationship between $\theta_1$ and $\theta_2$ is $\sin \theta_2 = n \sin \theta_1$. Thus, $r$ can be expressed as a function of $\theta_1$ by substituting the above expression in Eq. (A1). $r$ is therefore given by

$$r = z \frac{n \sin \theta_1}{\sqrt{1 - n^2 \sin^2 \theta_1}} + d \tan \theta_1, \tag{A2}$$

as given by Eq. (2).
References


Chapter 4: A QUANTUM ACCOUNTING AND DETECTIVE QUANTUM EFFICIENCY ANALYSIS FOR VIDEO-BASED PORTAL IMAGING

Abstract

The quality of images generated with radiographic imaging systems can be degraded if an inadequate number of secondary quanta are used at any stage before production of the final image. A theoretical technique known as a "quantum accounting diagram" (QAD) analysis has been developed recently to predict the detective quantum efficiency (DQE) of an imaging system as a function of spatial frequency based on an analysis of the propagation of quanta. It is used to determine the "quantum sink" stage(s) (stages which degrade the DQE of an imaging system due to quantum noise caused by a finite number of quanta), and to suggest design improvements to maximize image quality. We have used this QAD analysis to evaluate a video-based portal imaging system to determine where changes in design will have the most benefit. The system consists of a thick phosphor layer bonded to a 1 mm thick copper plate which is viewed by a T.V. camera. The imaging system has been modeled as ten cascaded stages, including: (i) conversion of x-ray quanta to light quanta; (ii) collection of light by a lens; (iii) detection of light quanta by a T.V. camera; (iv) the various blurring processes involved with each component of the imaging system; and, (v) addition of noise from the T.V. camera. The theoretical DQE obtained with the QAD analysis is in excellent agreement with the experimental DQE determined from previously published data. It is shown that the DQE is

1 This chapter has been accepted for publication in Medical Physics (1996). The title of the article is "A quantum accounting and detective quantum efficiency analysis for video-based portal imaging" by J.-P. Bissonnette, I. A. Cunningham, D. A. Jaffray, A. Fenster, and P. Munro.
degraded at low spatial frequencies (< 0.1 cycles/mm) by quantum sinks both in the number of detected x rays and the number of detected optical quanta. At higher spatial frequencies, the optical quantum sink becomes the limiting factor in image quality. The secondary quantum sinks can be prevented, up to a spatial frequency of 0.5 cycles/mm, by increasing the overall system gain by a factor of 18 or more, or by improving the modulation transfer function (MTF) of components in the optical chain.
4.1 Introduction

Medical x-ray imaging systems are generally designed to produce images having the highest quality possible for a specified dose. One measure of system performance is the detective quantum efficiency (DQE), which is a measure of how efficiently an imaging system transfers the signal-to-noise ratio (SNR) in the incident radiation beam. This SNR describes the ratio of the number of x-ray quanta to the random fluctuation in this number of quanta. The DQE is reduced if the imaging system does not detect all of the incoming x rays, adds noise to the image (i.e., noise due to electronic components or to variations in the intensity of bursts of light produced by the x-ray detector\(^1\)), or suffers from a low gain stage that results in an insufficient number of quanta at the final stage of the image-formation process to maintain the necessary SNR. The analysis described here does not account for “fixed pattern noise” or noise generated by variations in light output caused by localized energy deposition and optical transport (e.g., noise described by Nishikawa et al.\(^1\)).

A theoretical analysis of imaging systems, known as a quantum accounting analysis, has been introduced recently by Cunningham et al.\(^2\) to help design imaging systems. In this theory, which is based on expressions for the propagation of noise and signal derived by Rabbani et al.\(^3-5\) the average quantum fluence, modified by the modulation transfer function (MTF) squared, is determined for each stage of an imaging system and displayed on a “quantum accounting diagram” (QAD),\(^2\) similar to a “nomogram”.\(^6\) Designers can use a QAD to identify the stage with the fewest quanta at any specified frequency (i.e., a spatial-frequency-dependent “quantum sink”) which generally limits the SNR of the imaging system at that frequency. Therefore, designers can use the approximate QAD approach to optimize the performance of an imaging system by
ensuring that the quantum sink occurs only at the x-ray detection stage for any specified spatial frequency. A conventional "zero-frequency" version of this approach has been in use for at least 45 years, but ignoring the frequency dependence has been shown to give misleading results.\textsuperscript{2,8}

One area of medical imaging where the QAD theory can be useful is portal imaging. In a previous study, a zero-frequency analysis of a video-based portal imaging system was performed;\textsuperscript{9} however, this type of simplified analysis cannot account for the effects of improving spatial resolution or reducing additive noise.

In this chapter, we analyze a T.V. camera-based portal imaging system\textsuperscript{9} using QAD formalism to determine the noise-limiting processes of this imaging system and to determine which system parameters could be modified to improve the overall imaging performance. Furthermore, we present a theoretical calculation, derived from this QAD analysis, of the spatial-frequency-dependent DQE of this particular imaging system and compare the theoretical DQE with an experimental DQE determined using previously published data.\textsuperscript{10}

4.2 Background

4.2.1 QAD analysis

Characterization of the portal imaging system using the quantum accounting analysis theory\textsuperscript{2-5} requires representing the system as a serial cascade of multiple stages, where the number of quanta "leaving" each stage constitutes an effective input to the subsequent stage. This type of analysis can be used to describe linear and shift-invariant systems. The average number of quanta per unit area ($\bar{\Phi}$) as well as the noise-power spectrum (NPS) assuming
stationary noise can be determined at each stage of the image-formation process, and the number of quanta at any stage can be related to the average number of incident x rays. In this theory, quanta are propagated through two types of stages: (i) gain processes, where the number of quanta exiting a stage is related to the number of quanta entering by an average gain factor (e.g., amplification, collection efficiency, detection efficiency, and associated gain variance); and, (ii) stochastic spatial spreading processes, where the number of quanta is conserved. Stochastic spatial spreading occurs when quanta are dispersed randomly into a spatial distribution with a probability given by the point-spread function, which is related to the MTF.2,3 Stochastic spreading has also been described mathematically as a "stochastic convolution" process, and results from the statistical properties of the quanta (e.g., light photons) undergoing the spatial spreading.11 The order of each image-forming stage is important, and, in the quantum accounting analysis, each stage can be either a gain or a stochastic spreading process, but not both. Therefore, any physical process which involves both gain and spread must be represented by two separate stages (a gain stage and a spread stage) in the quantum accounting analysis.

4.2.1.1 Gain stages

When the $i^{th}$ stage consists of a gain only, the propagated mean fluence ($\overline{\Phi}_i$) and gain Poisson excess ($\varepsilon_{\phi_i}$) are\(^2\)

$$\overline{\Phi}_i = \overline{\Phi}_{i-1},$$

and

$$\varepsilon_{\phi_i} = \frac{\sigma_{\phi_i}^2}{\overline{\Phi}_{i-1}} - 1,$$

where $\overline{\Phi}_{i-1}$ is the mean number of quanta per unit area incident on stage $i$, $\overline{\Phi}_i$ is the average fluence gain associated with stage $i$, and $\sigma_{\phi_i}^2$ is the variance in this gain. The Poisson excess represents the relative amount by which the gain
variance is in excess of a gain which follows Poisson statistics. The gain may represent an amplification or an interaction probability. The latter is a binary selection process described by a probability \( \overline{g} \), that a given input quantum is propagated to the next stage. As a result, the outcome can be 1 or 0 only. For binary selection processes, the Poisson excess is \( \varepsilon_{\varphi} = -\overline{g} \), as a consequence of the binomial theorem.

4.2.1.2 Stochastic spreading stages

A spreading process is characterized by the MTF of the stage, \( T(\omega) \). When the \( i \)th stage is a spreading process, the propagated mean fluence and gain Poisson excess are\(^2\)

\[
\overline{\Phi}_i = \overline{\Phi}_{i-1}
\]

and \( \varepsilon_{\varphi} = -1 \)

since the spreading process involves no gain. Rather, a spreading process involves a random redistribution of individual quanta according to the point-spread function associated with the spreading process.\(^2,3,11\)

4.2.2 DQE analysis

The DQE of the portal imaging system can be derived, in terms of the signal and noise transfer characteristics of the system, using two approaches. The first approach makes use of measured values for the NPS and the MTF of the imaging system as a whole to calculate the DQE. This approach will be referred to throughout the chapter as the "experimental approach". In the second approach, the DQE is calculated using the QAD analysis, where the mean gains, MTF's, and Poisson excesses from individual stages are propagated.
4.2.2.1 Experimental approach

In terms of the NPS and MTF of an imaging system, the DQE can be written as

\[
DQE(\omega) = \frac{G^2 T_{\text{out}}^2(\omega) \bar{\Phi}_0}{S_{\text{out}}(\omega)},
\]

where \( \bar{\Phi}_0 \) is the mean x-ray fluence of the input radiation beam (in units of 1/mm²), \( S_{\text{out}}(\omega) \) is the NPS of the image (in units of 1/mm²), \( G \) is the average gain of the entire imaging system, and \( T_{\text{out}}(\omega) \) is the MTF of the entire imaging system. Note that Eq. (3) is slightly different from the more usual form, where \( \bar{\Phi}_0 \) is in the denominator. This is because our notation conforms to the notation of Cunningham et al.\(^2\)

4.2.2.2 QAD approach

The spatial-frequency-dependent DQE can also be expressed in terms of the average gain and MTF of individual stages, as well as any additive noise introduced in a stage. From Eqs. (1-2), the expression for the DQE derived by Cunningham et al.\(^2\) is

\[
DQE_{\text{QAD}}(\omega) = \frac{1}{1 + \sum_{i=1}^{M} \left( 1 + \varepsilon_i \frac{[T_i(\omega)]^2 + S_{s_i}(\omega)/\bar{\Phi}_i}{P_i(\omega)} \right)},
\]

where \( S_{s_i}(\omega)/\bar{\Phi}_i \) is the relative additive noise in the \( i \)th stage and \( P_i(\omega) \) is the product of the gains and squared MTF's up to and including the \( i \)th stage, given by

\[
P_i(\omega) = \prod_{j=1}^{i} g_j |T_j(\omega)|^2.
\]

Equation (5) describes the product of the mean gain and squared MTF of every stage up to and including the \( i \)th stage and can be interpreted as an effective number of quanta (x rays, light quanta or photoelectrons) which propagate the
image signal through the stages of the system, at each spatial frequency. A plot of \( P(\omega) \), as a function of stage number \( i \), yields the quantum accounting diagram. The stage with the lowest \( P(\omega) \) value corresponds to the limiting spatial-frequency-dependent quantum sink, and is generally the noise-determining stage. In general, the DQE given by Eq. (4) is degraded at any stage where \( P(\omega) \) is close to or less than unity. Therefore, this approach can be used to determine which stage(s) are the noise-limiting stages, and what changes in system design are required to prevent secondary quantum sinks and thus maximize image quality.

The DQE of an imaging system can therefore be described both by: (i) its overall signal and noise transfer properties [Eq. (3)]; and, (ii) the signal and noise transfer properties of each stage in a linear system model [Eqs. (4) and (5)]. In this article, the experimental DQE obtained using Eq. (3) and previously published data\(^{10}\) is compared with the theoretical DQE obtained using Eq. (4).

4.3 System description

The portal imaging system developed at the London Regional Cancer Centre, shown schematically in Fig. 4.1, consists of a T.V. camera (Video Optics, V1509 B camera) which detects the light produced by the "copper

![Diagram of the imaging system](image)

**Figure 4.1:** Schematic of the T.V. camera-based portal imaging system.
plate/phosphor screen* x-ray detector through a 45° mirror and a large aperture lens (Fujinon, 50 mm, F/0.7) with a demagnification factor of 22.1.9,10,12 The video signal is then digitized by an 8-bit frame-grabber (Infimed, Liverpool, NY) installed on a Sun Sparcstation 2 (Sun Microsystems, Mountain View, CA). The copper plate/phosphor screen detector consists of a 360 mg/cm² thick layer of Gd₂O₂S:Tb phosphor directly bonded to a 1 mm thick copper plate (Eastman Kodak, Rochester, NY). In order to minimize the contribution of the T.V. camera noise to the portal images, the camera is operated in a pulse-progressive mode, where the light emitted from the phosphor screen is accumulated on the lead-oxide target of the T.V. camera (Philips Components, XQ2182 tube) throughout a short irradiation (e.g., from 0.25 to 2.0 s).

The DQE and QAD analyses, as do all NPS-based methods, assume linear, shift-invariant systems with stationary noise processes. The MTF and NPS for the central region (~ 15x15 cm²) of images obtained with the portal imaging system have been measured previously using a 6 MV x-ray beam.13 In this region, the portal imaging system can be assumed to be linear9 and shift-invariant. The latter condition is closely approximated in the central quadrant of images obtained with the portal imaging system, where both lens vignetting and off-axis lens aberrations affect image quality minimally. Also, the random noise processes present in T.V. camera-based systems can be assumed to be stationary and ergodic in T.V.-based radiographic systems.14 Since all the measurements involving the lens and T.V. camera have been performed on the central axis (i.e., MTF) or on the central quadrant of images (i.e., NPS), the portal imaging system can be modeled, using the QAD approach, for a 6 MV beam to enable a comparison of the theoretical and measured DQE.
4.4 QAD analysis

For the purpose of the QAD analysis, the system has been divided into ten stages consisting of gain or spreading processes only, as shown in Fig. 4.2. This analysis requires that the gain, Poisson excess, and MTF of each stage described in this section be determined.

We have performed this analysis in the direction perpendicular to the scan lines of the T.V. camera only (i.e., vertical direction). While a full two-dimensional calculation is more comprehensive, our previous work has shown that the MTF of the portal imaging system is uncertain in the direction parallel to

![Diagram](image)

**Figure 4.2:** Block diagram showing the ten image-forming stages which have been used to analyze the T.V. camera-based system using the QAD theory. The QAD theory requires that the gain, Poisson excess, and MTF of each stage be determined.
the scan lines of the T.V. camera (i.e., horizontal direction) since the design and adjustment of the camera electronics (i.e., bandwidth of the preamplifier) affect the shape of the line-spread function (LSF) in this direction. The MTF measured in the perpendicular direction is generally worse than in the parallel direction, and hence our QAD and DQE analysis represents a “worst-case” estimate.

4.4.1 QAD analysis

4.4.1.1 Stage 1: Detection of primary x-ray quanta, $g_i$

The first stage is the deposition of x-ray energy in the x-ray detector. Only those x-ray photons which interact with the copper plate/phosphor screen detector and deposit energy in the phosphor screen produce optical quanta which contribute to the portal image. The probability ($g_i$) that an incident x ray interacts in this way has been calculated using a computer program based on the EGS4 Monte Carlo system. This program, which has been described by Jaffray et al., is used to generate an absorbed energy distribution (AED) in a specified material. The AED describes the number of incident x rays (photon histories) which deposit an energy between $E$ and $(E+\Delta E)$ in the phosphor layer of the x-ray detector as a function of energy $E$. One such AED is shown in Fig. 4.3. As far as our analysis goes, the only x-ray photons which are considered in the AED are those which actually deposit energy (i.e., $E>0$) in the phosphor layer. This is because x rays which interact with the detector but do not deposit energy in the phosphor layer do not generate the optical quanta which are necessary to form an image. The value of $g_i$ is determined as the area of the AED (i.e., the total number of energy deposition events) divided by the total number of x-ray photons incident on the detector. The AED was calculated ten times, using in-
Figure 4.3: Absorbed energy distribution for the copper plate/phosphor screen detector, when irradiated by a 6 MV x-ray beam. This distribution is used to establish $g_1$, $g_2$, and their respective Poisson excesses. The average energy deposited in the phosphor layer per x-ray which actually deposit energy in the phosphor layer (i.e., 0.44 MeV) results, on average, in 28,755 optical quanta being produced. The abrupt change in the AED at 0.25 MeV is caused by the poor resolution of the 6 MV x-ray spectrum used to calculate the AED.

dependent sequences of random numbers, and the values of $g_i$ obtained from each of the ten AED's were averaged to reduce the uncertainty in the value of $g_1$.

Each AED was calculated using $10^6$ photon histories and the 6 MV x-ray spectrum of Kubsad et al. The user-adjustable parameters which affect the transport of high-energy particles in the EGS4 program (e.g., the corrected parameter reduced electron step transport algorithm, the minimum total energy of photons and electrons which are transported (ECUT, PCUT), and the energy thresholds for the creation of secondary photons and electrons (AE, AP)] were set as described in Ref. 13.

For the 6 MV spectrum impinging on our x-ray detector (a 1 mm copper plate and a 360 mg/cm² gadolinium oxysulfide phosphor screen), we estimate
that approximately 3.5% of the incident x-ray photons deposit energy in the phosphor screen (i.e., \( g_1 = 0.0357 \pm 0.0001 \)).

4.4.1.2 Stage 2: Spread of high energy particles in the phosphor screen, \( T_2(\omega) \)

Spread of signal in the copper plate/phosphor screen detector occurs because of migration of high energy particles (i.e., scattered x rays, electrons set in motion by x-ray interactions in the x-ray detector, and bremsstrahlung) in both the metal plate and the phosphor screen as well as because of light spread in the phosphor screen.\(^{18}\) To separate these two effects and determine the spread due to high energy particles alone, the following technique was used. Two steel blocks (60 cm thick with a 72 mm groove cut along the length of one block) were clamped together to form a narrow slit. This slit collimator was carefully aligned with the x-ray source to produce a very narrow beam. The beam was used to irradiate the copper plate/phosphor screen, which was covered with a thin (~ 60 \( \mu \)m) sheet of opaque, black plastic to prevent optical photons from reaching Kodak EM-1, single-emulsion film which was placed in close contact with the plastic film. Thus, the LSF recorded by these films was due to the interaction of high energy particles with the film emulsion. These films were processed in a Kodak M6B X-Omat film processor. The LSFs recorded on two series of films were digitized twice with a Perkin-Elmer PDS scanning microdensitometer, first using an aperture which was 20 \( \mu \)m and then with an aperture which was 10 \( \mu \)m. A small sampling increment (20 \( \mu \)m first, 5 \( \mu \)m second) was used to minimize the effect of aliasing in the measurement. The measurements yielded similar MTF curves, suggesting little aliasing was present.

To determine the tails of the LSF accurately, 5 films were obtained, each exposed to a different x-ray dose (ranging between 250 and 32 000 monitor
Figure 4.4: The modulation transfer function [MTF] of: (i) the copper plate/phosphor screen detector; (ii) the copper plate/phosphor screen due to the spread of high energy particles only; (iii) the copper plate/phosphor screen due to diffusion of optical quanta only; (iv) the lens; and, (v) the T.V. camera and lens assembly. The MTFs associated with the copper plate/phosphor screen detector were measured using 6 MV x-rays. The uncertainty in the MTFs ranged between ±0.04 and ±0.06.

units, where 1 monitor unit = 1 cGy dose to the isocentre for a 10×10 cm² field at a depth of 5 cm in water). The film non-linearities were corrected by using the characteristic curve measured with the scanning densitometer for EM-1 film exposed to high-energy particles. The resulting LSF was processed using standard techniques used for screen/film combinations. The measured MTF of the copper plate/phosphor screen due to the transport of high energy particles, \( T_2(\omega) \), is shown in Fig. 4.4.

4.4.1.3 Stage 3: Generation of optical quanta in phosphor, \( g_3 \)

The AED's described in Sec. 4.4.1.1 were also used to determine the gain associated with the third stage \( (g_3) \), which is the conversion of x-ray quanta to optical quanta. The average number of optical quanta generated per x-ray which actually deposit energy in the phosphor layer is^{21}
\[ g_3 = \frac{\eta E_{ab}}{E_{opt}}, \]  

(6)

where \( E_{ab} \) is the mean energy absorbed in the phosphor layer per photon history, \( \eta \) is the energy conversion efficiency (i.e., the fraction of deposited energy converted to optical energy), and \( E_{opt} \) is the mean energy of the optical quanta produced. For \( \text{Gd}_2\text{O}_2\text{S}:\text{Tb} \), \( \eta \) is 0.15 and \( E_{opt} \) is 2.3 eV.\(^\text{22} \) The light emitted from \( \text{Gd}_2\text{O}_2\text{S}:\text{Tb} \) is approximately monochromatic, and hence the variance in \( E_{opt} \) can be ignored. A value of \( E_{ab} \) and its associated variance were determined from each AED in the following manner. Each AED was converted into probability density functions \( [f_{\epsilon}(E)] \) by normalizing the area of each to 1. The value of \( E_{ab} \) and its variance are therefore given by the first moment about the origin \( [\bar{E}_{ab} = \int_{0}^{\infty} E f_{\epsilon}(E)dE] \) and the second moment about the mean [i.e., \( \sigma^2_{E_{ab}} = \int_{-\infty}^{\infty} (E - \bar{E}_{ab})^2 f_{\epsilon}(E)dE \)], respectively. The ten values of \( E_{ab} \) and \( \sigma^2_{E_{ab}} \) were subsequently averaged. The average value of \( E_{ab} \) (0.44 MeV) and its variance (0.207 MeV\(^2 \)) were used in Eq. (6), resulting in \( g_3 = 2.88 \pm 0.01 \times 10^4 \) and \( \sigma^2_{g_3} = 8.8 \pm 0.1 \times 10^8 \). This surprisingly large value of \( \sigma^2_{g_3} \) is due to the very large variance in \( E_{ab} \). The corresponding Poisson excess is \( \varepsilon_{g_3} = (\sigma^2_{g_3}/g_3)^{-1} = 3.07 \pm 0.03 \times 10^4 \). The high value of \( \varepsilon_{g_3} \) suggests that, when high energy x rays are used, the energy absorption noise, which is related to the noise associated with the generation of optical quanta, is not Poisson noise. However, it is shown in the results section that this large excess value does not degrade the overall DQE of the T.V. camera-based system significantly.

Note that Eq. (6) ignores variations in light output due to spatial variations in energy deposition and light transport in phosphor screens of finite thickness.\(^1,13,21 \) Because of the high energy of the incident x rays, x-ray interactions in the x-ray detector do not deposit all of their energy locally. Thus, the situation
at megavoltage energies is much different than that encountered in diagnostic radiology.\textsuperscript{1}

4.4.1.4 Stage 4: Spread of optical quanta in the phosphor screen, $T_4(\omega)$

The fourth stage is characterized by the spread of optical quanta in the phosphor screen [$T_4(\omega)$]. It is not possible to measure $T_4(\omega)$ directly. Rather, only the MTF of the copper plate/phosphor screen combination, $T_2(\omega) \times T_4(\omega)$ (including spread due to radiation and optical processes), and $T_2(\omega)$ (the MTF due to radiation processes) can be measured directly. The product $T_2(\omega) \times T_4(\omega)$ has been measured previously,\textsuperscript{10} using a method similar to that described in Sec. 4.4.1.2. The value of $T_4(\omega)$ is therefore obtained by dividing the experimentally measured MTF of the copper plate/phosphor screen for the 6 MV x-ray beam from a linear accelerator (Varian 2100c) by $T_2(\omega)$.

4.4.1.5 Stage 5: Escape of optical quanta from the phosphor screen, $g_5$

The fifth stage involves the probability ($g_5$) that an optical quanta generated inside the phosphor escapes through the exit surface of the phosphor. For our 360 mg/cm\textsuperscript{2} phosphor, the value of $g_5$ has been calculated using the Kubelka-Munk theory\textsuperscript{23} to be 0.4 (David Trauernicht, Eastman Kodak, private communication). Because of the difference between the value of $g_5$ given by the screen manufacturer (i.e., $g_5 = 0.4$) and that reported from Monte Carlo simulations of optical light transport inside a phosphor screen of thickness similar to ours (i.e., $g_5 = 0.2$),\textsuperscript{21} the uncertainty in $g_5$ has been estimated to be ±0.2.
4.4.1.6 Stage 6: Collection of light quanta by the lens, $g_e$

The sixth stage involves the lens collection efficiency ($g_e$) of the imaging system, which is the fraction of photons exiting the phosphor that reach the target of the T.V. camera. The lens collection efficiency, estimated by taking the ratio of the area of the lens of the T.V. camera to half surface area of the sphere where light exiting the phosphor is emitted, can be expressed as:

$$g_e = \frac{\tau}{8f^2(1+m)^2},$$

(7)

where $f$ is the F-number of the lens, $m$ is the image demagnification factor, and $\tau$ is the transmittance of the lens. Equation (7) differs from the expression derived by Boyer by a factor of 2 since the escape probability of Sec. 4.4.1.4 already discriminates against light which is not emitted through the exit surface of the phosphor screen. The lens of our T.V. camera (Fujinon, model CF50L, NJ) has a F-number of 0.7, and we have assumed a transmittance of 0.9. For our geometry, the value of $m$ is 22.1. The resulting value of $g_e$ is $4.8 \times 10^{-4}$.

4.4.1.7 Stage 7: Spread of light quanta in the lens, $T_s(\omega)$

The optical spreading process in the seventh stage is described by the MTF of the lens [$T_s(\omega)$]. We can measure the combined MTF of the lens and T.V. camera assembly [i.e., $T_s(\omega) \times T_0(\omega)$]. However, it is essential to separate this measured MTF into the MTF's of the two individual components in order to calculate the quantum accounting diagram. In addition, this separation gives the opportunity to see how the QAD and the DQE change when we consider different combinations of lenses and T.V. cameras. Therefore, the MTF of the lens was obtained by dividing the measured $T_s(\omega) \times T_0(\omega)$ by the MTF of the T.V. camera only [$T_0(\omega)$]. $T_0(\omega)$ was obtained from the data sheet of the XQ2182 tube of the T.V. camera (Philips Components, Slatersville, RI). [Note that the tube MTF
reported by the manufacturer is, in fact, the product of the MTF of the tube and the MTF of the lens used in their measurements. However, these measurements are performed using a small lens aperture (i.e., F-number of 5.6), which reduces the loss in spatial resolution due to the lens. Therefore, we have assumed that the reported camera MTF is entirely due to the camera itself.]

The determination of the MTF of the lens and T.V. camera assembly \([T_f(\omega) \times T_0(\omega)]\) involved the measurement of the LSF of the lens and T.V. camera assembly,\(^{25,26}\) using standard image acquisition techniques.\(^{10,12,20}\) Two slabs of black plastic were glued together to form a narrow (85 \(\mu\text{m}\) wide, 5 cm long) light beam. These slabs were placed at the same location normally occupied by the x-ray detector, at the centre of the field of view of the T.V. camera. A total of 60 images of this narrow light beam were acquired, (slit images) and another 60 images were acquired with no light reaching the T.V. camera (background images). Each set of images was summed to reduce noise. The difference between the summed slit images and the summed background images resulted in the LSF image, which was used to generate the LSF and which was Fourier-transformed to yield \(T_f(\omega) \times T_0(\omega)\). The two sets of images were acquired by accumulating signals on the target of the T.V. camera for 1 s, which is the exposure time required to acquire clinical images, in order to ensure that time-dependent phenomena had similar effects on both the MTF measurements and the clinical images.\(^{10}\) These measurements were performed with a \(\Phi\) \(\phi\) narrow light beam placed at a slight angle from the direction parallel to the T.V. camera scan lines in order to improve the sampling of the LSF\(^{27}\) and to prevent asymmetries in the measured LSF.\(^{10}\) The measurement of \(T_f(\omega) \times T_0(\omega)\) was confirmed by measuring the square-wave response of the lens and T.V. camera assembly, which was converted to \(T_f(\omega) \times T_0(\omega)\) using the theoretical relationship derived by Coltman.\(^{28}\) In order to mimic the light emitted by the copper plate
phosphor screen detector, all of these measurements were performed with ambient light filtered by a green filter (Eastman Kodak, Wratten filter #74, Rochester, NY) mounted on the lens of the T.V. camera.

4.4.1.8 Stage 8: Detection of optical quanta by the T.V. camera, $g_b$

The eighth stage is described by the detection efficiency ($g_b$) of the T.V. camera, which is the probability that an incident light quantum releases a photoelectron in the target of the T.V. camera. The value of $g_b$, obtained by weighting the spectral response of the XQ2182 tube of the T.V. camera by the spectrum of light emitted by Gd$_2$O$_2$S:Tb, is 0.37. The spectral response of the XQ2182 tube was obtained from the tube's data sheet.

4.4.1.9 Stage 9: Spread in the T.V. camera, $T_o(\omega)$

The ninth stage involves the spreading process associated with the T.V. camera only [$T_o(\omega)$]. As mentioned in Sec. 4.4.1.7, $T_o(\omega)$ was obtained from the data sheet of the XQ2182 tube of the T.V. camera.

4.4.1.10. Stage 10: Additive noise, $S_o(\omega)$

The last stage involved the addition of noise by the electronics of the T.V. camera, described by the term $S_o(\omega) \sqrt{\Phi}$ in Eq. (4). The calculation of this additive term was performed using the NPS measured while no light was allowed to reach the lens of the T.V. camera. This NPS was measured using the two-dimensional technique described previously. Briefly, a large number of independent images were acquired and subtracted in pairs to remove any structural variations in the images. The two-dimensional NPS was calculated by averaging the modulus of the Fourier transform of the central portion (256x256 pixels)
of these subtracted images. Since the subtraction of statistically independent images doubles the variance, the NPS was divided by $2^{10.14.30}$ The slice through the two-dimensional NPS corresponding to the direction perpendicular to the camera scan lines was selected. The unitless term $S_a(\omega) / \Phi$, needed for the QAD analysis was obtained by dividing the NPS by the average pixel value of the images obtained with a 1 monitor unit irradiation, the number of ADC units per electron generated in the camera, and the pixel area scaled to the plane of the x-ray detector. The value of $S_a(\omega) / \Phi$ ranged between 1.1 and 6.8, depending on spatial frequency. The fluence-to-dose conversion factor, obtained by weighting fluence-to-dose conversion factors$^{31}$ with the 5 MV x-ray spectrum of Kubsad et al.$^{15}$, was $7.4 \times 10^{-8}$ cGy mm$^2$/x-ray at the isocentre of the linear accelerator, at a depth of 5 cm in water. Stage 10 describes the addition of noise with no quantum gain nor stochastic blur. This corresponds to $g_{10}=1$, $c_{10}=-1$, and $T_{10}(\omega)=1$.

4.4.2 Calculation of the DQE

4.4.2.1 Experimental calculation

The experimental DQE of the portal imaging system was first determined using Eq. (3) and previously measured MTF and NPS.$^{10}$ The DQE was calculated with the slices through the two-dimensional MTF and NPS which correspond to the direction perpendicular to the camera scan lines.

4.4.2.2 Theoretical calculation

The spatial-frequency dependent DQE of the T.V. camera-based portal imaging system is obtained using Eq. (4) and the values of the gains, Poisson
<table>
<thead>
<tr>
<th>Stage</th>
<th>Description</th>
<th>Type of process</th>
<th>Symbol</th>
<th>Gain ($g$)</th>
<th>Poisson excess $e_g$</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Absorption of x-ray energy in phosphor</td>
<td>Binomial</td>
<td>$g_1$</td>
<td>0.0357</td>
<td>-0.0357</td>
</tr>
<tr>
<td>2</td>
<td>Radiative spread in phosphor</td>
<td>Spread</td>
<td>$T_2(\omega)$</td>
<td>1</td>
<td>-1</td>
</tr>
<tr>
<td>3</td>
<td>Generation of optical quanta in phosphor</td>
<td>Amplification</td>
<td>$g_3$</td>
<td>$2.88 \times 10^5$</td>
<td>$3.07 \times 10^5$</td>
</tr>
<tr>
<td>4</td>
<td>Optical spread in phosphor</td>
<td>Spread</td>
<td>$T_4(\omega)$</td>
<td>1</td>
<td>-1</td>
</tr>
<tr>
<td>5</td>
<td>Escape probability of optical quanta</td>
<td>Binomial</td>
<td>$g_3$</td>
<td>0.40</td>
<td>-0.40</td>
</tr>
<tr>
<td>6</td>
<td>Collection by lens</td>
<td>Binomial</td>
<td>$g_6$</td>
<td>$4.8 \times 10^{-4}$</td>
<td>$-4.8 \times 10^{-4}$</td>
</tr>
<tr>
<td>7</td>
<td>MTF of lens</td>
<td>Spread</td>
<td>$T_6(\omega)$</td>
<td>1</td>
<td>-1</td>
</tr>
<tr>
<td>8</td>
<td>Detection in the T.V. camera</td>
<td>Binomial</td>
<td>$g_8$</td>
<td>0.372</td>
<td>-0.372</td>
</tr>
<tr>
<td>9</td>
<td>MTF of T.V. camera</td>
<td>Spread</td>
<td>$T_8(\omega)$</td>
<td>1</td>
<td>-1</td>
</tr>
<tr>
<td>10</td>
<td>Noise added by T.V. camera electronics</td>
<td></td>
<td>$S_\Phi(\omega)/\Phi$</td>
<td>1</td>
<td>-1</td>
</tr>
</tbody>
</table>

Table 4.1: Summary of the gains and Poisson excesses for the eight stages of the T.V. camera-based portal imaging system. For gain stages, $T_\Phi(\omega) = 1.0$.

<table>
<thead>
<tr>
<th>Spatial frequency</th>
<th>0</th>
<th>0.20</th>
<th>0.35</th>
<th>0.50</th>
<th>0.70</th>
</tr>
</thead>
<tbody>
<tr>
<td>$S_\Phi(\omega)/\Phi$</td>
<td>2.16</td>
<td>1.32</td>
<td>1.46</td>
<td>1.41</td>
<td>1.35</td>
</tr>
</tbody>
</table>

Table 4.2: Values of the relative additive NPS, $S_\Phi(\omega)/\Phi$, at five different spatial frequencies.

excesses, and relative additive NPS summarized in Tables 4.1 and 4.2. Many terms in Eq. (4) cancel, and the resulting DQE can be simplified to yield

$$DQE(\omega) = \frac{g_1 T_2^2(\omega)}{1 + \frac{e_g}{g_2} + \frac{1 + S_\Phi(\omega)/\Phi}{g_3 g_5 g_6 g_8 T_4^2(\omega) T_7^2(\omega) T_8^2(\omega)}}.$$  \hspace{1cm} (8)

The product $T_\Phi(\omega) \times T_\delta(\omega)$ corresponds to the MTF measured for the lens and T.V. camera assembly. Thus, it is not actually necessary to determine the MTF of each separately for the DQE, as only the MTF of the lens and T.V. camera combination (which was measured directly) is required in Eq. (8). It is necessary, however, to determine $T_\Phi(\omega)$ and $T_\delta(\omega)$ separately to obtain the QAD of the system, which may reveal secondary quantum sinks in these stages.

\[\text{The derivation of Eq. (8) is shown in Appendix A.}\]
4.5 Results

4.5.1 Absorbed energy distribution

The AED for the x-ray detector (i.e., 1 mm copper plate with a 360 mg/cm² thick gadolinium oxysulfide layer) and the 6 MV spectrum of Kubsad et al. is shown in Fig. 4.3. From this AED, the probability that incident x-rays deposit energy in the phosphor layer of the x-ray detector, \( g_n \), the gain associated from the conversion of x-ray quanta to optical quanta, \( g_3 \), and the Poisson excess associated with the latter gain, \( \varepsilon_{g_3} \), were determined as described in sections 4.4.1.1 and 4.4.1.3.

The AED shown in Fig. 4.3 shows discontinuities near 0.08 and 0.25 MeV. The discontinuity at 0.08 MeV is due to three sources: (i) the x-ray spectrum of Kubsad, which suggests that x-ray photons of energies between 0 and 0.25 MeV are equally probable; (ii) the filtering of the incident beam by the copper plate (there is an abrupt increase in the absorption of x rays by the 1 mm copper plate as the energy of the incident x rays decreases from 0.1 to 0.05 MeV); and, (iii) the K-edge of gadolinium. Thus, the incident spectrum contains photons ranging in energy between 0-0.25 MeV, the lower energy photons of this spectrum are preferentially absorbed in the copper plate, and these photons reaching the screen undergo primarily photoelectric interactions in the screen, resulting in the K-edge of gadolinium influencing the absorption process. The discontinuity at 0.25 MeV is caused by the low energy resolution (\( \Delta E = 0.25 \) MeV) of the x-ray spectrum used in our Monte Carlo simulations, which, in turn, causes an over-representation of x-ray energies just above 0.25 MeV and an under-representation of energies just below 0.25 MeV. Since most incident x-rays with energy below ~0.35 MeV deposit energy in the phosphor layer of the x-ray detector through photoelectric interactions, disconti-
nuities in the incident x-ray spectrum result in the underestimation of the height of photopeaks below 0.25 MeV and an overestimation of the height of photopeaks above 0.25 MeV. This causes a spurious peak in the AED at 0.25 MeV, caused by the input spectrum. Note that the discontinuity at 0.25 MeV does not dominate the shape of the AED shown in Fig. 4.3, and therefore has a negligible effect on the QAD parameters extracted from the AED.

4.5.2 Measurements

Figure 4.4 shows the MTF's used for the QAD analysis of the video-based portal imaging system. The error bars represent one standard deviation from the associated average MTF. Figure 4.4 shows that the loss in spatial resolution due to the transport of high energy particles [$T_2(\omega)$] is less than that caused by optical diffusion in the phosphor screen [$T_4(\omega)$]. Note that the $T_2(\omega)$ shown in Fig. 4.4 is probably an underestimate of the MTF due to the spread of high energy radiation since $T_2(\omega)$ can only be measured at the exit surface of the screen, where the lateral spread of high energy particles is maximal, whereas $T_2(\omega)$ should reflect the spread of high energy particles at the point where light is generated in the phosphor screen. Also, the phosphor screen and the EM-1 single-emulsion film used to record the signals were separated by a thin, low-density, opaque plastic film. The high energy particles can spread slightly while passing through the opaque plastic film. Figure 4.4 also shows that the MTF of the lens [$T_7(\omega)$] is much worse than the MTF of the camera tube [$T_9(\omega)$]. [Note that the value of $T_7(\omega)$, which was determined using the method described in Sec. 4.4.1.7, agrees well with the limited MTF data supplied to us by the lens manufacturer.] Since the MTF of the T.V. camera tube supplied by the tube manufacturer is much higher than the lens MTF, Fig. 4.4 suggests that there is
more potential for spatial resolution improvements with the lens than with the T.V. camera tube.

The frequency-dependent QAD of the portal imaging system is shown in Fig. 4.5, and the values associated with the QAD lines are shown in Table 4.3. In Fig. 4.5, the solid line represents the conventional zero-frequency QAD while the broken lines represent spatial-frequency-dependent values. It can be seen from Fig. 4.5 that, for many non-zero spatial frequencies, the number of optical quanta contributing to the image (i.e., at the last stage) is lower than the number of x-rays which deposit energy in the phosphor layer of the copper plate/phosphor screen detector. Therefore, the portal imaging system is not x-ray quantum limited over many frequencies of interest. In addition, in the final image-forming stage, the zero-frequency QAD is only slightly greater than the number of quanta at the x-ray detection stage, while the QAD at higher spatial frequencies is significantly less. The divergence between the QAD lines due to spreading processes, especially by $T_s(\omega)$ and $T_r(\omega)$, is shown clearly. Also, Fig. 4.5 shows that the major cause of signal loss is the poor lens collection effi-

<table>
<thead>
<tr>
<th>Stage #</th>
<th>Symbol</th>
<th>$\omega = 0$</th>
<th>$\omega = 0.25$</th>
<th>$\omega = 0.5$</th>
<th>$\omega = 0.74$</th>
<th>$\omega = 1$</th>
</tr>
</thead>
<tbody>
<tr>
<td>0</td>
<td>$\Phi_0$</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
<td>1</td>
</tr>
<tr>
<td>1</td>
<td>$g_1$</td>
<td>0.0357</td>
<td>0.0357</td>
<td>0.0357</td>
<td>0.0357</td>
<td>0.0357</td>
</tr>
<tr>
<td>2</td>
<td>$T_2(\omega)$</td>
<td>0.0357</td>
<td>0.0261</td>
<td>0.0169</td>
<td>0.0110</td>
<td>0.0073</td>
</tr>
<tr>
<td>3</td>
<td>$g_3$</td>
<td>1026</td>
<td>749</td>
<td>486</td>
<td>318</td>
<td>209</td>
</tr>
<tr>
<td>4</td>
<td>$T_4(\omega)$</td>
<td>1026</td>
<td>448</td>
<td>180</td>
<td>75</td>
<td>31</td>
</tr>
<tr>
<td>5</td>
<td>$g_5$</td>
<td>411</td>
<td>179</td>
<td>72</td>
<td>30</td>
<td>12</td>
</tr>
<tr>
<td>6</td>
<td>$g_6$</td>
<td>0.197</td>
<td>0.087</td>
<td>0.035</td>
<td>0.014</td>
<td>0.006</td>
</tr>
<tr>
<td>7</td>
<td>$T_7(\omega)$</td>
<td>0.197</td>
<td>0.067</td>
<td>$8.6 \times 10^3$</td>
<td>$5.2 \times 10^4$</td>
<td>$6.6 \times 10^5$</td>
</tr>
<tr>
<td>8</td>
<td>$g_8$</td>
<td>0.073</td>
<td>0.025</td>
<td>$3.2 \times 10^3$</td>
<td>$1.9 \times 10^4$</td>
<td>$2.4 \times 10^5$</td>
</tr>
<tr>
<td>9</td>
<td>$T_9(\omega)$</td>
<td>0.073</td>
<td>0.021</td>
<td>$2.1 \times 10^3$</td>
<td>$9.1 \times 10^5$</td>
<td>$7.3 \times 10^5$</td>
</tr>
</tbody>
</table>

**Table 4.3:** Values of the spatial-frequency dependent quantum accounting diagram [QAD] of the T.V. camera-based portal imaging system shown in Fig. 4.5.
Figure 4.5: Spatial-frequency dependent quantum accounting diagram [QAD] of the T.V. camera-based portal imaging system. The shaded area spans the stages involved with the x-ray detector of the imaging system.

ciency of the imaging system, and that the MTF of the lens significantly degrades image quality at high spatial frequencies. Note that the QAD is approximate since it does not account for noise added by electronic components of the imaging system, nor does it account for noise due to the Poisson excess in the gains.

Figure 4.6 shows the DQE obtained with the QAD approach using Eq. (8) (dotted line) as well as the DQE calculated using Eq. (3) and previously published data (solid line). In contrast to the QAD values, this result does account for the effect of additive and gain noise. Good agreement, over a range of 2.5 orders of magnitude, exists between the calculated and the measured DQEs. The error bars in the experimental DQE's were obtained by propagating the un-
Figure 4.6: The detective quantum efficiency [DQE] of the portal imaging system. The dotted line was obtained from the QAD analysis of the portal imaging system, the dashed line was obtained from the QAD analysis after the additive noise term was set to 0, and the solid line was obtained from previously measured data.\textsuperscript{10} The error bars represent the standard deviation in the measured DQE.

Uncertainties in the MTF and NPS of Ref. 10 using the method described by Bellington.\textsuperscript{32} Similarly, the error bars in the theoretical DQE were obtained by propagating the uncertainties in the gains and MTFs reported in this chapter. The dashed line in Fig. 4.6 shows the theoretical DQE obtained if the additive noise term of Eq. (8), $S_a(\omega)/\Phi_i$, is set to 0. Comparison of the dashed line in Fig. 4.6 with the solid line shows that a significant improvement in the DQE could be obtained if the additive noise could be eliminated.

4.5.3 Parameter sensitivity analysis

The expression derived for the DQE of the T.V. camera-based portal imaging system [Eq. (8)] depends on ten parameters, some of which depend on assumptions. In this section, we identify which parameters affect the DQE significantly, and what uncertainties may be acceptable in these parameters without affecting the conclusions of the analysis.
Changing the value of the gain factors listed in Table 4.1 results in a
greater improvement of the DQE obtained from Eq. (8) at high spatial frequen-
cies than at low spatial frequencies. For example, doubling the absorption of x-
ray energy in the phosphor screen \( (g_1) \) would increase the DQE by a factor of
2.5 at low spatial frequencies and 3.4 at high spatial frequencies. Doubling \( g_2 \)
would increase the DQE by a factor ranging between 1.9 at low spatial frequen-
cies and 3.4 at high spatial frequencies, and doubling any of the other gains
would increase the DQE by a factor ranging between 1.5 at low spatial frequen-
cies and 3.4 at high spatial frequencies. The improvement in the DQE can be
higher than the increase in the gain since the actual number of quanta in the
last image-forming stage increases and, furthermore, the relative additive NPS
[i.e., \( S_a(\omega)/\Phi_r \)] is reduced. If the relative additive NPS is negligible, then the
maximal increase in DQE would not exceed the increase in the gain factor.
Note that, as shown in Fig. 4.6, reducing \( S_a(\omega)/\Phi_r \) to zero while maintaining all
gain factors as listed in Table 4.1 would increase the DQE by a factor ranging
between 1.4 at low spatial frequencies and 7.7 at high spatial frequencies.
Also, note that the increase in \( g_1, g_2, g_3, \) and \( g_4 \) is limited since these parameters
cannot be greater than 1.

The other source of non-Poisson noise is at the stage where optical quanta are generated, and is accounted for by the Poisson excess \( \varepsilon_{s_{p}} \). Sur-
prisingly, if this noise was Poisson noise (i.e., \( \sigma^2_{s_p} = g_2 \), and \( \varepsilon_{s_{p}} = 0 \)), the DQE
would increase only by a factor of 1.35 at 0 cycle/mm, and would not change
considerably (i.e., less than 5%) for spatial frequencies higher than
0.5 cycle/mm. The modest influence of \( \varepsilon_{s_{p}} \) on the system DQE is due to the
secondary quantum sinks, which dominate at high spatial frequencies (see Fig.
4.6). Consequently, the value of the right-hand-side term in the denominator of
<table>
<thead>
<tr>
<th></th>
<th>0 cycles/mm</th>
<th>0.25 cycles/mm</th>
<th>0.5 cycles/mm</th>
<th>0.74 cycles/mm</th>
</tr>
</thead>
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<tr>
<td>$T_2(\omega)$</td>
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<td>1.04</td>
<td>1.09</td>
<td>1.12</td>
</tr>
<tr>
<td>$T_4(\omega)$</td>
<td>1.00</td>
<td>1.02</td>
<td>1.08</td>
<td>1.14</td>
</tr>
<tr>
<td>$T_7(\omega) \times T_9(\omega)$</td>
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<td>1.06</td>
<td>1.35</td>
<td>1.63</td>
</tr>
<tr>
<td>$T_7(\omega)$</td>
<td>1.00</td>
<td>1.06</td>
<td>1.32</td>
<td>1.47</td>
</tr>
<tr>
<td>$T_9(\omega)$</td>
<td>1.00</td>
<td>1.02</td>
<td>1.05</td>
<td>1.11</td>
</tr>
</tbody>
</table>

**Table 4.4:** Relative improvement of the DQE at four different spatial frequencies, resulting from an increase by 10% in the spatial frequency at which individual MTFs are 0.5. The greatest improvement is obtained by increasing the MTF of the lens and T.V. camera combination, $T_7(\omega) \times T_9(\omega)$.

Eq. (8) is much larger than the other terms in the denominator, thereby reducing the relative importance of $\varepsilon_g$ on the system DQE.

The MTF's associated with spreading stages also affect the DQE. Table 4.4 shows the improvements to the DQE if the frequency at which each MTF is 0.5, $f_{0.5}$, is increased by 10%. Clearly, such a modest improvement in the MTF of the lens [$T_7(\omega)$] or the MTF of the lens and camera combination [$T_7(\omega) \times T_9(\omega)$] is sufficient to improve the DQE at high spatial frequencies. In contrast, similar changes in the other MTF's only have a moderate influence on the DQE at high spatial frequencies. These results show that the MTF of the lens is a major determinant of the spatial resolution of the T.V. camera-based imaging system.

To summarize, the parameter sensitivity analysis shows that, if the individual gains are increased by identical factors, the DQE is most affected by changes in the x-ray detection probability, $g_1$. The DQE is less sensitive to changes in other gain stages at low spatial frequencies, and modestly sensitive to changes in $\varepsilon_g$. As far as increments of the MTF's are concerned, the DQE at high spatial frequencies is considerably affected by changes in the lens MTF [$T_7(\omega)$], and only moderately affected by changes in the other MTF's [$T_2(\omega)$, $T_4(\omega)$].
and \( T_\delta(\omega) \). Also, the DQE can be increased considerably if additive noise can be eliminated.

4.6 Discussion

This chapter presents the first experimental demonstration of the QAD theory. Only Monte Carlo simulations have been used previously.\(^2,^8,^{11}\) The DQE calculated using the QAD appears to be slightly greater than the measured DQE; however, they agree within statistical uncertainty, giving confidence that the QAD analysis shown in this chapter can be used as a tool to analyze the performance of T.V. camera-based portal imaging systems and to suggest what modifications should improve image quality. Note that our analysis applies only to the central region of the video-based portal imaging system since the system is not shift-invariant over the entire field of view. Fortunately, linearity and shift-invariance can be reasonably assumed for the central quadrant of images obtained with our system.

The expression derived for the DQE presented in this chapter, Eq. (8), complements a previous zero-spatial frequency DQE derived in Ref. 9 for video-based portal imaging systems. In Ref. 9, the zero-spatial frequency DQE was derived assuming that all processes are either Poisson-distributed gains or binary selections. In Eq. (1) of Ref. 9, \( P_1 \) corresponds to our \( g_1 \); \( P_2 \) to the product of our \( g_3 \) and \( g_5 \); \( P_3 \) to our \( g_7 \); and, \( P_4 \) to our \( g_8 \). Equation (8) is consistent with Ref. 9, but represents a more comprehensive derivation which accounts for non-Poisson noise and evaluates the DQE for all spatial frequencies. Equation (8) demonstrates that maximizing spatial resolution (i.e., the MTF's) and minimizing non-Poisson noise [i.e., \( \varepsilon_g \), and \( S_\delta(\omega)/\Phi \)] is necessary to maximize the DQE. However, the implications from the zero-spatial frequency DQE derived in Ref. 9
still hold: to maximize the DQE, the product of the gains and squared MTF's \( P(\omega) \) should be much greater than unity.

The main limitation to the QAD analysis is the uncertainty in some of the gain factors involved. The largest uncertainty is in the value of the probability that an optical photon generated inside the phosphor screen exits through the exit surface of the screen \( (g_s) \): the value of \( g_s \) used in this analysis is twice as large as that determined by Radcliffe et al.\(^{21}\) for a screen thickness similar to that of our x-ray detector. Using the value of \( g_s \) suggested in Ref. 21 would result in a theoretical DQE which is almost identical to the experimental DQE.

The QAD plot (Fig. 4.5) indicates that the T.V. camera-based portal imaging system is not quantum-noise limited. This result confirms the conclusions of an earlier study.\(^{10}\) Note that these results agree with those obtained by Mah et al.,\(^{30}\) where they suggest that x-ray quantum noise can only be observed when the lens collection efficiency is dramatically increased. To remove these secondary quantum sinks, \( P(\omega) \) increases of at least 1.8, 18, and 392 are required to ensure that the primary quantum sink dominates for 0.25, 0.5, and 0.74 cycles/mm spatial frequencies, respectively. This can be achieved by increasing the gain factors or MTFs in the optical chain of the imaging system. However, changes in the design of the T.V. camera-based system would have difficulty eliminating all of the secondary quantum sinks. The QAD analysis shows that the lens collection efficiency, \( g_0 \), is the major cause of signal loss. The value of \( g_0 \) can be increased by increasing the area of the light sensor. Replacing the current T.V. camera (useful target area of 2.06 cm\(^2\)) by a large-area charge-coupled device (CCD) camera [useful target area of 37.21 cm\(^2\) has been reported (Dalsa Megasensor CA-D9-5120, Waterloo, Ontario)] would decrease the demagnification factor involved in the calculation of \( g_0 \), resulting in an increase of \( P(\omega) \) by a factor of about 18, assuming no changes in other gains
or MTFs. However, a large-area CCD may not be practical because of the large size of the lens required to cover such a large area light sensor. Furthermore, this increase, in combination with other changes in the current system design, is unlikely to remove the secondary quantum sinks for the highest spatial frequencies (> 0.7 cycle/mm) where an improvement factor of 392 is required. While eliminating additive noise and improving the MTF of the lens and T.V. camera combination is likely to have a significant and beneficial effect on image quality, it is unlikely for video-based portal imaging systems to eliminate all secondary quantum sinks for spatial frequencies above 0.5 cycle/mm.

The parameter sensitivity analysis performed on the derived expression for the DQE of the portal imaging system shows that the DQE can be affected, in order of decreasing efficiency, by changes in: (i) the detection efficiency, \( g_s \); (ii) \( S_s(\omega)/\eta \); (iii) the MTF of the lens and T.V. camera assembly, \( T_1(\omega) \times T_6(\omega) \); (iv) the optical gain, \( g_5 \); (v) either of \( g_3 \), \( g_6 \), or \( g_0 \); and, (vi) the MTF of the x-ray detector, \( T_2(\omega) \times T_4(\omega) \). Conversely, the noise associated with the optical gain affects the DQE of the current system only minimally. Therefore, improving the x-ray detector detection efficiency is the most efficient way to improve the current system, even though the lens collection efficiency is the most important cause of signal loss. Nonetheless, research groups are investigating the use of solid-state technology to eliminate the losses due to the poor lens collection efficiency. Already, the portal imaging system based on an amorphous silicon detector shows promise of significant improvements in QAD and DQE.\textsuperscript{33}

4.7 Conclusions

This chapter presents the first application of the QAD analysis theory to an existing imaging system. By analyzing each of the stages which constitute the T.V. camera-based portal imaging system, an expression for the spatial-
frequency dependent DQE was obtained for a T.V. camera-based portal imaging system. We have compared the DQE obtained from previously measured MTF and NPS with the DQE obtained from the QAD analysis theory and found good agreement.

The QAD analysis of our portal imaging system shows that the system is not x-ray quantum-limited, and that increasing the $P(\omega)$ by 18 is needed to remove secondary quantum sinks for spatial frequencies up to 0.5 cycles/mm; such an increase would bring the corresponding QAD lines of Fig. 4.5 above the primary quantum sink level. Also, the QAD shown in Fig. 4.5 demonstrates that the light collection efficiency of the lens is the most significant factor degrading image quality. The poor MTF associated with this lens further degrades image quality at high spatial frequencies. Sufficient increases in gain may be achieved by increasing the efficiency of the optical chain of the portal imaging system.

Our analysis shows that the most efficient ways to increase the DQE involve increasing the x-ray detection probability, reducing the noise added by the camera electronics and the video frame-grabber, and increasing the MTF of the lens and T.V. camera combination.
References


Chapter 5: OPTIMAL PHOSPHOR THICKNESS FOR PORTAL IMAGING

Abstract

A theoretical approach known as quantum accounting diagram (QAD) analysis has been used to calculate the spatial-frequency-dependent detective quantum efficiency (DQE) of two portal imaging systems: one based on a video camera and another based on an amorphous silicon array. The spatial frequency-dependent DQEs have then been used to determine indices of displayed and perceived image quality. These indices are figures of merit which can be used to optimize the design of linear imaging systems. We have used this approach to determine which of eight phosphor screen thicknesses (ranging between 67 and 947 mg/cm²) is optimal for the two designs of portal imaging systems. The physical characteristics (i.e., detection efficiencies, gains, and MTF's) of each of the eight x-ray detectors have been measured and combined with the physical characteristics of the remaining components to calculate the theoretical DQE's. In turn, the DQE's have been used to calculate theoretical indices of displayed and perceived image quality for two types of objects: a pelvis object and a point-like object. The maximal indices of displayed and perceived image quality were obtained with screen thicknesses ranging between 358 and 947 mg/cm², depending upon the imaging system design and the object being imaged. Importantly, the results showed that there is no single optimal screen thickness. The optimal thickness depended upon imaging task.

1 This chapter was accepted to Medical Physics, August 1996. The title of the article is "Optimal phosphor thickness for portal imaging" by J.-P. Bissonnette, I.A. Cunningham, and P. Munro.
(e.g., detecting large, low-contrast structures, as opposed to detecting edges and small structures). Nevertheless, the results showed that there were only modest improvements in the indices of image quality for phosphor screens thicker than 350-400 mg/cm².
5.1 Introduction

A number of on-line portal imaging systems are being developed to verify radiation field placement in radiation therapy. The most common of these video-based portal imaging systems - suffer from low light collection efficiency in their optical chain, which degrades image quality. One way to improve image quality is to increase the light generated by the imaging system by increasing the thickness of the phosphor screen which forms part of the x-ray detector. Unfortunately, increasing the thickness of the phosphor screen degrades spatial resolution, and hence image quality. Therefore, there must be an optimal screen thickness which balances the competing effects of light output and spatial resolution.

Two previous studies have attempted to optimize the thickness of phosphor screens for video-based portal imaging systems. In the first of these studies, an analysis of the signal-to-noise ratio (SNR) was performed for Gd$_2$O$_2$S:Tb screen thicknesses varying between 150 and 1000 mg/cm$^2$. This study was limited to the zero-frequency SNR of the x-ray detector alone. The second study used a simplified spatial-frequency-dependent analysis which made assumptions about the spatial resolution requirements for portal imaging and ignored spatial resolution degradations and noise sources other than those due to the x-ray detector. Clearly, better estimates of optimum phosphor screen thickness are required.

One method of evaluating system performance is by comparing the ability of different imaging systems to allow a user to detect structures of interest in images generated by the different systems. The detectability of simple patterns has been shown to be related with the SNR of an image. Thus, the SNR can be used as an "index of physical image quality" to compare different imag-
ing system designs quantitatively. If only the physical parameters of an imaging system [i.e., modulation transfer function (MTF), noise-power spectrum (NPS), and detective quantum efficiency (DQE)] are combined to calculate the SNR for a desired detection task, then an index of displayed image quality is obtained.\textsuperscript{12} If psycho-physical parameters related to the human visual system are incorporated in the calculation of the SNR, then an index of perceived image quality is obtained.\textsuperscript{12}

In this chapter, we present a method to optimize the design of linear imaging systems. We have applied this method to optimize the thickness of a Gd$_2$O$_2$S:Tb screen for portal imaging systems based on video cameras\textsuperscript{4,5} and amorphous silicon arrays.\textsuperscript{14} This method, which in principle can be applied to any linear imaging system, is based on the calculation of displayed and perceived image quality indices.\textsuperscript{12} These image quality indices, in turn, result from DQE's calculated using the quantum accounting diagram (QAD) theory.\textsuperscript{5,15}

5.2 Background

5.2.1 QAD and DQE analysis

The two portal imaging systems studied in this chapter (i.e., a system based on a TV camera and a system based on a hydrogenated amorphous silicon array) have been analyzed using the QAD theory.\textsuperscript{15} In this theory, linear imaging systems are described as a cascade of multiple stochastic spread, fluence gain, and additive noise stages, where the number of quanta leaving each stage becomes an effective input to the following stage. When the $i$th stage consists of a fluence gain ($\bar{g}_i$), the mean propagated fluence ($\Phi_i$) and NPS [$S_i(\omega)$] are given by\textsuperscript{15-17}

$$\Phi_i = \bar{g}_i \Phi_{i-1} \text{ and } S_i(\omega) = \bar{g}_i^2 S_{i-1}(\omega) + \sigma_{\bar{g}_i}^2 \Phi_{i-1},$$  \hspace{1cm} (1)
where $\bar{\Phi}_{i-1}$ is the mean fluence incident on stage $i$, $\sigma_{\Phi}^2$ is the variance in the average gain, and $S_{i-1}(\omega)$ is the NPS of the previous stage. The corresponding relationships for a stochastic spread stage are\textsuperscript{15-18}

$$
\bar{\Phi}_i = \bar{\Phi}_{i-1} \quad \text{and} \quad S_i(\omega) = \left| T_i(\omega) \right|^2 S_{i-1}(\omega) + \left[ 1 - \left| T_i(\omega) \right|^2 \right] \bar{\Phi}_{i-1}, \quad (2)
$$

where $T_i(\omega)$ is the MTF associated with the stochastic spread process.

In turn, the DQE of an imaging system can be expressed in terms of the average gain, MTF, and additive noise [i.e., $S_a(\omega)/\bar{\Phi}_i$] of individual stages. The theoretical expression for the DQE of any linear imaging system is\textsuperscript{15}

$$
DQE(\omega) = \frac{G^2 T_{out}^2(\omega) \bar{\Phi}_0}{S_{out}(\omega)} = \frac{1}{1 + \sum_{i=1}^{M} \left( \frac{1 + \varepsilon_g |T_i(\omega)|^2 + S_a(\omega) / \bar{\Phi}_i}{P_i(\omega)} \right)^{-1}}, \quad (3)
$$

where $\varepsilon_g$ is the Poisson excess [i.e., $\varepsilon_g = (\sigma_{\Phi}^2 / g_i - 1)$] associated with $\bar{g}_i$, $G$ is the average gain of the entire imaging system (also given by the product of all $\bar{g}_i$), $T_{out}(\omega)$ is the MTF of the entire imaging system [also given by the product of all $T_i(\omega)$], $S_{out}(\omega)$ is the NPS of the entire imaging system, $\bar{\Phi}_0$ is the mean x-ray fluence of the input radiation beam, $M$ is the number of stages, and $P_i(\omega)$ is the product of the gains and MTF's up to and including the $i$th stage, given by

$$
P_i(\omega) = \prod_{j=1}^{i} \bar{g}_j |T_j(\omega)|^2. \quad (4)
$$

Equation (4) can be interpreted as an effective number of quanta (x rays, optical quanta or photoelectrons) which propagate the signal through each of the image-forming stages of an imaging system, at each spatial frequency. A plot of $P_i(\omega)$ as a function of stage number $i$ yields the quantum accounting diagram.\textsuperscript{15} The stage with the lowest $P_i(\omega)$ value corresponds to the quantum sink for the corresponding spatial frequency.
5.2.2 Image quality indices

As stated in Sec. 5.1, the probability of detection of a signal by a human observer correlates with the SNR, which, in turn, depends on the imaging task and the physical characteristics of an imaging system.\textsuperscript{10-13,19-21} Loo et al. have performed a study to determine which one of ten different models of the SNR correlates best with human observer performance.\textsuperscript{12} In this study, two types of models have been defined: \textit{displayed} models, which assume that the human visual system is a perfect observer, and \textit{perceived} models, where the physical properties of the human visual system are accounted for. Loo et al. have found that the models which correlate best with human observer performance are based on the statistical decision theory.\textsuperscript{12} The square of the one-dimensional SNR derived from this theory is\textsuperscript{12}

\[
SNR_{SD}^2 = \frac{\left( \int_0 S^2(\omega) T_{out}^2(\omega) d\omega \right)^2}{\int_0 S^2(\omega) T_{out}^2(\omega) S_{out}(\omega) d\omega}
\]

(5a)

where \( S(\omega) \) is the Fourier-transform of the object being imaged. From the middle term of Eq. (3), Eq. (5a) can also be expressed as

\[
SNR_{SD}^2 = \frac{\left( \int_0 S^2(\omega) T_{out}^2(\omega) d\omega \right)^2}{\int_0 \frac{G^2 S^2(\omega) T_{out}^2(\omega) \Phi_0 d\omega}{DQE(\omega)}}
\]

(5b)

where \( DQE(\omega) \) is the DQE of the imaging chain. The index of perceived image quality derived from the statistical decision theory model is given by\textsuperscript{12}

\[
SNR_{SD}^2 = \frac{\left( \int_0 S^2(\omega) T_{out}^2(\omega) T_{sys}^2(\omega) d\omega \right)^2}{\int_0 \frac{G^2 S^2(\omega) T_{out}^4(\omega) T_{sys}^4(\omega) \Phi_0 d\omega}{DQE(\omega)}}
\]

(6)
where $T_{\text{eye}}(\omega)$ is the MTF of the human visual system. Of the physical parameters describing the human visual system, only $T_{\text{eye}}(\omega)$ is considered in this work. This is because the other parameters, which include the light collection efficiency of the eye and internal noise in the visual system, may be uncertain or unknown. While the magnitude of the internal noise in the human visual system can be estimated for the special case where system noise is Poisson-distributed,\textsuperscript{22} this estimation may not be appropriate for video-based portal imaging systems, where noise is not Poisson-distributed.\textsuperscript{23}

5.2.3 Optimization of phosphor screen thickness

The phosphor screen thickness which optimizes the detectability of structures of interest for portal imaging system occurs where a plot of image quality index (i.e., $\text{SNR}_{sd}^2$) versus screen thickness reaches a maximum. Thus, to determine the optimal screen thickness, we must determine the DQE of the portal imaging system, as required for the calculation of image quality indices [see Eqs. (5b) and (6)] for all of the screen thicknesses evaluated in this study.

5.3 QAD description of the portal imaging systems

5.3.1 Video-based system

The video-based system used in this study, developed at the London Regional Cancer Centre, has been described in detail previously.\textsuperscript{5,24} Briefly, this system consists of a TV camera which detects the light signal emitted from a copper plate/phosphor screen detector through a 45° mirror and a large-aperture lens. For the purposes of the QAD analysis, the system has been divided into a number of stages, as shown in Fig. 5.1 and detailed previously.\textsuperscript{5}
The theoretical equation for the DQE of the video-based portal imaging system \( [DQE_{\text{video}}(\omega)] \), obtained using the QAD analysis, is given by\(^5\)

\[
DQE_{\text{video}}(\omega) = \frac{g_1 T_2^2(\omega)}{1 + \frac{\varepsilon_{2s} + 1 + S_4(\omega)/\Phi}{g_3 g_5 g_6 T_4^2(\omega) T_7^2(\omega) T_6^2(\omega)}}.
\]  

(7)

where \( g_1 \) is the probability that an incident x-ray deposits some energy in the phosphor screen, \( T_2(\omega) \) is the MTF of the x-ray detector due to the transport of high-energy particles, \( g_2 \) is the average number of optical quanta generated per interacting x-ray, \( \varepsilon_{2s} \) is the Poisson excess associated with \( g_3 \), \( T_4(\omega) \) is the MTF associated with the spread of optical quanta in the phosphor screen, \( g_5 \) is the probability that an optical quanta generated inside the phosphor screen escapes through the exit surface of the phosphor, \( g_6 \) is the light collection efficiency of the lens of the video camera, \( T_7(\omega) \) is the MTF of the lens of the TV camera, \( g_8 \) is the light detection efficiency of the video camera, \( T_9(\omega) \) is the MTF of the TV camera, and the term \( S_4(\omega)/\Phi \) describes the relative amount of additive noise from the electronic components. The DQE of the video-based portal imaging system, weighted by the MTF of the human eye, is given by

\[
DQE_{\text{video,\,w}}(\omega) = DQE_{\text{video}}(\omega) T_{\text{eye}}^2(\omega).
\]  

(8)

5.3.2 System based on an amorphous silicon array

One device which may overcome the bulkiness and low light collection efficiency of TV camera-based portal imaging systems is the amorphous silicon array.\(^{14}\) We have modeled a portal imaging system based on this array as a series of nine cascaded stages [see Fig. 5.1]. Using Eq. (3), the theoretical DQE of a portal imaging system based on this array \([DQE_{\text{array}}(\omega)]\) is

\[
DQE_{\text{array}}(\omega) = \frac{g_1 T_2^2(\omega)}{1 + \frac{\varepsilon_{2s} + 1 + S_4/\Phi}{g_3 g_5 g_6 T_4^2(\omega) T_6^2(\omega)}}.
\]  

(9)
Figure 5.1: Block diagram showing the stages which have been used to model, using the QAD theory, (A) the video-based portal imaging system and (B) the portal imaging system based on the amorphous silicon array. The MTF of the human visual system is required for the computation of an index of perceived image quality.
where $g_1$, $T_2(\omega)$, $g_3$, $\varepsilon_g$, $T_4(\omega)$, and $g_5$ are as defined in Eq. (7). The primed quantities denote QAD parameters which apply to the amorphous silicon array. These quantities are the probability that a light quantum exiting the phosphor screen is intercepted by the sensitive region of the amorphous silicon array, $g'_e$, the probability that an incident light quantum releases an electron-hole pair in the array, $g'_f$, (the present study assumes that $g'_f$ explicitly includes the loss of optical quanta due to the mismatch of the indices of refraction of the amorphous silicon array and the phosphor screen) and the intrinsic MTF of the amorphous silicon array, $T_e^{2r}(\omega)$. The expression for the DQE of this system, weighted by the MTF of the human eye $[DQE_{array,w}(\omega)]$, is

$$DQE_{array,w}(\omega) = DQE_{array}(\omega) T_{eye}^2(\omega).$$

The QAD parameters for the amorphous silicon array have been described in the literature, and summarized here (we have chosen to model a 450 $\mu$m pitch array since it is the only array for which all the QAD parameters are available in the literature). The value of $g'_e$, given by the fill factor of the amorphous silicon array, is 0.62. The value of $g'_f$ is 0.63 for the light emitted by Gd$_2$O$_2$S:Tb, and $T_e^{2r}(\omega)$ is obtained from Ref. 25. Finally, the noise added by the electronic components $(S_a/\Phi)$ is obtained by dividing the measured dark noise per pixel (in Ref. 26, the dark noise for a single element of the amorphous silicon array is 11,200 electrons) by the number of charge-carriers created in a single monitor unit exposure from the 6 MV x rays (1 monitor unit $= 1$ cGy dose to the isocentre for a 10x10 cm$^2$ field at a depth of 5 cm in water). Since no spatial-frequency-dependent NPS for dark noise is available, we have assumed that $S_a/\Phi$ is constant for all spatial frequencies.

5.4 Materials and methods
5.4.1 X-ray detectors

Eight x-ray detectors, each consisting of a 1 mm copper plate bonded to a layer of Gd$_2$O$_2$S:Tb, were evaluated in this study. Six of these detectors, with phosphor thicknesses of 221, 273, 358, 404, 721, and 947 mg/cm$^2$, were supplied by the Eastman Kodak research laboratories. The other detectors were commercially-available screens (Kodak Lanex Regular and Kodak Lanex Fast Back, corresponding to thicknesses of 67 and 134 mg/cm$^2$, respectively; Richard Jebo, Eastman Kodak) placed in contact with a 1 mm thick copper plate.

The six QAD parameters associated with each x-ray detector [and required by Eqs. (7) and (9)] were determined using the same methods as those described in Ref. 5. Three of these parameters were obtained from Monte Carlo simulations: (i) the x-ray detection efficiency ($g_1$), (ii) the average number of light quanta per interacting x ray ($g_3$), and (iii) the Poisson excess in $g_2$. Two other parameters were determined from MTF measurements: (i) the MTF due to the spread of high energy particles in the x-ray detector [$T_2(\omega)$], and (ii) the spread of optical quanta in the phosphor screen [$T_4(\omega)$]. The remaining QAD parameter, the probability that an optical quanta generated inside the phosphor screen escapes through the exit surface of the phosphor ($g_5$), was obtained from light output measurements.

5.4.1.1 Monte Carlo simulations

Monte Carlo simulations have been performed in order to calculate the absorbed energy distribution (AED) for 6 MeV x rays incident on the eight x-ray detectors considered in this study. The AED is a plot of the number of incident x rays which deposit an energy between $E$ and $E+\Delta E$ in the phosphor layer of the x-ray detector as a function of $E$ (see, for example, Fig. 3 in Ref. 5 or Figs. 6-8 in Ref. 27). All AED's were computed using a computer program, based on the
EGS4 Monte Carlo system, which has been described by Jaffray et al.\textsuperscript{27} Each AED was calculated using $10^6$ photon histories and the 6 MV x-ray spectrum of Kubsad et al.\textsuperscript{28} The user-adjustable parameters which affect the transport of high energy particles in the EGS4 system were set as described in Ref. 27. For each of the eight phosphor thicknesses, three independent AED's were computed (by using different sequences of random numbers) to estimate the uncertainties in the Monte Carlo simulations.

Three of the six QAD parameters needed to describe the eight x-ray detectors were determined from AED's. First, the probability that an incident x-ray interacts with the ray detector and leads to an energy-deposition event in the phosphor layer of the x-ray detector ($g_1$) was obtained by dividing the total number of energy deposition events which occurred in a single AED calculation by the total number of x-ray histories. Then, the gain in fluence due to the conversion of x-ray quanta to optical quanta ($g_3$) was determined from the mean energy absorbed per interacting x-ray, $\overline{E_{ab}}$. Multiplying $\overline{E_{ab}}$ by the energy conversion efficiency of Gd$_2$O$_2$S:Tb (0.15) and dividing by the mean energy of the light quanta produced by Gd$_2$O$_2$S:Tb (2.3 eV) resulted in $g_3$.\textsuperscript{5,9,29} Finally, the Poisson excess $\epsilon_p$ was calculated as $\epsilon_p = \sigma_p^2 / g_3 - 1$ (see Sec. 5.2.1).\textsuperscript{5}

5.4.1.2 Light output measurements and the escape fraction

To determine the fraction of the light exiting the x-ray detectors ($g_5$), the light output of the eight x-ray detectors was measured on exposure to 6 MV x-rays (Varian 2100c, Palo Alto, CA) with the emulsion side of Kodak EM-1 film placed in direct contact with the exit surface of the phosphor screens. The detector and film were placed in a light-tight container and exposed to a 5×5 cm$^2$ x-ray field at 130 cm from the x-ray source, yielding optical densities ranging between 0.75 and 3.4. The films were processed in a Kodak M6B X-Omat film
processor and the optical density of the films was measured with a film densitometer (X-Rite, Grandville, MA, model 301). The measured optical densities were corrected for the non-linear response of the film. This measurement was performed twice to reduce experimental uncertainty.

The relative light output obtained from these films is a measure of the product of the interaction probability \(g_1\), the number of optical quanta generated per interacting x ray \(g_3\), and \(g_5\). Therefore, dividing the relative light output (determined experimentally) by the product of \(g_1\) and \(g_3\) (determined by Monte Carlo simulations) yielded a quantity proportional to \(g_5\). These quantities were normalized so that the value of \(g_5\) for the 358 mg/cm\(^2\) screen was 0.4. (The value of \(g_5\) for the 358 mg/cm\(^2\) screen had been determined earlier.\(^3\))

5.4.1.3 MTF measurements

The MTFs of the eight x-ray detectors [i.e., \(T_2(\omega) \times T_4(\omega)\)] were measured using standard methods.\(^{4,5,23,30-34}\) Briefly, two steel blocks (60 cm thick) were clamped together to collimate the 6 MV x rays from a medical linear accelerator to a narrow (~ 60 µm) x-ray beam. This narrow beam was used to irradiate the x-ray detectors, and the resulting line-spread function (LSF) was recorded with Kodak EM-1, single emulsion film. To measure the tails of the LSFs accurately, five films were acquired, each exposed to a different x-ray dose (ranging between 1 and 729 monitor units). All films were processed in a Kodak M6B X-Omat film processor. The LSFs recorded on these films were digitized with a Perkin-Elmer PDS scanning densitometer, using an aperture which was either 10 or 20 µm, and a sampling increment which was either 5 or 20 µm. The resulting LSF's were processed using standard techniques.\(^{4,23,30,31,33}\) The MTFs were measured three times for the 358 and 947 mg/cm\(^2\) screens, twice for the
Lanex Regular and 721 mg/cm² screens, and once for the four remaining screen thicknesses.

The above technique was modified so that we could measure the MTF due to the spread of high energy particles (i.e., scattered x rays, electrons set in motion in the detector, and bremsstrahlung), $T_2(\omega)$. To prevent optical photons from reaching the film, a thin (~ 60 μm) sheet of opaque, black plastic film was inserted between the phosphor screen and the film. Five films (exposed to x-ray irradiations ranging between 90 and 24 300 monitor units) were used to estimate the tails of the LSF accurately. The MTF due to the spread of high energy particles was measured twice for the 358, 721, and 947 mg/cm² screen thicknesses. The MTF's for the five remaining screen thicknesses were estimated by linear interpolation between the measured MTF's and the MTF for a 1 mm copper plate alone, which was published earlier.33

The measurement technique presented in the above paragraph underestimates systematically the value of the MTF due to the spread of high energy particles [$T_2(\omega)$]. This is because $T_2(\omega)$ can only be measured at the exit surface of the screen, where the lateral spread of high energy particles is maximal, whereas $T_2(\omega)$ should reflect the spread of high energy particles at the point where light is generated in the phosphor screen. Also, the black plastic sheet inserted between the phosphor screen and the film further increases the spread of high energy particles. Therefore, our measurement technique overestimates the width of the measured LSF, and, consequently, underestimates $T_2(\omega)$.

The MTF due to the spread of optical photons inside the phosphor screen, $T_4(\omega)$, cannot be measured directly. For the purposes of this work, $T_4(\omega)$ is determined by dividing the MTF of the x-ray detector [i.e., $T_2(\omega) \times T_4(\omega)$] by the underestimated $T_2(\omega)$ (see paragraph above). Therefore, the $T_4(\omega)$ presented in this work are overestimated systematically.
5.4.2 Computation of image quality indices

For the two portal imaging systems considered in this study, the indices of displayed and perceived image quality were calculated using the statistical decision theory model described in Sec. 5.2.2. The DQE's required by this model were evaluated using Eqs. (7) and (9), and the MTF of the human visual system required by Eq. (6) was obtained from the literature (a viewing distance of 50 cm was assumed). All spatial frequencies were related to the plane described by the x-ray detector. For the video-based portal imaging system, Eq. (7) was used to determine the DQE in the direction perpendicular to the camera scan lines and, for the system based on an amorphous silicon array, Eq. (9) was evaluated along the direction of the data lines since the MTF in this direction is slightly lower than that in the direction of the FET control lines. Indices of image quality were determined for each of the eight x-ray detectors described in Sec. 5.4.1, using a fluence-to-dose conversion factor of $7.4 \times 10^{-8}$ cGy mm$^2$/x ray.

The calculation of the image quality indices requires that the object spectrum be known [see Eqs. (5) and (6)]. We have chosen to determine image quality indices for (i) a point-like object and (ii) anatomic structures found in an AP pelvic image. For the point-like object, the object spectrum was set to 1.0 for all spatial frequencies where the DQE was non-negligible. The indices calculated with such a spectrum would be representative of a very small object, such as radio-opaque markers.

The object spectrum for anatomic structures found in an AP pelvic image was obtained as follows. Ten portal films of an anthropomorphic phantom (PIXY phantom patient, Victoreen Inc., Carle Place, NY) were obtained using the 4 MV beam from a medical linear accelerator (Varian 600 C, Palo Alto, CA). The films were digitized, using a Konica KFDR-S laser film digitizer, with a
sampling increment of 175 μm. All of the digitized images were registered using a semiautomatic registration algorithm and subsequently summed to reduce film and x-ray quantum noise. The magnitude of the Fourier transform of this summed image gave the two-dimensional object spectrum for the pelvic image. The slice through the two-dimensional object spectrum which corresponded to the vertical direction was selected for the calculation of the indices of image quality for pelvic structures.

All of the computations described in this section were performed on a Power Macintosh 7100/80, (Apple computers, Cupertino, CA) using the Kaleidogaph software package (Synergy Software, Reading, PA), except for the Fourier transform of the summed pelvis image, which was performed on a Sun Sparcstation 10 (Sun Microsystems, Mountain View, CA) using the AVS5 software package (Advanced Visual Systems, Waltham, MA).

5.5 Results

5.5.1 X-ray detectors

The parameters determined using Monte Carlo simulations (i.e., \(g_1\), \(g_3\), and \(g_5\)) are listed in Table 5.1. The escape probabilities \(g_5\) listed in Table 5.1

<table>
<thead>
<tr>
<th>Phosphor screen</th>
<th>Interaction probability, (g_1)</th>
<th>Optical gain, (g_3)</th>
<th>Escape probability, (g_5)</th>
<th>Poisson excess in optical gain</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lanex Regular</td>
<td>0.0139 ± 1 × 10^{-4}</td>
<td>10 404 ± 5</td>
<td>0.86 ± 0.43</td>
<td>7900 ± 56</td>
</tr>
<tr>
<td>Lanex Fast Back</td>
<td>0.0193 ± 1 × 10^{-4}</td>
<td>16 927 ± 115</td>
<td>0.75 ± 0.38</td>
<td>14 095 ± 513</td>
</tr>
<tr>
<td>222 mg/cm²</td>
<td>0.0260 ± 2 × 10^{-4}</td>
<td>22 640 ± 112</td>
<td>0.55 ± 0.27</td>
<td>21 417 ± 370</td>
</tr>
<tr>
<td>273 mg/cm²</td>
<td>0.0296 ± 1 × 10^{-4}</td>
<td>25 303 ± 88</td>
<td>0.45 ± 0.22</td>
<td>25 139 ± 479</td>
</tr>
<tr>
<td>358 mg/cm²</td>
<td>0.0357 ± 1 × 10^{-4}</td>
<td>28 755 ± 122</td>
<td>0.40 ± 0.20</td>
<td>30 653 ± 261</td>
</tr>
<tr>
<td>404 mg/cm²</td>
<td>0.0391 ± 1 × 10^{-4}</td>
<td>30 381 ± 229</td>
<td>0.37 ± 0.18</td>
<td>33 491 ± 499</td>
</tr>
<tr>
<td>721 mg/cm²</td>
<td>0.0598 ± 1 × 10^{-4}</td>
<td>37 544 ± 307</td>
<td>0.23 ± 0.11</td>
<td>45 168 ± 474</td>
</tr>
<tr>
<td>947 mg/cm²</td>
<td>0.0735 ± 2 × 10^{-4}</td>
<td>40 746 ± 210</td>
<td>0.19 ± 0.09</td>
<td>49 836 ± 190</td>
</tr>
</tbody>
</table>

Table 5.1: Summary of the gains for each of the eight phosphor screen thicknesses considered in this study. These figures were obtained from Monte Carlo simulations and measurements of the relative light output. The Poisson excess in the optical gain is included for completeness.
are derived from Fig. 5.2. Figure 5.2 shows that the amount of light produced in the screen (dashed line) increases linearly with increasing phosphor thickness while the light emission from the screen (solid line) increases in a non-linear fashion. This difference is due to optical attenuation inside the phosphor.\textsuperscript{9,24}

Figure 5.3 shows the detector MTF [5.3 (A)] as well as the MTF due to radiation transport only [5.3 (B)], $T_2(\omega)$, and the MTF due to optical diffusion only [5.3 (C)], $T_4(\omega)$, when the detectors are irradiated by a 6 MV x-ray beam. As expected, the MTFs shown in Fig. 5.3 (A) decrease as the thickness of the phosphor layer increases,\textsuperscript{4,7} as do the MTFs due to the spread of high energy [Fig. 5.3 (B)]. However, the MTFs due to spread of optical quanta in the phosphor screen shown Fig. 5.3 (C) are identical, within experimental uncertainty, for spatial frequencies above 0.7 cycles/mm. Figures 5.2 and 5.3 show that, for screen thicknesses beyond 404 mg/cm$^2$, there is a modest gain in light output, but the spatial resolution of the detectors continues to degrade while the blur due to optical diffusion inside the phosphor screen remains constant. Note that, as mentioned in Sec. 5.4.1, our measurement technique systematically underestimates the blur due to the spread of high energy particles. This underestimation causes an overestimate of the blur due to the spread of optical quanta. This overestimate is evident in Fig. 5.3 (C), where the MTFs derived for the thicker screens are higher than that of thinner screens at low spatial frequencies ($\omega < 0.7$ cycle/mm). The error bars shown in Fig. 5.3 indicate that the uncertainty in the MTF's of Fig. 5.3 (A) is smaller than ±0.06, and the uncertainty in the MTF's of Fig. 5.3 (B) is smaller than ±0.04.

5.5.2 QAD and DQE calculations

The frequency-dependent QAD's of the two portal imaging system considered in this study are shown in Fig. 5.4 for a screen thickness of
Figure 5.2: Relative light emission recorded by film when illuminated by light from the eight phosphor screen thicknesses (solid), and the relative number of quanta produced per incident x-ray (dashed), obtained from Monte Carlo simulations. Both quantities have been normalized to 0.4 at the 358 mg/cm$^2$ screen. The screens were irradiated with a 6 MV x-ray beam.

358 mg/cm$^2$. The thin lines represent the QAD for the video-based portal imaging system while the thick lines represent the QAD for the system based on the amorphous silicon array. Figure 5.4 clearly demonstrates that, for the video-based portal imaging system, the number of quanta at the last image-forming stage is lower than the number of x-rays which deposit energy in the phosphor layer of the x-ray detector for non-zero spatial frequencies.$^5$ On the other hand, for the system based on the amorphous silicon array, the number of quanta at the final image-forming stage is much higher than the number of x-rays which deposit energy in the phosphor layer of the x-ray detector. Therefore, for the spatial frequencies shown in Fig. 5.4, the video-based portal imaging system is not quantum-limited at the x-ray detection stage while the system based on the amorphous silicon array is.
Figure 5.3: (A) MTFs for the copper plate/phosphor screen x-ray detectors when irradiated with 6 MV x-rays. (B) Radiation MTFs for the copper plate/phosphor screen x-ray detectors when irradiated with 6 MV x-rays. The MTF for the copper plate alone was determined previously. (C) MTFs obtained by dividing the MTFs in (A) by the corresponding MTFs in (B), which reflect the blur due to the diffusion of optical quanta in the phosphor screens. The MTFs in (B) are underestimated, and thus cause an overestimate, at low spatial frequencies, of the MTFs in (C) [see Sec. 5.4.1]. The error bars represent one standard deviation about the mean.
Figure 5.4: The spatial-frequency-dependent QAD for the video-based portal imaging system and the system based on an amorphous silicon array for 0 (solid), 0.5 (dashed), and 1.0 (dotted) cycles/mm. The shaded area spans the stages involved with the x-ray detector (1 mm Cu + 358 mg/cm² phosphor screen) of the imaging system. Clearly, the video-based system suffers from severe quantum sinks in the optical chain while the system based on the amorphous silicon array does not.

The effect of screen thickness on the DQE of the two portal imaging systems considered in this study is shown in Fig. 5.5. In general, thicker screens improve the x-ray detection probability and the optical gain, and thus the DQE at low spatial frequencies. However, thicker screens have poor MTFs, and thus, at high spatial frequencies, the DQE obtained with these screens is lower than DQE’s obtained with some of the thinner screens. When the DQE is weighted by the MTF of the human eye, the DQE’s at low spatial frequencies are reduced due to the poor MTF of the human eye at these frequencies. (Note that the DQE’s weighted by the MTF of the human eye presented in Fig. 5.5 are not involved in the computation of indices of perceived image quality; see Sec. 5.4.2.) For identical screen thicknesses, the theoretical DQE’s calculated for the portal imaging system based on an amorphous silicon array are always higher than those calculated for the video-based system.
Figure 5.5: Theoretical DQE's for the video-based portal imaging system [(A) and (B)] and the system based on the amorphous silicon array [(C) and (D)] when irradiated with 6 MV x rays. Also shown are the theoretical DQE's weighted by the MTF of the human visual system. The error bars were determined from the uncertainty in the QAD parameters used for the calculation of these DQE's.
5.5.3 Indices of image quality

Figure 5.6 shows the indices of image quality computed for the two portal imaging systems considered in this study. For the video-based portal imaging system viewing a point object, the highest indices of displayed (solid) and perceived (dotted) image quality were obtained with the 947 and 721 mg/cm² screens, respectively. For pelvic structures, the maximal indices of displayed and perceived image quality were both obtained with the 947 mg/cm² screen. However, the error bars on Figs. 5.6 (A) and (B) show that there is no statistical difference in the indices of image quality for the 358, 404, 721, and 947 mg/cm² screens for the video-based portal imaging system.

Figure 5.6 also shows the image quality indices calculated for a portal imaging system based on an amorphous silicon array. Both the indices of displayed and perceived image quality are maximal at 358 mg/cm² for a point object, and at 947 mg/cm² for pelvic structures. Note that, for identical screen thicknesses, the indices of image quality obtained with the portal imaging system based on an amorphous silicon array are always higher than those calculated with the portal imaging system based on a video camera (see Fig. 5.6).

In general, the highest indices of image quality for pelvic structures are obtained with a screen thicker than for a point object. This is because most of the information contained in pelvic structures is concentrated in the lowest spatial frequencies ($\omega < 0.1$ cycles/mm), where degradation in contrast transfer due to increasing screen thickness is modest.
Figure 5.6: Indices of displayed and perceived image quality calculated for the video-based portal imaging system and the portal imaging system based on an amorphous silicon array. These displayed (solid) and perceived (dashed) indices were calculated when either a point object or a pelvic structures are imaged with 6 MV x rays. For the system based on the amorphous silicon array, the error bars are of the order of the symbol size.
Figure 5.7 shows images of a contrast-detail phantom which have been acquired, using the video-based portal imaging system, for four of the phosphor screen thicknesses evaluated in this study. This phantom consists of an aluminum plate with circular holes of different depths (4.6, 3.3, 2.4, 1.7, 1.3, 0.9, 0.7, 0.5, 0.4, and 0.3 mm) and different diameters (12.7, 9.8, 7.5, 5.8, 4.4, 3.4, 2.6, 2.0, 1.5, and 1.2 mm) arranged in rows and columns, respectively, such that object contrast decreases from left to right and object size decreases from top to bottom. These images have been acquired using identical 1 monitor unit irradiations and have been displayed using an identical number of gray levels. The image obtained with the Lanex Fast Back screen is the sharpest of the four images of Fig. 5.7, but the low light output of the screen prevents displaying low contrast structures with a high enough SNR for detection (only seven columns of holes can be seen). From Fig. 5.7, it is clear that increasing phosphor screen thickness increases the image contrast. However, while the 721 mg/cm² screen produces the highest display contrast, the low spatial resolution of this screen prevents the detection of low-contrast structures (again, only seven columns of holes can be seen while eight columns are visible with the 404 mg/cm² screen). Therefore, the visual evaluation of these contrast-detail images agrees reasonably well with the index of perceived image quality calculated for the video-based portal imaging system.
Figure 5.7: Images of a contrast-detail phantom acquired with the video-based portal imaging system using four different screen thicknesses.
5.6 Discussion

This chapter presents a method, based on the determination of indices of image quality calculated from the DQE of an imaging system, for the optimization of imaging system design. In principle, this method is applicable to any linear imaging system. However, the video-based portal imaging system is not shift-invariant over the entire field of view because of off-axis distortions in the MTF of the lens of the video camera. Fortunately, linearity and shift-invariance can be reasonably assumed for the central quadrant of images obtained with the video-based system.

We have used this method to optimize the phosphor screen thickness of the x-ray detector used in two portal imaging system designs: one based on a video camera, and another based on an amorphous silicon array. For a point object imaged with the video-based portal imaging system, the optimal screen thickness is 947 mg/cm\(^2\) screen when the human visual system is ignored, and 721 mg/cm\(^2\) when the human visual system is accounted for; for a point object imaged with the amorphous silicon array, the optimal screen thickness is 358 mg/cm\(^2\), whether the human visual system is accounted for or not. On the other hand, for pelvic structures, the highest indices of displayed and perceived image quality are all obtained with a 947 mg/cm\(^2\) screen for the two systems evaluated in this study. It is unclear whether increasing screen thickness beyond 947 mg/cm\(^2\) would further increase or decrease image quality indices.

One of the most gratifying aspect of our results is that the index of perceived image quality agrees well with our visual perceptions. Even more importantly, our results show that the optimum screen thickness depends on imaging task. The task of detecting large, low contrast structures (e.g., pelvic structures) requires thicker screens than that of detecting small structures or edges (e.g., point-like object). The results suggest that there is no single screen
thickness which is optimal for portal imaging and that many criteria (including imaging task) need be considered.

Fortunately, the results shown in Figs. 5.5 (A), 5.5 (B), 5.6 (A) and 5.6 (B) show that there are relatively modest changes in the indices of image quality as screen thickness is increased beyond 350-400 mg/cm². Therefore, increasing screen thickness beyond these values is unlikely to lead to significant benefits. Moreover, manufacturing of thicker screens becomes more and more challenging. Flaws and point defects become more common, adding a "fixed pattern noise" to the images. This noise is visually distracting and difficult to correct. Given these conditions, we feel that screen thicknesses in the range 350-400 mg/cm² are the most appropriate for video-based portal imaging systems.

One of the interesting observations from our study is how the spread of signals in the detector due to optical diffusion and the spread of signals due to the transport of high energy quanta change with screen thickness. Conventional wisdom (gained by experience with screens ≤ 400 mg/cm²) suggests that most of the losses in spatial resolution in copper plate/phosphor screen detectors are due to optical diffusion. However, the results shown in Figs. 5.5 (B) and (C) suggest that, for screens thicker than 400 mg/cm², optical diffusion plays an increasingly minor role. This is because while optical attenuation limits the distance that the optical quanta can be transported (see Fig. 5.2), no such limitations in particle transport occur for high energy quanta in these screens. These results have significance for designers attempting to replace phosphor screens with alternative scintillators. The results show that transport of high energy quanta as well as optical quanta will become a concern as the thickness of the scintillator is increased.

One limitation of our model is the systematic underestimate of the MTF due to the spread of high energy particles [T₂(ω)] on the DOE's calculated using
Eqs. (7) and (9). This underestimate has a negligible effect on the DQE of the video-based portal imaging system, but leads to an underestimate of the DQE of the system based on the amorphous silicon array for non-zero spatial frequencies. This is because, as shown by Fig. 5.4, the video-based portal imaging system suffers from severe secondary quantum sinks while the system based on the amorphous silicon array does not. In a system which is not x-ray quantum limited, electronic noise dominates and system gains are low. Under such conditions, both Eqs. (7) and (9) can be expressed as

$$DQE \propto T^2_s(\omega) T^2_a(\omega).$$

(11)

From Eq. (11), it is clear that it is not necessary to separate the MTF of the copper plate/phosphor screen into its radiation and optical components to determine the DQE of the video-based portal imaging system [in contrast with the measured MTF due to spread of high energy particles, the accuracy of the detector MTF [i.e., $T_2(\omega) \times T_a(\omega)$] is not affected by a systematic underestimate]. However, Eq. (11) cannot apply to the portal imaging system based on the amorphous silicon array since this system is x-ray quantum limited (see Fig. 5.4). Therefore, the systematic uncertainty in $T_2(\omega)$ has no impact on the accuracy of our optimization for the video-based portal imaging system, but may affect the accuracy of the optimization for the system based on the amorphous silicon array. The systematic underestimate in $T_2(\omega)$, which becomes more important as screen thickness increases, causes a systematic underestimate of the indices of image quality for the system based on the amorphous silicon array. Therefore, the optimal screen thicknesses for this imaging system may be higher than the thicknesses suggested by the results presented in this chapter.

This chapter has presented a technique, which merges the detailed analysis of the design of imaging systems (i.e., QAD theory) with the physical characteristics of the human visual system (i.e., MTF of the human eye) and the
psycho-physics of signal detection (i.e., statistical decision theory), for the optimization of the design of hypothetical or actual imaging systems. With this method, we have succeeded in demonstrating how simple (i.e., phosphor screen thickness) and radical (i.e., replacing the video camera by an amorphous silicon array) changes to the design of an existing portal imaging system influence its performance. This technique can presumably be applied to other linear imaging systems as well as evaluate novel imaging technologies. Therefore, this technique can not only assist imaging system designers, but result in improved portal imaging systems.

5.7 Conclusions

This chapter presents a method for the optimization of imaging system design which accounts for the MTF of the human visual system. This method, based on the calculation of indices of perceived image quality derived from DQE's calculated using the OAD theory, has been applied to find which of eight phosphor screen thicknesses is optimal for two portal imaging system designs. We have found that the optimal phosphor thickness depends on the object being imaged, and that ignoring the MTF of the human visual system may result in an inaccurate optimization. For the video-based portal imaging system, there is no statistical difference between the images of perceived image quality calculated for screens of thicknesses ranging between 358 and 947 mg/cm². However, practical limitations favor the lower limit of this range.

Both the DQE's and the indices of physical image quality calculated for an imaging system based on an amorphous silicon array are much larger than those calculated for a video-based portal imaging system. Therefore, future portal imaging systems based on an amorphous silicon array may be better suited for portal imaging than the current video-based technology.
References


Chapter 6: SUMMARY AND FUTURE WORK
6.1 Summary

The purpose of this thesis was to optimize the design (Chapters 3, 4, and 5) and use (Chapter 2) of a video-based portal imaging system. The imaging performance of this video-based system was compared with portal films and an amorphous silicon array-based portal imaging system.

Our results show that the image performance of a portal imaging system depends not only on the design of the imaging device itself, but also on other factors such as the size of the x-ray source of the linear accelerator and the location where the portal imaging is mounted on the accelerator. The optimization of radiographic magnification described in Chapter 2 demonstrated that the imaging performance of a portal imaging system can be improved without changing the design of the system. The results show that the optimal magnification for portal films and video-based portal imaging systems differ significantly: portal films should be placed as close to the patient as possible (radiographic magnification of 1.0) while a radiographic magnification of about 1.7 is required for video-based portal imaging systems since the projection of a larger image compensates for the lower spatial resolution of the video-based system. Our results show that optimizing radiographic magnification has a considerable benefit on image quality. Fortunately, due to engineering constraints, most commercially-available on-line portal imaging devices are mounted so that the radiographic magnification is in the range 1.3 - 1.7.

Optimizing the radiographic magnification of an imaging system, however, can not compensate for poor system design. In Chapter 3, we have evaluated a new x-ray detector, which was believed to be a high efficiency and high resolution alternative to the metal plate/phosphor screen-based x-ray detectors. Unfortunately, we found that this new detector, which consists of a thick,
high-density scintillating glass, is not suitable for portal imaging applications. Despite its higher quantum efficiency, the glass scintillator has low light output and spatially-variant resolution (caused by the divergence of x rays within the 12 mm thick detector itself). Since low light collection efficiency is the major limitation of video-based portal imaging systems and since portal imaging systems must be able to image large x-ray fields (where x-ray divergence is large at the edge of the field), the glass scintillator is not suited to portal imaging.

More importantly, we found that choosing imaging system components on the basis of one desirable parameter (e.g., x-ray quantum efficiency) without considering other relevant parameters wastes time, money, and effort when assessing whether a proposed modification to an existing imaging system is beneficial or not. Clearly, the optimization of the design of an imaging system requires a more efficient approach.

Chapters 4 and 5 present an approach for the optimization of the design of linear imaging systems. Chapter 4 describes the first experimental verification of a recently-introduced formalism [quantum accounting diagrams (QAD)] for optimizing system design. The QAD formalism considers how each individual component of an imaging system effects the detective quantum efficiency (DQE) of the entire system. The QAD analysis has been used to reveal the limitations and suggest efficient methods of improving the performance of the existing portal imaging system. The good agreement between the DQE calculated using the QAD formalism and the DQE measured experimentally (Fig. 4.6) gives confidence in the accuracy of the QAD analysis.

In Chapter 5, the QAD formalism has been extended to include indices of perceived image quality. This optimization approach not only accounts for all noise and spatial resolution characteristics of the imaging system itself, but also includes of the physical characteristics of the human observer. Furthermore,
this approach calculates a single figure of merit (rather than curves) which makes the selection of the optimal system design easier. We have applied this method to optimize the phosphor screen thickness for portal imaging systems based on a video camera and on an amorphous silicon array. Our results show that, in portal imaging, the optimal phosphor screen thickness depends on what object is being imaged, and that ignoring the physical characteristics of the human visual system may result in an inaccurate optimization. Therefore, one screen thickness cannot be optimal for all portal imaging situations. Furthermore, the results shown in Chapter 5 suggest that replacing the optical components of the video-based portal imaging system (i.e., lens and TV camera) by an amorphous silicon array detector may result in striking improvements in image quality.

Is optimizing the design and use of a video-based portal imaging system sufficient to overcome the limitations of such systems? While significant improvements in the DQE of a video-based portal imaging system can be achieved (Jaffray et al.\textsuperscript{1,2} have managed striking improvements in image quality by improving the efficiency of the optical chain and drastically reducing noise added by electronic components), current video camera technology cannot overcome the loss of information-carrying quanta in the optical chain of video-based portal imaging systems, especially at high spatial frequencies (i.e., \(\geq 0.7\) cycles/mm). Fortunately, novel technologies already eliminate such losses. Because of their high radiation resistance, large-area amorphous silicon arrays can be placed in direct contact with conventional metal plate/phosphor screen x-ray detectors, which results in a compact portal imaging device where about 40\% of the light emitted by the phosphor screen is collected. This is a considerable improvement over the poor efficiency encoun-
tered in the video-based system described in this thesis (~0.02%). Furthermore, the spatial resolution of the amorphous silicon array is much higher than that of the lens and T.V. camera combination of the video-based system.

The most exciting implication from Chapters 4 and 5 is that the image performance of any linear imaging system can be determined numerically, without the need for experimental measurements. Therefore, the beneficial or deleterious effect of modifying the characteristics of any stage in the imaging system can be predicted, without having to construct the modified imaging system first. This capability permits numerical evaluations to see how improved components would improve the overall performance of the imaging system. In addition, this approach allows visual modeling of the influence of these design changes. Following the QAD model of an imaging system, a computer program can simulate the passage of image-forming quanta through individual gain and blur stages, resulting in an accurate representation of the signal and noise encountered in a final image. Such a program is described in Sec. 6.2.4.

The analytical framework presented in this thesis can evaluate the impact of these novel imaging technologies on future portal imaging systems since many of the physical factors involved in imaging with high-energy x-rays are accounted for accurately. These factors include x-ray absorption and quantum noise, the physical size of the x-ray focal spot, the physical characteristics of system components, additive noise, as well as the spatial information contained in the x rays transmitted through anatomical structures. Therefore, not only has this framework shown that there is little room for improvement in the design of video-based portal imaging systems, designs involving novel technologies can be evaluated virtually and thus lead quickly to considerable improvements in portal image quality, with efficient use of time, effort, and money.
6.2 Future work

This thesis has investigated many of the physical parameters associated with portal imaging systems, presented methods to evaluate the merit of proposed modifications to an existing imaging system, and provided a deeper understanding of the many processes which lead to the formation of a portal image. The approaches presented in this thesis provide a framework which can, in principle, be applied to any other linear imaging system and allow the evaluation of imaging system designs before components are purchased. The following sections present some avenues for future work.

6.2.1 Improved understanding of copper plate/phosphor screen detectors

As mentioned in Chapters 4 and 5, the two main sources of uncertainty in the DQE calculated using QAD formalism, for video-based portal imaging systems, are associated with the copper plate/phosphor screen detector. The first significant source of uncertainty is that of the MTF due to the spread of high energy particles in the phosphor layer of the x-ray detector (This quantity is henceforth referred to as “radiation MTF”). As mentioned in Sec. 5.4.1.3, the radiation MTF is uncertain because this MTF cannot be measured at depth inside the phosphor screen. As explained in Chapter 5, this uncertainty has little effect on the DQE's calculated for the video-based system, but may affect the shape of the DQE calculated for a portal imaging system based on an amorphous silicon array. To help understand the effect of the radiation MTF on image quality, a more comprehensive model of the MTF of the x-ray detector must be developed. One approach may be the extension of current models of phosphor screens, where a screen can be considered as a composite of many thin layers, each with its own radiation MTF and MTF due to scatter of light quanta. These MTFs could be determined and combined for each layer to yield a more accurate
model the MTF of the copper plate/phosphor screen. Since the radiation MTF can only be measured experimentally at the exit surface of the screen, Monte Carlo simulations of the dose deposited by high energy particles in the phosphor screen are required to estimate reasonably the radiation MTF for each phosphor layer.

The second source of uncertainty is in the estimation of the probability that a light quantum emitted inside the phosphor screen escapes from the screen through the exit surface of the screen. In contrast with the radiation MTF, this escape probability has little effect on the DQE calculated for an imaging system based on the amorphous silicon array, but has a non-negligible effect on both the magnitude and, to a lesser extent, the shape of the DQE of the video-based portal imaging system (see Sec. 4.5.3). The combination of the radiative and optical spreading processes described in the paragraph above may result in a more accurate estimate of the escape probability since the amount of light generated in a single layer of the phosphor screen is related to the dose deposited in this layer. This needs to be studied.

6.2.2 QAD theory and index of perceived image quality

The QAD theory allows a more comprehensive model of the image-forming stages than that presented in Chapter 5. Ideally, the QAD model involving the psycho-physical characteristics of the human visual observer should include the physical characteristics of the display device\(^5\) as well as those the human visual system. Currently, many of these parameters are unknown (e.g., gain in fluence of the display monitor and internal noise in the human visual system) or uncertain (e.g., collection of light by the human visual system and adjustment of display contrast). Furthermore, viewing conditions may vary
greatly. Nonetheless, for a well-defined viewing condition (e.g., human observer at 50 cm from a display monitor with no environmental illumination), some of these parameters (e.g., the MTF and NPS of a display monitor, the amount of light emitted from the display monitor and collected by the iris of the human eye, ...) can be determined accurately from models or experimental work. By properly cascading those parameters, a comprehensive model of an imaging system, starting from the detection of an incident x-ray pattern to the internal noise in the human visual system, can be achieved. Such a model would allow a comprehensive optimization of the viewing environment.

6.2.3 General medical imaging

In this thesis, we have demonstrated the usefulness of the QAD theory, and we have expanded its use towards the determination of indices of image quality for portal imaging. The use of such methods can be easily expanded to any other systems which can be represented as a cascade of linear stages. These methods are attractive since the complicated statistical concepts involved with signal detection can be expressed, almost intuitively, in terms of the multiplication factors and point spread functions which make up the stages of any linear, isoplanatic imaging system. Therefore, any imaging system designer can use the tools presented in this thesis to determine which components are best suited for a given medical imaging application.

6.2.4 Image simulator

To complement the theoretical approaches described in Chapter 4 and 5, we have written Monte Carlo code to simulate images obtained with any linear imaging system. This "image simulator" allows the visual demonstration of
the beneficial or deleterious effects of a proposed system modification or allows
the comparison of alternative imaging technologies without the investment of
time, effort, and money involved in building prototype imaging systems. Simu-
lation can also demonstrate visually the sensitivity of image quality to specific
physical parameters or to a specific component of the system.

The simulation of the portal imaging system is performed as follows. First, a Poisson-distributed incident fluence is generated, with a user-defined average number of simulated x-rays per pixel. Any user-defined "object" can be present in the incident fluence. Then, these simulated x rays are propagated through binary selections, amplifications and stochastic convolutions,7-9 as required by the physical parameters of the system to be simulated. For a binary selection process, random numbers, ranging between 0 and 1, are generated for each incident quantum to determine whether the quantum is propagated to the next stage or not. For an amplification stage, each incident quantum is multiplied by a random amplification factor, which is distributed according to the probability density function of this factor. Finally, for a stochastic spread stage (i.e., blur), two random numbers are generated to scatter each incident quanta to a random location: one number for the distance and the other for the direction.

Using the QAD model of the video-based portal imaging system (see Tables 4.1 and 4.3, as well as Fig. 4.6), we have simulated images for each image-forming stage of this system (the two sources of blur associated with the x-ray detector are combined, and additive noise is neglected). Figure 6.1 shows the resulting image for each system component (i.e., x-ray detector, lens, and T.V. camera). All of the images are displayed using the same number of gray levels. Losses in image quality (i.e., decreasing contrast and resolution, and increasing noise) due to low detection probabilities and blur stages are evident.
Figure 6.1: Images generated from the Monte Carlo image simulator for each component of the video-based portal imaging system. Each image accounts for the various gain and blur stages of the imaging system. These images have been simulated using an average background of 500 quanta per pixel at the input stage. All the point-spread functions used in the simulation are assumed symmetric.
To demonstrate the utility of the image simulator, Fig. 6.2 shows images at the final image-forming stage of the video-based portal imaging system when only one of the physical parameters is changed. Fig. 6.2 (B) is the same as Fig. 6.1 (I), and Fig. 6.2 (A) was simulated using an escape probability of 0.9 instead of 0.4. Clearly, increasing the efficiency of the optical chain increases image quality. The images shown in Fig. 6.3 demonstrate the effect of modifying phosphor screen thickness on image quality.

6.3 Future of portal imaging

The DQE curves and the indices of image quality shown in Chapter 5 show quite clearly that using an amorphous silicon array may result in potentially striking improvements in portal image quality. This is because using the amorphous silicon array eliminates the secondary quantum sink and the poor spatial resolution associated with video-based systems. Furthermore, the amorphous silicon array offers a compact design which would be easy to attach to a linear accelerator gantry.
Figure 6.2: Demonstration of the effect of decreasing the escape probability of light generated in the phosphor screen from (A) 0.9 to (B) 0.4. The loss of contrast due to the lower escape probability is evident.
Figure 6.3: Demonstration of the increasing phosphor screen thickness on image quality from video-based portal imaging systems. These simulated images show that the high light output of the 947 mg/cm$^2$ does not compensate for the poor spatial resolution of this screen.

Another promising development for medical imaging in general, and portal imaging in particular, is amorphous selenium plates. Because of its non-crystalline structure, amorphous selenium is highly resistant to radiation damage, like amorphous silicon, and can thus be placed directly in the primary radiation beam to form images. Existing prototypes are built by depositing selenium onto a metal plate which acts as an electrode as well as radiation dose...
Figure 6.4: Schematic of the amorphous selenium plate system. After irradiation, the latent image on the amorphous selenium plate is read non-destructively using an electrostatic probe. The latent image is erased by exposing the plate to light.

Buildup material. Prior to irradiation, the selenium plate is charged uniformly, thus creating a uniform electric field of 10 V/µm in the amorphous selenium (see Fig. 6.4). This field causes the electric charge-carriers, which are created by the x-rays absorbed in the amorphous selenium plate, to drift towards both surfaces of the plate. A large fraction of these charge-carriers recombine, but a sufficient number of carriers reach the surfaces of the plate to dissipate the charge on the detector surface. After irradiation, a latent image, consisting of an array of electric charges which map out the structure of the object being imaged, remains on the detector. The latent image is read out, non-destructively, using a scanning line of electrostatic probes. Therefore, the amorphous selenium plate can be read out many times to yield many images which can be subsequently averaged to yield a low-noise portrait image, if necessary.

In addition to its relative simplicity and compactness, the amorphous selenium plate system has the potential to offer high resolution images. This is because the high resistivity of selenium severely limits the lateral migration of electrical charge-carriers. Also, this design does not involve optical stages, which are common to video-based and amorphous silicon array systems. Consequently, there is no loss of spatial resolution due to diffusing optical quanta.
Therefore, the spatial resolution of the amorphous selenium plate is limited only by (i) the transport of high energy particles created when x-rays interact with the plate and (ii) the size of the electrostatic probe. Furthermore, this technology is likely to produce very low noise images because of the electrostatic read-out of the image.\textsuperscript{10} Therefore, amorphous selenium plates may compete fiercely with amorphous silicon arrays as the detector of choice for future portal imaging systems.

6.4 Final remarks

In conclusion, this thesis describes tools the optimization of portal imaging systems. These tools integrate most of the various fundamental factors which limit the quality of portal images, including the physical size of the x-ray focal spot, x-ray absorption and quantum noise, the noise and spatial resolution characteristics of various system components (including those of the human observer), as well as the spatial information contained in the x-rays transmitted through anatomical structures. This thesis points out which physical parameters require improvements or have a significant impact on overall image quality, and presents theoretical tools to help system designers save time, effort, and money when examining alternative approaches for portal imaging. The methods and results presented in this thesis will lead to improved portal image quality, and therefore to improved radiotherapy treatment field verification, local tumor control, and, hopefully, better treatment outcome.
References


APPENDIX A: DERIVATION OF THE THEORETICAL DQE FOR THE VIDEO-BASED PORTAL IMAGING SYSTEM
The detective quantum efficiency, DQE, of an imaging system describes the ability of that system to transfer signal and noise. The DQE is defined as

\[ DQE = \frac{\text{SNR}_{\text{out}}^2}{\text{SNR}_{\text{in}}^2}. \] (A.1)

where \( \text{SNR}_{\text{in}} \) and \( \text{SNR}_{\text{out}} \) are the input and output signal-to-noise ratios of the imaging system. The theoretical expression for the DQE, in terms of the physical parameters (image-forming "stages") which characterize the components of that imaging system, is given by

\[ DQE(\omega) = \frac{1}{1 + \sum_{i=1}^{M} \left[ 1 + \varepsilon_{g_i} \left( \frac{T_i(\omega)^2 + S_{A_i}(\omega)}{\Phi} \frac{\prod_{j=1}^{i} g_j T_j(\omega)^2}{\prod_{j=1}^{i} g_j T_j(\omega)^2} \right) \right]}. \] (A.2)

where \( \bar{g}_i \) characterises the mean gain factor for a fluence gain stage, \( T_i(\omega) \) is the MTF associated with a blurring stage, \( \varepsilon_{g_i} \) is the Poisson excess (i.e., \( \varepsilon_{g_i} = (\sigma_{g_i}^2/\bar{g}_i) - 1) \) associated with \( \bar{g}_i \), and \( S_{A_i}(\omega)/\Phi \) describes the relative additive noise injected by a system component. The video-based portal imaging system can be modelled as a number of gain, blur, and additive noise stages. With each stage is associated a Poisson excess which represents the relative amount by which the variance of a physical process is in excess of a process which follows Poisson statistics. For a process which follows Poisson statistics, the Poisson excess is 0. For a binary selection process, described by the probability that a given input quantum is propagated to the next stage, \( g_i \), the Poisson excess is given by \( -g_i \), as a consequence of the binomial theorem. For a deterministic process (i.e., a noise-free gain), the Poisson excess is always -1.

In this appendix, the derivation of the theoretical DQE of the video-based portal imaging system is presented, stage-by-stage. The first stage is the probability that an x-ray photon deposits energy in the phosphor layer of the phosphor layer of the x-ray detector, \( g_1 \). This gain stage is a binomial selection proc-
ess. Therefore, the Poisson excess in $g_1$ is $-g_1$, and the DQE after this stage is given by

$$DQE(\omega) = \frac{1}{1 - \frac{1 - g_1}{g_1}} = g_1$$  \hspace{1cm} (A.3)

The second stage is the blur in the x-ray detector cause by the transport of high energy particles, $T_2(\omega)$. The Poisson excess associated with this blur stage is $-1$, and the theoretical DQE after this stage is given by

$$DQE(\omega) = \frac{1}{1 + \frac{1 - g_1}{g_1} + \frac{1 - T_2^2(\omega)}{g_1 T_2^2(\omega)}} = g_1 T_2^2(\omega).$$  \hspace{1cm} (A.4)

The third stage is the gain, $g_3$, associated with the conversion of x-ray quanta to optical quanta. Since this gain process is neither a Poisson or a binary selection process, its Poisson excess is given by $\varepsilon_{g_3}$, and the theoretical DQE after this stage is

$$DQE(\omega) = \frac{1}{1 + \frac{1 - g_1}{g_1} + \frac{1 - T_2^2(\omega)}{g_1 T_2^2(\omega)} + \frac{1 - \varepsilon_{g_3}}{g_3 T_2^2(\omega)}} = \frac{g_1 T_2^2(\omega)}{1 + \frac{1 + \varepsilon_{g_3}}{g_3}}.$$  \hspace{1cm} (A.5)

The fourth stage is the blur in the x-ray detector cause by the transport of optical quanta, $T_4(\omega)$. Again, the Poisson excess associated with this blur stage is $-1$, and the theoretical DQE after this stage is given by

$$DQE(\omega) = \frac{1}{1 + \frac{1 - g_1}{g_1} + \frac{1 - T_2^2(\omega)}{g_1 T_2^2(\omega)} + \frac{1 - \varepsilon_{g_3}}{g_3 T_2^2(\omega)} + \frac{1 - T_4^2(\omega)}{g_3 T_2^2(\omega) T_4^2(\omega)}} = \frac{g_1 T_2^2(\omega)}{1 + \frac{\varepsilon_{g_3}}{g_3} + \frac{1}{g_3 T_4^2(\omega)}}.$$  \hspace{1cm} (A.6)

The fifth stage, the probability that a light quantum emitted inside the phosphor screen exits the screen through the exit surface ($g_5$), is a binary selection process. The Poisson excess is $-g_5$, and the theoretical DQE after this stage is given
by (the intermediate terms of the denominator, which are identical to those in preceding equations, are omitted for conciseness)

\[ DQE(\omega) = \frac{1}{1 + \frac{1 - g_1}{g_1} + \ldots + \frac{1 - g_5}{g_1g_2g_5}\frac{1 - g_6}{T_2^2(\omega)T_2^2(\omega)}} \]

\[ = \frac{g_1T_2^2(\omega)}{1 + \frac{g_3T_2^2(\omega)}{g_1g_3g_5g_6T_1^2(\omega)T_4^2(\omega)}} \]  \hspace{1cm} (A.7)

The sixth stage, described by the probability that a light quantum exiting the phosphor screen \((g_6)\), is also a binomial process. The Poisson excess is \(-g_6\), and the theoretical DQE after this stage is given by

\[ DQE(\omega) = \frac{1}{1 + \frac{1 - g_1}{g_1} + \ldots + \frac{1 - g_6}{g_1g_3g_5g_6}\frac{1 - g_6}{T_2^2(\omega)T_4^2(\omega)}} \]

\[ = \frac{g_1T_2^2(\omega)}{1 + \frac{g_3T_2^2(\omega)}{g_1g_3g_5g_6T_4^2(\omega)}} \]  \hspace{1cm} (A.8)

The seventh stage is the blur due to the lens of the T.V. camera, \(T_7(\omega)\). The Poisson excess associated with this blur stage is \(-1\), and the theoretical DQE after this stage is given by

\[ DQE(\omega) = \frac{1}{1 + \frac{1 - g_1}{g_1} + \ldots + \frac{1 - T_7^2(\omega)}{g_1g_3g_5g_6T_2^2(\omega)T_4^2(\omega)}} \]

\[ = \frac{g_1T_2^2(\omega)}{1 + \frac{g_3T_2^2(\omega)}{g_1g_3g_5g_6T_4^2(\omega)}} \]  \hspace{1cm} (A.9)

The last gain stage, the probability that a light quantum incident on the lens of the T.V. camera generates a photoelectron in the camera \((g_6)\), is a binary selection process. The Poisson excess is \(-g_6\), and the theoretical DQE after this stage is given by
\[\text{DQE}(\omega) = \frac{1}{1 + \frac{1}{g_1} + \ldots + \frac{1 - g_\theta}{g_1 g_3 g_5 g_6 T_2^2(\omega) T_4^2(\omega) T_7^2(\omega)}} \cdot (A.10)\]

\[= \frac{\epsilon_{g_2}}{1 + \frac{1}{g_3 g_5 g_6 T_4^2(\omega) T_7^2(\omega)}} \cdot (A.11)\]

The last blur stage, which is associated with the T.V. camera itself \(T_6(\omega)\), has a Poisson excess of \(-1\), and the theoretical DQE after this stage is given by

\[\text{DQE}(\omega) = \frac{1}{1 + \frac{1 - g_1}{g_1} + \ldots + \frac{1 - T_6^2(\omega)}{g_1 g_3 g_5 g_6 T_2^2(\omega) T_4^2(\omega) T_7^2(\omega) T_9^2(\omega)}} \cdot (A.11)\]

\[= \frac{\epsilon_{g_2}}{1 + \frac{1}{g_3 g_5 g_6 T_4^2(\omega) T_7^2(\omega) T_9^2(\omega)}} \cdot (A.11)\]

The last stage involves the addition of noise from electronic components. The final DQE is therefore given by

\[\text{DQE}(\omega) = \frac{1}{1 + \frac{1 - g_1}{g_1} + \ldots + \frac{1 + S_\omega(\omega)/\Phi}{g_1 g_3 g_5 g_6 T_2^2(\omega) T_4^2(\omega) T_7^2(\omega) T_9^2(\omega)}} \cdot (A.12)\]

\[= \frac{\epsilon_{g_2}}{1 + \frac{1 + S_\omega(\omega)/\Phi}{g_3 g_5 g_6 T_4^2(\omega) T_7^2(\omega) T_9^2(\omega)}} \cdot (A.12)\]

Equation (A.12) shows the theoretical expression which was used in Chapters 4 and 5 to describe the DQE of the video-based portal imaging system. The derivation of the DQE for a portal imaging system based on an amorphous silicon array is similar, and is not explicitly performed in this Appendix.