Radiologic Image Assessment Using Information Loss Theory by Specially Designed Low Contrast Detail Phantoms and Extending it to CT

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Abstract

All radiographic imaging modalities are imperfect; they produce images that are affected to some degree by loss of object information or details. This can be addressed by exposing the subject being imaged to beams of different exposure values or extending the exposure time, but this result in an increase in the radiation dose delivered to the subject, a factor which must be minimised in clinical applications. In order for radiographic imaging modalities to be calibrated to minimise the dose delivered to a patient while still capturing images of sufficient detail to facilitate diagnosis, various methods of image quality (IQ) assessment have been developed which determine the efficiency and low contrast detectability of these modalities.

A further approach to the reduction of the dose delivered is applied in the latest types of Computed Tomography (CT) modalities, in which incomplete slices are obtained and utilised in the reconstruction of images. These slices are made through very short periods of time which also reduces the dose to the patient but leads to some artefacts in the image in addition to some extra information loss. IQ assessment is therefore critical to these new types of CT modalities as well [1].

Assessing IQ is normally either done by using equipment (objective method) or by visualisation of images by professionals (subjective method) [more information about different methods of image quality evaluation is discussed in (Chapter 2, Section 2.8)]. Objective assessments of image quality associated with diagnostic imaging systems are most often equipment-based, such as noise analysis or modulation transfer functions [2]. These methods do not consider the effects of the image assessor, who is typically a radiologist, nor the effects of the viewing system and conditions.
Image quality can also be assessed subjectively. The subjective method is based on the observer’s perception of the assessor. This method requires multiple observers who individually identify a visible object (threshold details) for every detailed parameter available in the image. The human decision criteria are considered a fundamental element, as they can be included in the imaging chain when evaluating image quality, due to their crucial role in the medical diagnosis process. In this method, the radiologists usually assess the diagnostic images with the receiver operator characteristics (ROCs) to compare the performance of different imaging systems [more details about the ROC in (Chapter 2, Section 2.8.3)]. Despite the fact that ROC analysis considers the whole imaging chain, such as human observers and equipment, this analysis is a time-consuming process and cannot be readably adopted as a quality assurance (QA) method in a busy clinical practice [3].

The most common alternative approach to assess IQ is the use of a contrast detail phantom (CDP). A CDP can provide useful information on contrast detail detectability and is considered the most reliable form of IQ assessment, particularly in low-contrast conditions [3]. In fact, the CDP is referred to as a low-contrast detail (LCD) phantom, and the commercially available phantom is called CDRAD 2.0. The CDRAD phantom is made of acrylic (Perspex; polymethyl methacrylate) which is 10 mm thick and in which 225 cylindrical holes of various sizes and depths are drilled. The diameter of the holes varies in size from 0.3 mm to 8 mm. This range is equally distributed across 15 depths. These depths range from 8 mm (providing high contrast) to 0.3 mm (providing low contrast). Hence, the CDRAD phantom uses the air–acrylic interface to create image contrast [3]. This method involves both equipment and observers.
This thesis is mainly focused on the use of CDPs in the evaluation of IQ in two X-ray-based modalities: Conventional radiography and CT. The evaluation of the former modality includes computed radiography (CR) and digital radiography (DR) systems. A modified CDRAD phantom which is based on the CDP approach is proposed and evaluated. This modified phantom utilises a smaller attenuation differential than unmodified CDRAD and it is more closely representative of the tissues found in the human body.

The three main aims of this thesis are then as follows:

**The first aim:**

To modify the current CDRAD phantom by replacing the air-filled holes fully with water or contrast media. Replacing the air-filled holes at the same phantom with the contrast media can create a gradation of contrast measurements that can be varied and extended by adding different amounts (concentrations) of contrast media into the holes (Chapter 5) because air attenuates much less radiation compared with other media, including Perspex. After creating a low contrast phantom by replacing the air-filled holes with water at the CDRAD, the LCD of these two interfaces (air-Perspex and water-Perspex) is investigated by utilising of information loss (IL) theory in the DR system (Chapter 6). The investigation of applying the IL theory with the CDRAD is extended to include the evaluation of the image quality of the CR system using two different techniques anti-scatter grid and non-grid on the unmodified CDRAD containing air filled holes (Chapter 7). Finally, the study of the grid effect includes the flat panel direct (DR) system by using the IQF_{inv} factor to assess the image quality of different DR systems with and without the grid by using the CDRAD phantom containing air-filled holes (Chapter 8).
The second aim:
To create a special contrast-detail phantom to evaluate the IQ and assess the LCD of CT scanners more efficiently than the commercially available Catphan phantoms by including a wide dynamic range that can be modified to assess any required level of LCD, which will be called CTCDP. Like the CDRAD phantom used in conventional radiography, the CTCDP incorporates a central slide that contains holes of different diameters. The diameter of these holes incrementally increases in size from the middle of the phantom to the phantom edge, as follows: 1.0 mm, 2.5 mm, 5 mm, 6 mm, 7.5 mm, 9.0 mm, 10.0 mm, 11.0 mm and 12.5 mm (more details about the CTCDP in Chapter 9). The detection of the LCD at this phantom is measured by two factors: IQF and IL (Chapters 9 and 10). This phantom complements the existing Catphan and extends its applicability to much lower contrast values and also it extends its contrast dynamic scale.

The third aim:
To investigate the effects of the contrast media on the CTDI value and relate it to the CT numbers by using theoretical and experimental methods. The theoretical method employed a new derivation of the known CTDI formula. This new derivation accounts for the presence of the contrast media as a factor when determining the dose enhancement. The experimental part includes the determination of the dose enhancement by using the contrast media and Gafchromic films (Chapter 11).

These studies will be of great value to the radiology community and to all the CT users because it develops a method of estimating the level of information loss during imaging procedures that significantly enhances the X-rays based modalities currently employed.
Declaration

I hereby certify that all of the work described within this thesis is the original work of the author; the work has not been submitted previously in whole or in part, to quantify for any other academic award; the content of the thesis is the result of work which has been carried out since the official commencement date of the approved research program; and any published (or unpublished) ideas and/or results from the work of the others are fully acknowledged in accordance with the standard referencing guidelines.

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Dedication

This thesis is dedicated to my parents, to my lovely wife Wala and my precious daughter Mayar for their fully support and understanding.
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Figure 11.1: Calculated dose enhancement factor from mass energy absorption coefficient for different beam energies and at different concentration of iodine contrast agent in the target.

Figure 11.2: The area of the ten pixel values measurement from the contrast enhancement phantom at different medium; iodine contrast media 20%, iodine contrast media 10%, distilled water and Perspex phantom.

Figure 11.3: CT gafchromic films; top sample and the rest as immersed in the reservoirs of various concentrations of the contrast media.

Figure 12.1: A) The prototype samples of the contrast detail phantom, it shows two different filling materials. B) a radiograph image of the prototype sample.
Publications


**General abbreviations and Acronyms**

**A**
Amorphous Selenium  a-Se · 20
Amorphous Silicon Thin Film Transistors  a-Si-TFT · 3
Analogue Digital Converter  ADC · 38
Area Under the Curve (AUC) · 53
As Low As Reasonable Achievable  ALARA · 5
Automatic Exposure Control  AEC · 86

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**B**
Binding Energies  BEs · 22

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**C**
Computed Radiography  CR · 2
Computed Tomography Angiography  CTA · 70
Computed Tomography Contrast Detail Phantom  CTCDP · 8
Computed Tomography CT · 3
Computed Tomography Dose Index  CTDI · 5
Computed Tomography Pulmonary Artery (CTPA) · 94
Contrast Detail Phantom  CDP · 6
Contrast Media  CM · 23
Contrast to Noise Ratio  CNR · 45
CsI  Cesium Iodide · 46

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**D**
Detective Quantum Efficiency  DQE · 20
Detector Element  Del · 39
Digital Radiography  DR · 2
Direct Digital Radiography  DDR · 64
Dose Enhancement Factor  DEF · 210
Dose Length Product  DLP · 114
Dual Source Computed Tomography  DSCT · 67

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**E**
Electrocardiogram  ECG · 77

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**F**
Field of Measurement  FOM · 70
Filtered Back Projection  FBP · 82
Flat Panel Detector  FPD · 46

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**G**
Gadolinium Oxysulphides  GOS · 46

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**H**
Hounsfield Units  HU · 3

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**I**
Image Plate  IP · 2
Image Quality factor inverse  IQFinv · 17
Image Quality factor  IQF · 9
Image Quality  IQ · 4
Image Receptor  IR · 2
Indirect Digital Radiography  IDR · 64
Information Loss  IL · 9
Iterative Reconstruction  IR · 82

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**K**
Kinetic Energy Released in Matter  KERMA · 48

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**L**
Low Contrast Detail  LCD · 7

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**M**
Modulation Transfer Function  MTF · 47
Multi Detector Computed Tomography  MDCT · 67

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**N**
Noise Power Spectrum  NPS · 47

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**P**
Photo Stimulated Phosphor  PSP · 156
Photostimulated Luminescence  PSL · 29
### Q

- **Quality Assurance (QA)** · 6
- **Quality Control (QC)** · 6

### R

- **Receiver Operator Characteristics (ROC)** · 6

### S

- **Scan Field of View (SFOV)** · 73
- **Scatter to Primary Ratio (SPR)** · 110
- **Screen Film (SF)** · 2
- **Signal to Noise Ratio (SNR)** · 6

### T

- **Thin Film Transistor (TFT)** · 28
- **Three-Dimensional (3D)** · 2
- **Total Information Loss (TIL)** · 15
- **Transmitted Information (TI)** · 50
- **Two-Dimensional (2D)** · 2

### V

- **Visual Grading Analysis (VGA)** · 20

storage phosphor screen SPS · 27
Storage Phosphor Screens SPSs · 20
Chapter 1 represents project overview, hypotheses, aims of the thesis, general outcomes and outline of the thesis.
Chapter 1 Project overview

1.1 Project overview

Projection radiography was established after the discovery of the X-ray by Wilhelm Rontgen in 1895 [4]. The image in projection radiography is a two-dimensional (2D) projection of the attenuation properties of all the three-dimensional (3D) tissues along the paths of the X-rays; in other words, the 3D information is collapsed into 2D. In clinical applications, an X-ray tube emits X-ray photons that interact with the patient. There are three alternatives for each photon: 1) It can penetrate the patient without interacting (such photons are termed primary photons); 2) It can interact with the patient and be completely absorbed by depositing its energy; 3) It can interact and be deflected from its original direction and may deposit part of its energy (such photons are termed secondary photons). Primary photons pass through the tissues without interaction, and then they are recorded by the image receptor (IR). Secondary photons produce a certain amount of background radiation, which degrades the image contrast. These types of photons are called scattered radiation [4] and are the product of X-ray interaction with matter, which is discussed in detail in (Chapter 2, Section 2.2).

A significant development in projection radiography technology has occurred in the last three decades, wherein digital IRs have been introduced [5]. These digital receptors offer high sensitivity, large dynamic range and lower noise compared with the traditional screen-film (SF) systems. The conventional SF systems have been replaced with digital systems of various types first with computed radiography (CR) and later with digital radiography (DR) [5]. The CR system was the first digital radiography system to be introduced and is based on an image plate (IP) which is coated on one side with a layer of photo-stimulable phosphor material. This IP retains the information of the incident photons (i.e. latent image), which can
be retrieved in CR by a stimulation process using a read-out-laser [5]. The DR system is characterised by a direct readout-matrix of electronic elements. These elements are made of thin layers of amorphous silicon thin-film transistors (a Si-TFT element). The DR systems are divided into two types according to material of the detector: 1) indirect conversion TFT detectors; and 2) a direct conversion system [5]. The CR and DR systems are discussed in details in (Chapter 2, Sections 2.3 - 2.6).

The most significant recent development in X-ray diagnostic radiology began with the announcement of computed tomography (CT) by Hounsfield in 1972 [4]. The image in CT is formed using the rotation of a well-collimated X-ray pencil beam around the patient. This beam is attenuated by the tissues along its path, and then the transmitted radiation is detected by the detectors in the gantry. To produce one projection, the tube detector assembly scans the target through a full rotation around it. This process is repeated at multiple viewing angles, with a minimum of 180 projections received with a rotational increment of 1 degree. From these projections, a 2D discrete distribution of the linear attenuation coefficient ($\mu_{\text{tissue}}$) is reconstructed as an image by computation process [4]. In practice, the Hounsfield units (HU) or CT numbers are used instead of $\mu_{\text{tissue}}$ where the Hounsfield unit is defined by [4]:

$$\text{HU} = 1000 \cdot \frac{\mu_{\text{tissue}} - \mu_{\text{water}}}{\mu_{\text{water}}}$$

(1.1)

$\mu_{\text{water}}$ is the linear coefficient of the water. The recent CT modalities are discussed in detail in (Chapter 3, Section 3.1).
The CT scan is superior to conventional radiography in two ways: 1) It produces a cross-sectional image that prevents the anatomical structures from being superimposed, which happens in projection imaging due to the compression of 3D structures onto 2D recording systems. 2) The sensitivity of the CT scan to low contrast in X-ray attenuation is greater than that normally achieved by all project imaging techniques by a factor of at least ten [6]. However, the CT modalities and digital radiography systems provide good spatial and contrast resolution, resulting in better image quality (IQ) for daily clinical cases, which allow for the detection of tiny and subtle lesions[7].

The IQ in digital radiography and CT modalities is influenced by spatial resolution, contrast and noise in the image. Spatial resolution is the ability limits of an imaging system to represent small distinct anatomic features within the object being imaged. The radiographic contrast is proportional to the magnitude of the signal variation between the structures being imaged and the surrounding areas and is therefore affected by the subject contrast [8]. Noise generates random variations of signals that can obscure useful information in diagnostic images. Accordingly, scatter radiation is a significant factor that can degrade the subject contrast and, in turn, the IQ [8]. The factors that affect the image quality of the digital radiography and CT modalities are discussed in detail in (Chapter 2, Section 2.7 and Chapter 3, Section 3.5).

The most commonly used technique for minimising the scatter radiation in digital radiography is the insertion of the anti-scatter grid between the patient and the IR. The grid selectively absorbs a large amount of scattered radiation and hence improves the image contrast. However, the use of an anti-scatter grid has the disadvantage of increasing patient
radiation exposure [2]. Anti-scatter grid is discussed in detail in (Chapter4, Section 4.4). This thesis will investigate the efficiency of utilising the anti-scatter grid technique on the IQ for the CR and the DR systems in Chapters 7 and 8.

The evaluation of IQ in radiological departments and the measures that are taken to ensure high IQ are critical for better diagnostic outcomes. The improvement of the IQ is usually associated with an increased dose to the patients and the public. In accordance with the As Low As Reasonable Achievable (ALARA) principle, to maintain a radiation dose that is as small as possible while providing adequate IQ, the dose must be optimised with the IQ [9]. A common technique to improve the IQ in CT modalities is to inject contrast media in to the patient. The effect of the radiographic contrast media is discussed in detail in (Chapter 4, Section 4.6). The enhancement of the iodine contrast media is associated with an increase in radiation dose [10, 11]. A number of studies have explored the potential radiation hazards of CT scans. According to Pearce et al. [12], there was one additional case of a brain tumour or leukaemia out of every 10,000 patients who had head CT scans and there is a significant correlation between cancer induction and CT scanning [12]. These studies indicate that potential risks from CT do exist, and this has led to demands to both establish a specific threshold and limit the radiation dose to the lowest possible level [13]. In this thesis, the effect of dose enhancement caused by iodine contrast media has been introduced into the computed tomography dose index (CTDI) formula theoretically and validated experimentally. This derived formula including the dose enhancement can be used in estimating the delivered dose to the public and patients. With the inclusion of the contrast media effect, the dose enhancement that was caused by the contrast media is also linked for the first time to the CT number in Chapter 11.
The assessment of the IQ for digital radiography and CT scan can be addressed by either objective or subjective methods. Objective methods are not affected by human perception and are based on the measurement of image data such as signal to noise ratio (SNR). These methods are used for routine quality control (QC), as they are suitable in detecting drifts in equipment performance. However, the medical diagnosis process is not only dependent on the formed image but also on the observer’s perception. Hence, radiologist and the radiographer are also important in assessing the quality of medical images. A number of studies have shown the importance of psychophysical factors in IQ as detected by the observer [14, 15].

The subjective method is based on human perception and decision making criteria. Multiple observers individually identify a visible object (threshold details) for every detailed parameter available in a given image. The human decision criteria are considered a fundamental element of the medical diagnosis process, and hence can be included in the imaging chain when evaluating the image quality [16, 17]. In this subjective method, radiologists usually assess the diagnostic images with the receiver operator characteristics (ROCs) to compare the performance of different imaging systems [9]. Different methods of IQ evaluation including ROC are discussed in detail in (Chapter 2, Section 2.8).

Although ROC analysis considers the whole imaging chain, such as human observers and equipment, this analysis is a time-consuming process and therefore not suitable for adoption as a quality assurance (QA) method in busy clinical practices. The most common alternative approach to assess IQ is the use of a contrast detail phantom (CDP) [3, 18]. A CDP can provide useful information on contrast detail detectability and is considered the most reliable
method for IQ assessment, particularly in low-contrast conditions [19]. In such conditions, the CDP is referred to as a low-contrast detail (LCD) phantom and the commercial phantom is called CDRAD 2.0. In this case, equipment and observer are combined in image assessment.

The CDRAD phantoms use the air-acrylic interface to create image contrast. This interface represents high subject contrast to the CDRAD phantom due to the high attenuation difference between the Perspex and air medium. A number of studies have used the CDRAD phantom in evaluating the IQ of digital radiography systems [16, 20, 21]. Because the CDRAD phantoms are applied to assess IQ in clinical cases, there is a demand to establish different scales of contrast subjects in CDRAD phantoms. CDRAD phantoms are discussed in details in (Chapter 4, Section 4.1). The rationale for this thesis stems from the potential to provide these different scales of contrast subjects by modifying the current type of commercially available CDP i.e. (CDRAD). It will investigate the effects of replacing the air in the CDP, changing the conventional air-Perspex combination to a water-Perspex and contrast media-Perspex combination. This study is expected to detect lower contrast levels of the phantom when using the water-Perspex combination. The proposed novel water or contrast media-Perspex interface with the CDRAD could provide a form of CDP that is more closely related to human tissue and thus better represents the contrast-detail imaging conditions encountered in radiology. Accordingly, this thesis will examine the feasibility of using the CDRAD phantom with all holes filled with contrast media or water as a multi-scale contrast-measuring device in most clinical institutions. This concept is discussed in detail in Chapter 5.
The LCD is a vital parameter in CT IQ control procedures. It is defined as the ability to distinguish between materials with similar attenuation properties. The detection of small objects can be affected by noise, particularly if the contrast is low. The measurement of LCD is obtained using phantom images such as the Catphan phantom, which contains objects of varied contrast and sizes [22]. These measurements are carried out with human observers scoring the images. Radiologists determine the smallest object of the lowest contrast that they are able to visualise [22]. The Catphan phantom is explained in detail in (Chapter 3, Section 3.7 and Chapter 9, Section 9.1). However, in brief, the Catphan phantom is limited by the relatively small number of different materials with varying subject contrast that can be selected. This makes the measuring scale short. Moreover, as Giron et al. [22] observed nearby objects could be included inside the samples. Consequently, there is a strong demand to create a specially designed phantom that has the ability to provide varied sizes of objects and can also accommodate different concentrations of contrast media with a larger phantom design to mimic larger body parts wherein more scatter radiation occurs, affecting the low contrast detectability and hence the image quality [23, 24]. This thesis aims to introduce such a newly designed computed tomography contrast detail phantom (CTCDP) to detect the LCD. The specially designed CTCDP is discussed in detail in (Chapter 4, Section 4.2).

The process of quantifying IQ and determining the image contrast objects begins with the observer attempting to visualise the different discs-shape diameters in the CDRAD phantom images. The observer then completes a feedback sheet about the detected object. The observer identifies the threshold visible thickness for each detail diameter and the location of the corner detail whenever it exists. Then, the observer scores are corrected using the four nearest neighbour’s method, as advised in the CDRAD manual [16]. The CDRAD correction and evaluation method is discussed in detail in (Chapter 4, Section 4.1).
The Image Quality factor (IQF) is calculated to assess the observer’s detectability:

\[
IQF = \sum_{i=1}^{15} C_i.D_{i,j}
\]  \hspace{1cm} \text{(1.2)}

Where \(D_{i,j}\) represents the threshold \((j)\) diameter in contrast column “\(Ci\)” that is observable. The summation is over all the columns.

Moreover, the CDRAD phantom can be used to evaluate the degradation of IQ as information loss (IL):

\[
H(x) = \sum p(x) \log_2 \left( \frac{1}{p(x)} \right)
\]  \hspace{1cm} \text{(1.3)}

\(p(x)\) : The probability.  \(H(x)\) : The information content. The probability \(p(x)\) is discussed in detail in (Chapter 6, Sections 6.3.2.3 and 6.3.2.4).

Information loss (IL) quantification is based on information theory, which was established by Shannon in 1948 [25]. Entropy is the key measure, and the predicated random different-value uncertainty can be measured by it [25]. Information entropy is discussed in detail in (Chapter
This theory can be applied to measure the IL in medical images using CDP [7] as follows:

\[
\text{IL} = H \text{ (no information loss)} - H_\infty
\]  

(1.4)

The Niimi et al. [7] study showed that the IL increased as the diameter of the phantom holes decreased [7]. Another study was done by the same group using the same method of measuring IL. In the latter experiment, however, they used a mammography CDP to obtain phantom images with a digital mammography system, and the results were the same [26]. To quantify the performances of CT modalities and digital radiography such as DR and CR, the IL theory will be applied to the CDRAD phantom and the specially designed CTCDP in this thesis.

This thesis outlines an innovative method that permits a calculation of the amount of IL, particularly in recent modalities of CT scanners and digital projection-type images such as the digital radiography and computed radiography. Chapters 6, 7 and 10 discuss the employment of the IL theory in evaluating the IQ. This project will greatly benefit CT users and the radiology community, who will be able to utilise this method to detect the level of IL during imaging procedures.
1.2 Hypotheses

1. In conventional radiography, which includes computed radiography (CR) and direct digital radiography (DR) systems, the CDPs are designed to provide useful information on contrast detail detectability and have been shown to be one of the most reliable and commonly adopted phantoms for IQ assessment, especially in low-contrast conditions. However, the subject contrast of the commercially available CDPs (CDRAD) phantom is relatively high because it represents attenuation differences between Perspex and air in contrast to the human tissues which shows low subject contrast.

2. Scattered radiation is considered a noise source that causes image degradation in radiography. It decreases the dynamic range of available X-rays on the exit side of the patient. As a result, the contrast subject and the signal-to-noise ratio (SNR) are decreased because they have insufficient signals and contain more quantum noise. The scanned slots of the DR detectors have scatter rejection capabilities and, hence, do not need a grid to eliminate the scattered radiation. However, anti-scatter grids must be used in the area detectors of DR and CR radiography when the scattered radiation is dominant.

3. One of the main reasons for the increasing usage of CT as an imaging modality is its ability to display images with information on adjacent tissues using detailed subject contrast. In other words, CT is highly efficient in LCD detectability. Assessment of the capability of a particular imaging chain in the range of contrast detectability level is performed using special-type phantoms, with the insertion of various high-density and low-density materials. However, variability in the density and atomic numbers of the materials in the commercially available CT low contrast detectability phantom limits the dynamic range of assessment.
4. Contrast media enhances the visualisation of the anatomical structures in radiological examinations; however, this process is associated with an increase in the radiation dose to patients.

1.3 Aims of the thesis

1. To modify the current commercial CDP for conventional radiography, including computed radiography (CR) and direct digital radiography (DR) systems, thereby expanding the phantom’s ability to provide different contrast scales and providing it with a low contrast scale better suited to mimic the real x-ray examination where the attenuation difference between adjacent soft tissues is very small.

2. To investigate the role of the anti-scatter grid at conventional radiography, including computed radiography (CR) and flat-panel direct digital radiography.

3. To design a contrast detail phantom that includes various sized diameters, and which can accommodate different concentrations of contrast media for testing the low contrast detectability of CT modality at any level of required detectability.

4. To investigate the effect of contrast media on the level of IL at two x-ray-based modalities: at direct digital radiography (DR) system and computed tomography (CT).

5. To calculate the dose enhancement in CT scan as a result of using contrast media and include it as a new factor on CTDI equation.

6. To utilise the information loss theory in assessing the quality of the phantoms images obtained from two X-ray-based modalities: conventional radiography [including computed radiography (CR) and digital radiography (DR)] and computed tomography (CT).
Two types of phantoms will be used in this project. The first phantom is commercially available to test the low contrast detectability of conventional radiography. This will include computed radiography (CR) and digital radiography (DR) systems. The second type of phantom will be designed to identify the IL in CT images through detecting the low contrast differences.

1.4 Outcomes/Benefits

1. Improving the identification of artefacts and IL which potentially lead to the misinterpretation of images in X-ray-based modalities: conventional radiography [including computed radiography (CR) and digital radiography (DR)] and computed tomography (CT).

2. Providing better knowledge about the IL and IQF methods in image quality assessment.

3. Quantifying the dose enhancement associated with using contrast media.

4. Creating a specially design phantom for quality assurance testing, especially for CT scans.

5. Supporting a new perspective in developing digital IQ.

1.5 Outline of the Thesis

This thesis is divided into 12 Chapters; the description of each Chapter is as follows:

Chapter 2 mainly reviews the basics of X-ray sources, including a description of most of the X-ray production forms. It explains in detail the two conventional radiography systems that are available at medical institutions; CR and DR systems, and describes the differences between them. It also discusses the factors that affect the IQ and the potential methods for evaluating the quality of the radiographic images achieved by subjective methods such as
radiologists and objective methods, such as automated low contrast detectability software analyses.

Chapter 3 reviews the recent CT scanners and the IQ parameters affecting the CT scan, such as spatial and contrast resolution. It demonstrates the factors that have an impact on the radiation dose and the IQ. Both Chapters 2 and 3 discuss the factors affecting the low contrast detectability in two X-ray-based modalities: conventional radiography [including computed radiography (CR) and digital radiography (DR)] and computed tomography (CT).

General information about all materials used in this thesis are summarised in Chapter 4. It also contains information about the phantoms that were utilised in this project, such as CDRAD 2.0 phantom and the CDP specially designed for a CT scan. The materials and methods sections in Chapters 5 – 11 contain specific details related to the individual experiment.

Chapter 5 introduces a CDP that is a modification of the commercially available CDRAD phantom (more information about the CDRAD phantom is discussed in Chapter 4, Section 4.1). The air in the holes was replaced with a medium that has absorption characteristics similar to the base material acrylic, hence decreasing the subject contrast material. Firstly, the air-filled holes within the CDRAD phantom were filled fully with distilled water. Subject contrast in the phantom is reduced when the holes are filled with water instead of air. This modified CDP more closely replicates the low-subject contrast commonly encountered in non-contrast radiologic examination where attenuation between adjacent soft tissues in the human body is very similar. Secondly, the air-filled holes within the CDRAD phantom were filled with a 30% concentration of iodinated contrast media (Omnipaque 350; GE Healthcare)
mixed with distilled water. The introduction of iodinated contrast media creates a CDP that more closely replicates the attenuation encountered in radiologic contrast examination. The results presented in this experiment prove that the conventional form of the CDRAD can be extended to include various ranges of attenuating materials filling the holes instead of only using air and that using a modified form of the CDP for projection imaging systems will be a valuable addition to radiology departments.

Chapter 6 expands on the work outlined in Chapter 5 by again using the modified CDPs. The CDRAD in this Chapter use two media: air-filled holes and water-filled holes. The commercially available CDRAD is used in the visual assessment of LCD and also can be used as a tool for maintenance/assessment of image quality [more information about LCD is discussed in (Chapter 2, Section 2.8.6)] [7]. This visual assessment is usually performed using an image quality factor (IQF) [more information about the calculation of the IQF is discussed in (Chapter 6, Section 6.3.2.2)]. Information loss theory has also been applied to the visual assessment of CDP images. Niimi et al. [7] introduced the term total information loss (TIL), and demonstrated that TIL was lower when higher radiation dose was used, and when the diameters of the air-filled holes within the CD phantom were larger [7]. In this Chapter, the TIL is employed to quantify the performance of two CDPs: one represents relatively high-subject contrast, air-Perspex CDP, and the other representing relatively low-subject contrast, water-Perspex CDP. This Chapter concludes that the material within the holes of the CDP influences TIL and IQF measurements. It was shown that the modified CDP, water-Perspex, with its lower inherent subject contrast that more closely represents soft tissue imaging in radiology, had higher TIL and IQF measurements. These higher measurements provide a more realistic account of TIL and IQF for soft tissue radiology imaging.
Chapter 7 employs the image quality factor (IQF) and the information loss (IL) theory (from Chapter 6) to assess the impact of applying the anti-scatter grid on the image quality obtained by CR system. In diagnostic X-ray imaging, the image contrast is degraded by scattered radiation affecting the image quality. One method to greatly reduce scattered radiation is to place the grid between the patient and the image receptor [27]. The typical CDRAD phantom (air-filled holes) is used in this Chapter with two different techniques: with anti-scatter grid and without grid. Then the obtained images from both techniques were analysed utilising both the IQF and IL theory. This Chapter implies TIL is more sensitive method compared to the IQF method because the IL differs depending on the distribution of detection rate. Also, the TIL calculation allows the evaluation of the variation between two techniques in bits for multiple observers [28]. This proves that TIL is more reliable indicator of the performance of the improvement to IQ that use of the anti-scatter grid. The TIL of non-grid CDRAD phantom was higher by 103 bits compared to the CDRAD phantom using anti-scatter grid results, implying that using the grid in a CR system can significantly improve the efficiency of detecting the low contrast details and reduces the amount of information loss and enhances the imaging quality.

Chapter 8 continues the investigation of the impact of the anti-scatter grid on different digital system: the digital radiography (DR) systems. These systems are based on using detectors that are technically made to reject most of the scatter; this reduces the demand for the grids [27, 29]. Hence, the applicability of grids in this imaging modality has been debatable. Three different flat panel DR systems were investigated: the Agfa digital flat panel system (Agfa DX-D 600), the Philips ProGrade DR retrofitted to existing Philips Optimus 65 X-ray machine and the Shimadzu, RAD speed radiography system with the anti-scatter grid technique applied. The typical CDRAD phantom with air-filled holes was used in this
Chapter. To evaluate the performance of each DR system with the grid technique, an automated CDRAD analyser software was employed to calculate the IQFinv from the CDP data for each system. This Chapter concludes that utilising the anti-scatter grid technique in the DR system is vital for removing scattered radiation. Our results clearly show the importance of anti-scatter grids in DR systems to improve image quality by reducing the level of scattered radiation reaching the image receptors.

Chapter 9 introduces a special design of CDPs for CT scanners called CTCDP. This phantom is made of acrylic plates (Perspex; polymethyl methacrylate) and was designed to test the LCD of CT scanners. The phantom has a circular shape to simulate a large body area, e.g. the chest or abdomen, and contains multiple holes and different diameter sizes (more information about this phantom is discussed at Chapter 9). In this Chapter, the new phantom design to complement the existing Catphan phantom used in CT image quality assessment was tested and was demonstrated to be valuable for extending the contrast ranges to lower limits and increase the number of measuring points. The IQF method was applied to validate this newly designed CTCDP to quantify the LCD of objects using the CT imaging modality. The IQF scores recorded by all participants demonstrated efficiency with regard to the detection of low-subject contrast (small diameter) and high-subject contrast (large diameter).

Chapter 10 continues the investigation of the LCD at our phantom CTCDP. The assessment of the image quality is achieved by employing the information loss (IL) theory to the CTCDP. This Chapter concludes that the application of the information loss factor provides a good indication of phantom performance. It allows the amount of information loss with regard to each diameter size in the CTCDP to be calculated.
Chapter 11 demonstrates the modified form of the general formula of the CTDI. The experiment conducted consists of three parts: 1) A new derivation of the CTDI formula to accommodate the dose enhancement factor was outlined. 2) The contrast enhancement that result from the inclusion of contrast media is determined; then the average of CT numbers was compared among holes’ images to indicate the contrast enhancement in regard to the Perspex phantom material. 3) Gafchromic films are applied to measure the dose enhancement due to the inclusion of the contrast media. This Chapter concludes that the new modified formula of CTDI can be utilised in calculating the estimated dose delivered to patients using a CT scanner. Its measurements with the Gafchromic films show that some level of dose enhancement occurs when introducing the contrast media.

Chapter 12 summarises the conclusion and the findings of the thesis for all Chapters.
Chapters 2 and 3 review the backgrounds for two X-ray-based modalities: conventional radiography [including computed radiography (CR) and digital radiography (DR) systems] and computed tomography (CT).
Chapter 2 Low-Contrast Detectability in Digital Radiography

In most healthcare institutions, the digital technologies of computed radiography (CR) and digital radiography (DR) are widely used in place of the traditional screen-film (SF) system. The CR system uses storage-phosphor screens (SPSs), such as BaFBr:Eu$^{2+}$, while the DR system uses amorphous selenium (a-Se; direct conversion) and CsI (indirect conversion). The image quality (IQ) of these digital systems is influenced by multiple factors such as contrast, spatial resolution, un-sharpness and artefacts [30].

IQ in DR and CR systems can be assessed by using two different methods: objective methods, such as detective quantum efficiency (DQE), information entropy, the Rose model method and pixel signal to noise ratio (pixel SNR) and subjective methods, such as visual grading analysis (VGA) and receiver operating characteristic (ROC) analysis, which rely on the human perception like the radiologist or the radiographer [17]. During the diagnosis process, it is vital to detect small lesions with low contrast. Contrast Detail Phantoms (CDPs) are utilised in the evaluation process of low contrast detail (LCD) detectability as part of both the subjective and objective methods [3], which will be explained in (Chapter 2, Section 2.8.6).

The aim of this Chapter is to review the X-ray production and how it interacts with matter prior to generating diagnostic images. Also, it will describe both CR and DR systems and discuss the factors that affect IQ on these systems. The assessment methods for evaluating IQ
will be discussed including the subjective and objective methods. Finally, the importance of the LCD detectability and its role in assessing the IQ by using the CDPs will be examined.

2.1 X-ray production

When the electrons are accelerated between the cathode and the anode in the X-ray tube, they acquire kinetic energy. This energy transfers to the target atoms due to the interaction process that occurs when the electrons hit the target. In this interaction, penetration into the target occurs at a very small depth of approximately 0.25 to 0.5 mm. As a result, two types of energy are generated: 1) heat, which forms when the kinetic energy is converted to thermal energy, and 2) X-rays, which are formed when the kinetic energy is converted to electromagnetic energy [2, 31, 32].

An atom consists of a nucleus that is surrounded by orbiting electrons. The nucleus has a positive charge and contains neutrons and protons. Electrons around the nucleus occupy energy levels called shells; these shells can have specific numbers of electrons. For example, the maximum number of electrons that can exist in the K-shell (the inner shell) is two. The next L-shell can have 8 electrons, while the M-shell and N-shell can contain 18 and 32 electrons, respectively. Incoming electrons (e.g., from the cathode) can cause excitation when they interact with any of these orbiting electrons. As a result of these interactions, the orbiting electrons can rise to higher orbits. In the case of an ionization process, an incoming electron completely removes the orbiting electron from the atom [2, 31, 32].
2.1.1 Characteristic X-ray

When ionization occurs to an electron, it ejects the electron from the atom, causing a vacant space. The hole that results from the ejected electron at the inner shell (e.g., the K-shell) is rapidly filled by electrons from outer shells (e.g., the M-shell). When this occurs, an X-ray photon is generated due to the excess energy of the electron that moves from the outer shell to the inner shell. The created energy of X-ray is equal to the difference between the binding energies (BEs) of the electrons contributed in the process. In Figure 2.1a, a projectile electron with high energy ionizes the atom, generating a hole in the K-shell due to target electron ejection. This hole is occupied by an electron from the M-shell producing a characteristic X-ray photon with energy of 66.7 keV. If the interaction occurs in a tungsten atom, the X-ray energy photon produced is calculated according to the binding energy. For example, the BE of the K-shell is 69.5 keV minus the BE of the M-shell (2.8 keV), which is equal to 66.7 keV. This calculation can be applied to all different shells, such as the M-, N-, O- and P-shells, when their electrons fill the K-shell. When this occurs, the X-rays produced are called K X-rays, since the X-rays result from K-shell ionisation. Ionization can also occur to M-shell electrons, which can be replaced by outer electrons. Because this emission is characterised by a target element, it is known as characteristic radiation [2, 31, 32].
2.1.2 Bremsstrahlung

The nucleus of the target atom affects the projectile electron through an electrostatic field when this electron gets close to the target atom’s nucleus. The electron changes its path and slows down as it passes close to the nucleus; hence, it travels in a different direction with decreased kinetic energy. This kinetic energy loss is represented in X-ray form Figure 2.2. This X-ray photon’s production is called Bremsstrahlung, which originates from German terms brems, meaning braking, and strahlung, meaning radiation. The X-ray emission from Bremsstrahlung varies in energy due to the loss of some or all of the electrons’ kinetic energy. As a result, an electron beam would have an energy extending from zero to maximum energy. The low energy of Bremsstrahlung occurs when the nucleus slightly influences the
incoming electron, which generates a low-energy X-ray photon and then resume with reduced energy. When the incoming electron loses all its energy or stops, this results in high energy production for Bremsstrahlung X-rays. The intensity (I) produced by Bremsstrahlung radiation is proportional to the electron energy (E) and the atomic number of the target (Z) as follows:

\[ I \propto ZE^2 \]

This explains the importance of heavy target atoms (e.g., tungsten), since the production of Bremsstrahlung is directly related to the atomic number of the material [2, 31, 32].

Figure 2.2: Bremsstrahlung is generated when a projectile electron changes its path through the atomic nucleus [31].

### 2.2 X-ray interaction with matter

When ionizing radiation such as X-rays interacts with matter, approximately 8 interaction processes occur. Of these, two fall within the range of X-rays: photoelectric absorption, Compton scattering.
2.2.1 Photoelectric absorption

The photoelectric effect occurs at the X-ray energies utilised in diagnostic imaging. The range of this energy is about 50 to 150 kV, which is equal to or larger than the binding energy of the inner orbital electron, which is considered a requirement for a photoelectric absorption. This phenomenon occurs when the full energy of an incident photon is transferred to the inner orbital electron due to its interaction with the inner shell electron. The photon disappears after this interaction, since it is completely absorbed. The binding energy of the orbiting electron is overcome by the energy transferred from the incident photon, which causes the electron to be released from the atom. This released electron is emitted in a wide range of angles and it is called a photoelectron. Because of the incident photon attempts to conserve energy and momentum in the interaction, it needs high incident photon energy for small angles \([2, 31, 32]\). The remaining energy is transformed to the kinetic photoelectron energy, which permits it to emit through the attenuating material (Figure 2.3). The atoms of the attenuating material are dissipated by the kinetic energy of the photoelectrons until they achieve resting status. The vacancy in the inner electron shell of the atom created by the photoelectron ejection causes unstable atoms and in these circumstances electron transitions from outer shells occur. As a result, a radiation photon is generated as a characteristic X-ray related to the difference in binding energy of the two electrons shells that contribute to the transition process; hence, this characteristic X-ray is absorbed by the medium. The probability of photoelectric occurrence is dependent upon the photon energy e.g. the photon with energy less than 15 keV does not have enough energy to eject an L electron \([2, 31, 32]\).
2.2.2 Compton scattering

In Compton scattering, a free electron and incident photon collide, resulting in absorption processes for both. Energy is transferred from the X-ray beam to the atom of the attenuating medium, and scattering occurs due to the change in the incident photon path. In the attenuating material, the binding energy of the orbital electron becomes more vulnerable due to the increase in incident photon energy; hence, free electrons are released in a process that can be illustrated as follows: an incident photon collides with a free electron, and the photon transfers some of its kinetic energy to the electron. The electron is released and travels through the attenuating material in either a side direction or a forward direction (Figure 2.4). Until the electron settles, it dissipates its kinetic energy through several electron particle interactions. The incident photon still exists, but with reduced energy, since it transferred some of its kinetic energy to the electron. Moreover, its path is deflected from its original path. Compared to the incident photon, the scattered photon has a greater wavelength and a decreased frequency [2, 31, 32].
In X-ray diagnostic radiology, images are produced by the interaction of X-ray photons which have been transmitted through the patient with the imaging plate or the detectors in CR and DR systems. Recently, most medical institutions have replaced SF systems with technology such as Computed Radiography (CR) and digital radiography (DR). The following sections will describe these two technologies in more detail.

2.3 Digital radiography and computed radiography systems

The CR system creates images through an indirect conversion process that utilises storage phosphor plates associated with the individual image readout process. This is applied in two steps: First, the X-rays are acquired through a storage phosphor screen (SPS), such as BaFBr:Eu2+. Second, light emitted from the SPS is acquired by photo detectors, which then transform the luminescence into digital images [30].
The DR system transforms X-rays into electrical charges, utilising Thin Film Transistor (TFT) arrays in order to apply a direct readout process. This system can use either direct or indirect processes to transform X-rays into electric charges. An X-ray photo conductor such as Amorphous Selenium (a-Se), is available in direct conversion detectors, which transform X-ray photons into electric charges in one stage [33]. In contrast, an indirect conversion system uses two stages. The first stage provides a scintillator like CsI, which transforms X-rays into visible light. The second stage provides the amorphous silicon photodiode array, which transforms light into electric charges (Table 2.1) [30].

<table>
<thead>
<tr>
<th>Detector technology</th>
<th>Capture element</th>
<th>Coupling element</th>
<th>Charge readout</th>
</tr>
</thead>
<tbody>
<tr>
<td>Computed radiography</td>
<td>BaFBr:Eu²⁺</td>
<td>Photostimulated luminescence</td>
<td>Photomultiplier tube; signal digitization</td>
</tr>
<tr>
<td>(CR)</td>
<td></td>
<td>(PSL) light-guide</td>
<td></td>
</tr>
<tr>
<td>Direct radiography</td>
<td>a-Se</td>
<td>None</td>
<td>TFT array</td>
</tr>
<tr>
<td>(DR)</td>
<td></td>
<td>Contact layer</td>
<td></td>
</tr>
<tr>
<td>Indirect conversion</td>
<td>CsI or Gd₂O₂S</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 2.1: Three components of digital detectors [30].

2.4 Computed radiography (CR)

Computed radiography has historically been the main digital technology available for projection radiography [34]. It utilises a photostimulable detector instead of a traditional screen film cassette [34]. The X-ray photons receive the storage phosphor plate inside the cassettes. This technology permits the radiographer to acquire a plain radiographic image similar to that acquired by the traditional screen film system. The major difference lies in the creation of the latent image and the image processing [35]. There are three steps in CR imaging: exposure, readout and erasure. The radiography cassette has an image plate (IP) or SPS, which contains a detective layer of photostimulable crystal. This detective layer is
comprised of the phosphor family BaFX:Eu, where X can be a halogen, such as Br, Cl or I. The latent image can be stored in a typical SPS for a specific period of time [30, 36]. The phosphor crystal is deposited into plates in solid material in the selected scintillator. When the SPS is exposed to X-ray photons, the radiation energy excites electrons, causing them to move to high energy levels (Figure 2.5 a & b) [30].

![Figure 2.5: SPS exposure and the PSL process](image)

The excited electrons present an unstable level of energy in the atoms. The crystal structure of the phosphor stores the absorbed X-ray energy. The latent image generated by these high energy levels provides a spatial distribution of the electrons at the storage phosphor detectors. Additional light energy with appropriate wavelengths can stimulate and, hence, release the tapped energy; this process is called photostimulated luminescence (Figure 2.5 c) [30]. Following the X-ray exposure and the formation of a latent image, a CR reader device is used to scan the SPS [37]. In this readout process, the photostimulable screen is scanned by a red laser beam, which stimulates the emission of blue light photons through excitation (Figure
2.6). Scanning each pixel of the detective layer of the image plate with a high-energy laser beam of a particular wavelength will cause a release of stored energy, since the emitted light has a different wavelength than the laser beam. This stimulates the Photostimulable Luminescence Process (PSL), causing blue light emission proportionate to the original X-ray photons and releasing the excited electrons to a lower energy level. The photodiode collects the blue light and transforms it to an electrical charge. The digital images are generated by converting the electrical charges to images using an analogue-to-digital device [30]. The last step in the CR imaging process involves erasing the residual signals. After readout, the latent image has residual electrons which are trapped at higher energy levels. These are erased by a high-intensity white light, and new electrons are not introduced from the ground energy level [36].

Figure 2.6: The process indicates CR storage phosphor plate exposure and readout [38].
2.5 Digital radiography (DR)

Digital radiography uses a large-area X-ray detector (flat panel system) that has a layer sensitive to X-rays and an electronically readable system based on TFT arrays. There are two types of TFT detectors: indirect conversion TFT, which uses a light-sensitive TFT photodiode and a scintillator layer, and direct conversion TFT, which utilises a TFT charge collector and an X-ray-sensitive photoconductor layer [30, 34].

2.5.1 Direct conversion system

This system uses a-Se as a semiconductor material because a-Se offers high spatial resolution and characteristic X-ray absorption [34]. The electric field is applied across the selenium layer before the flat panel is exposed to X-rays. The electrons are produced by X-ray exposure, and they interact within the a-Se layer. Due to the presence of the electric field, the absorbed X-ray photons are transformed into electric charges and then drawn directly to the charge-collecting electrode [30]. The total collected charge is proportional to x-ray beam intensity, which are produced and then moved vertically to the selenium layer with low lateral diffusion. The charges are stored at the TFT charge collector until the readout (Figure 2.7). The collected charged is quantified and amplified to a digital code value representing the corresponding pixels [30].
2.5.2 Indirect conversion system

This system uses gadolinium oxysulphide, or CsI, as an X-ray detector. The phosphor and scintillation used in the indirect conversion system can be either structured or unstructured (Figure 2.8). The phosphor material in the structured scintillator is a needle-like structure that is perpendicular to the screen surface. This decreases the lateral scattering of light photons and increases the number X-ray photon interactions [30]. The spatial resolution in unstructured scintillators is reduced due to high levels of light scatter. The beam is absorbed when the scintillation layer is exposed to X-ray photons, which it converts to fluorescent light. Through the use of an a-Si photodiode array, this light is converted to an electric charge. In indirect conversion detectors, the scintillators and the a-Si photodiode circuitry are placed on the top layers of the TFT. However, the direct conversion device uses a semiconductor layer. The active area of the detectors is divided into the TFT switch (for the
readout process) and an integrated array of image elements (the pixels), and each element contains photodiodes (Figure 2.9) [30].

![Figure 2.8: An unstructured scintillation (left image) and a structured scintillation (right).](image)

![Figure 2.9: Indirect and direct X-ray detection. In an indirect conversion, a TFT converts X-ray photons into visible light using a scintillator layer. Then, a photodiode converts the visible light into electrical charges, and the TFT arrays can be read out row by row. In direct-conversion flat-panel detectors, amorphous selenium is used to convert X-ray photons directly into electrical charges, which are then stored in capacitors to be read out by TFT arrays [34].](image)
2.6 Comparison of DR with CR systems

In a DR system, the field size should be large enough for all radiographic examinations. This size must have an active area of about 43 x 43 cm that permits both horizontal and vertical orientations without detector rotation [39]. However, the CR system has different cassette sizes, including standard dimensions for typical plain radiography (e.g., 18 x 18, 24 x 24, 30 x 35, 43 x 43); these contain IPs, which are utilised for the appropriate areas to be examined. The size of pixels and spacing (i.e., the distance between pixel centres) control the maximum spatial resolution, and the spatial resolutions of CR and DR systems are affected by pixel size. For example, the system resolution in the CR system ranges from 100 to 200 µm [39], depending on the detector size in the cassettes, while the system range of the DR detectors is between 127 and 200 µm. The readout time is important for improving image quality and workflow efficiency, which are controlled by the technology type of the system. The readout process in the DR system takes about 1.3 seconds, while, in the CR system, the size of the IP determines the readout speed (i.e., a smaller IP takes about 30 to 40 seconds, which is faster than a large IP) [39].

DR and CR are competing technologies that have existed for the past fifteen years [30]. The type of phosphor used in both systems is vital and affects their performance. The main advantage of the CR system is portability and flexibility, particularly in the operation room (OR), the neonatal imaging area and trauma area. In contrast, the DR system provides great productivity. Compared to the CR system, the DR system provides greater performance and has a conversion efficiency of 20 to 35 percent. Moreover, its conversion efficiency is 25 percent better than the screen film system for chest radiography. Though the DR system has historically been more expensive than the CR system, this has recently changed, and the cost
of DR is decreasing. Moreover, the CR system has become more compact and less expensive, allowing it to reach the low-end market. For these reasons, it is anticipated that these two imaging technologies will exist for long time [30].

2.7 Image quality parameters

Image quality (IQ) describes how well a diagnostic image displays information regarding the physiology and anatomy of a patient, including any changes to the anatomy structures caused by trauma or diseases [40]. IQ is affected by five major image characteristics of: contrast, noise, spatial resolution [41], un-sharpness (blurring) and artefacts [42].

2.7.1 Contrast

The contrast resolution can be described as the ability to transform subtle density differences in a patient’s tissue into image information. It is a measure of the scale of the calculated signal variations between physically different regions of an imaged object (Figure 2.10) [40], and it is produced by a varied attenuation of X-radiation via tissues like glandular and adipose tissues in mammography. The X-ray spectrum affects image contrast, which is controlled by the anode material, the tube voltage applied and the X-radiation filtration. Through the image receptor, the radiation contrast is transformed into variances in optical density in the radiograph (image contrast) or into variances in pixel values for digital imaging (histogram) and post-processing on the monitor [4].

The image contrast is the product of signal contrast and detector contrast, as follows:

\[ C_i = C_s \times C_d \]  \hspace{1cm} (2.1)
The signal contrast ($C_s$) relies on the source of the energy and the physical properties of the imaged object. It defines the energy ranges that are emitted by an object. The detector contrast ($C_D$) relies on the method of the signal emitted by an imaged object that has been modified (e.g., with a scatter submission grid), detected and recorded. Both contrast signals and contrast detectors are vital to generate image contrast. The contrast adjustment is used in digital imaging to reduce or enhance image intensity variations in order to make them observable to human eyes [40].

In the digital image, the image contrast is influenced by the width settings and the window levels. This great contrast results in stronger bright and dark image areas. The different linear attenuation coefficient for X-rays allows one to distinguish between the varied internal structures of the human body, such as its soft tissues, lungs and bones [4].
Figure 2.10: Different contrasts and resolutions of an elbow. a) The elbow image shows good contrast and resolution, with a low level of noise. b) The image displays low noise and high spatial resolution, but it is useless because it has almost zero contrast. c) The image shows extremely poor spatial resolution, as well as high contrast and low noise. d) This image has great spatial resolution; however, the greater noise level has destroyed the contrast information [40].

2.7.2 Noise

Noise overlaps image information and can be defined inside homogenous tissue regions via fluctuations in the brightness of viewing stations or in the optical density of the radiograph [4]. The X-ray image is produced by individually absorbed X-ray photons inside the region of the radiological image. The overall image is formed through the contributions of each single X-ray photon. Therefore, several X-ray photons are absorbed for each image area, resulting in lower fluctuation due to noise. The quantum noise is determined by the X-ray quanta absorbed by the image receptor. The efficiency of the transformation (of the absorbed energy) in the image receptor for different information carriers (visible light) or charge carriers (electrons), as well as the number of X-ray photons, can also be involved in the noise. This is due to the limitations related to the number of light photons that form the visible image in the viewing station [4]. Random noise occurs due to statistical fluctuations inside the imaging system (e.g., the digital system, including quantisation noise, electrical noise) or the SF system (e.g., graininess noise), and these have a major impact on the total image noise. This type of noise is not linked to a specific location at the receptor. In contrast, fixed pattern noise is correlated with fixed locations on the receptor, and it includes spatial differences in screen thicknesses in the SF system. In digital radiography, this type of noise is related to position-dependent light collection efficiency at the CR plate reader, while, in the DR system, it is related to the variations in preamplifier gains. The digital system has the ability to eliminate this noise by utilising digital post-processing techniques [4].
Another type of noise is quantization noise. This type of noise occurs when the digitization process of the analogue detector output voltage produces discrete pixel values. To quantize the image data into grey-scale values, the digital imaging system uses an analogue digital converter (ADC)[8]. These grey-scale values are controlled by electronic binary number channels (on/off) inside the ADC (named bits). The maximum number of grey-scale values that can be encoded is equal to $2^n$, where $n =$ the bit number. The quantization noise increases when signal encoding errors occur due to an insufficient number of quantization steps. Minimal quantization noise can be achieved when the digital detectors for projection radiography use 10 to 14 bits (1024 to 16384 analogue-to-digital units) in their output images [8].

Scattered radiation is considered a noise source that causes image degradation in radiography. It decreases the dynamic range of available X-rays on the exit side of the patient. As a result, the contrast subject and the signal-to-noise ratio (SNR) are decreased because they have no enough signal and contain more quantum noise [8]. The scanned slots of the DR detectors have scatter rejection capabilities and, hence, do not need a grid to eliminate the scattered radiation. However, anti-scatter grids must be used in the area detectors of SF, DR and CR radiography when the scattered radiation is dominant, [8] which it is when patient thickness $\geq 10$ cm [43]. It is vital to use grids in the CR system due to great scatter sensitivity of Barium halide ($k$-edge = 35 KeV). In digital detectors, the grid decreases the noise (i.e., eliminates scattered radiation) and the signal (i.e., an incomplete transmission of primary radiation). This depends on the grid design and the scatter-to-primary ratio in the beam i.e the ratio is higher for larger patients [8].
2.7.3 Spatial resolution

Spatial resolution is the ability of an imaging system to permit two adjacent structures to be demonstrated as separate. It is known as the sharpness of the image and relates to the distinctness of the image’s edges. Image blurring results in spatial resolution loss [8]. This is affected by several factors, such as the patient’s motion in relation to the image receptor and the X-ray source, detector element (del) effective aperture size and geometric factors, like the size of the focal spot of the X-ray tube [8].

In a CR system, the laser light beam scattering that occurs during the image plate readout causes a loss in spatial resolution. As a result, there is a de-excitation of locations in the phosphor, which is somewhat larger than the laser beam size and, hence, is larger than the separation of laser positions. In this case, blurring extends beyond the size of the pixels [38]. The CR system allows the plate to be read from both sides, since the thicker phosphor causes more blurring and greater scattering. Moreover, the use of structured crystalline geometry permits greater thickness and an enhanced detection of efficiency, without spatial resolution loss [38].

In a DR system, the spatial resolution is influenced by two factors. The first factor is related to the indirect system that affects the spread of the light photon in the process of converting X-rays to light. To overcome this problem, many manufacturers use structured converters (Cesium iodide), which are created in narrow, parallel columnar structures [8]. Thus, incident X-ray photons occur along the long dimension of these columns. This method retains spatial resolution and improves absorption efficiency through thicker absorbers or longer columns. However, direct conversion in the DR system is not affected by this issue, since the electron
spread inside the photoconductor material is minimal on account of the electrons being accelerated toward the TFT [8]. The second factor that has an impact on the spatial resolution in the DR system is related to the size of the del. The single quantity or the charge read from del contributes to all X-rays absorbed inside an individual del during radiation exposure. Thus, a partial volume effect or decreased contrast is produced due to the smearing out of a patient’s structures when these structures are smaller than the del size [8].

### 2.7.4 Un-sharpness

Un-sharpness is a blurring of the well-defined boundaries or edges of a subject. Un-sharpness can be classified using four components: subject un-sharpness, receptor un-sharpness, motion un-sharpness and geometric un-sharpness [42].

The boundaries of anatomical structures are not always well-defined edges. An object’s shape has an impact on the projection of sharp edges onto the image receptor (Figure 2.1). In this Figure, the left image illustrates sharp edges, while the right image indicates blurred edges. This image un-sharpness is named subject un-sharpness, and it can be caused by either the object shape or the object composition [42].
Figure 2.11: The left image represents free subject un-sharpness because the boundaries of the trapezoid are parallel to the X-ray’s path. The right image indicates subject un-sharpness resulting from different edge densities [42].

The image receptors in every display technique add image un-sharpness called receptor un-sharpness. For instance, the display device in a digital radiography system provides different levels of un-sharpness (Figure 2.12). In this Figure, three display formats are presented for the same data. Fine matrix data are presented in the left image (0.2 x 0.2 mm pixels), and the image is continuous and smooth. The right image displays the same data for a coarser matrix (0.6 mm x 0.6 mm pixels), which adds significant un-sharpness to the image [42].
Figure 2.12: This digital chest radiograph indicates multiple pulmonary nodules at three levels of spatial resolution. The decreased receptor un-sharpness is characterised by a finer display (the left image). Left image: 0.2 x 0.2 mm pixels. Centre image: 0.4 x 0.4 mm pixels. Right image: 0.6 x 0.6 mm pixels [42].

Motion has the most significant impact on radiologic images causing un-sharpness. The boundaries in a patient are projected on the image receptor in varied areas due to the motion created by producing the image [42]. As a result, the boundaries expand beyond a finite distance, forming a blurred image. To overcome this issue, a short examination time is used as a main rule in radiography, since both involuntary and voluntary motions are present to some degree when imaging any anatomic regions [42].

The image forming process is responsible for geometric un-sharpness. It is influenced by the distance between the source and the object or patient, by the distance between the object and the image receptor and by the size of the radiation source [42]. For example, the boundaries in the object are blurred over a finite region of the image when a small-sized radiation source is used. Moreover, the blurring level increases when the distance between the image receptor
and the object increases. To avoid geometric un-sharpness, the distance between the object and the detectors should be decreased as much as possible, and the object should be moved farther from the radiation source [42].

### 2.7.5 Artefacts

Artefacts are parts of the image that can mimic clinical features, obscure abnormalities or impair image quality [44].

Exposure artefacts are one of many artefact types that occur in DR systems. Despite that DR systems have a wide dynamic range and linear response digital image receptors that can produce good image quality, overexposure or underexposure can affect the image quality of this system (Figure 2.13) [44]. As a result, the radiograph appears much sharper when the display size is smaller, which decreases the uncertainty of adjacent pixels. Ghost artefacts are very common in DR systems [45]. The scintillation layer emits light, which is then digitized by the photodiodes; finally, a radiograph is produced. The simulated photodiodes trap charges; however, the released charges can persist beyond the readout causing the artefact. Applying subsequent exposures in short period of time will eliminate the artefact’s appearance [44].
In a CR system, a double exposure artefact can be produced when using the same cassette for two exposures without erasing the cassette in between. As in the DR system, ghost artefacts occur in the CR system when effective saturation of the image receptor image occurs. To avoid such artefacts, a correct erasure setting must be employed (since the radiation can be trapped within the image plate for several minutes) [44]. Underexposure artefacts also occur in the CR system, making the generated images appear grainy. Proper exposure factors must be considered, particularly in the CR system, to enhance the image quality. More pattern artefacts are produced due to the selection of low grid line rates (e.g., 33 lines/cm) [46], which are oriented with grid lines that are parallel to the plate reader scan lines. The use of grids with 60 lines/cm or more can reject these artefacts from the CR system [46].

2.7.6 Image quality and radiation dose

It is vital to minimize the radiation dose to the patient, as well as to enhance image quality. The concept of ‘as low as reasonably achievable’ (ALARA) is associated with image quality,
and this concept concentrates on the radiation doses that are delivered as a result of medical imaging procedures. These are important because of the need to maintain low radiation doses. In some cases, such as radiography applied to children, the patients have high radiosensitivity to ionization radiation [47].

DR and CR systems provide significant patient dose reductions to patients, as well as better IQ than screen-film radiography [48]. Optimising IQ and lowering the dose requires the optimisation of the whole imaging chain through, for example, acquisition techniques and detector efficiency [48, 49].

Acquisition parameters, such as kVp and mA, control the amount of the radiation dose and, hence, the SNR and contrast-to-noise ratio (CNR) [50]. When altering the kVp, the penetrating power of the beam or beam quality is changed. Moreover, the mA is responsible for the beam quantity or for the number of photons in the primary beam. If an increased X-ray quality is required for more penetrating primary beams, a higher kVp must be applied to allow the electrons move faster in the tube current. This will decrease the scatter radiation inside the body and reduce the dose, due to the great penetration associated with a high kVp. It is necessary to maintain a dose level that is as low as possible and that has a high kVp, without affecting IQ [51]. Moreover, in digital radiography, the dose mainly affects the noise in the images and, hence, degrades IQ. For example, the image density is influenced by underexposure, resulting in increased image noise [50]. The photons are distributed in a random manner inside the X-ray image, which is considered an important source of quantum noise. In digital radiography, the pixel values that are correlated with individual detector elements vary around their anticipated values. The square root of the exposure level is
proportional to the noise in the detector element [50]. Therefore, higher exposure levels lead to a better IQ and an improved SNR. Therefore, if the SNR needs to be improved by a factor of two, the dose must be increased by a factor of four (the quantum noise is assumed to be predominant noise source) [50]. This improvement in IQ must be evaluated against the increase in the dose absorbed by the patient [50].

Another aspect of digital radiography is that the differences in digital detectors’ radiography approaches lead to differences in IQ and dosages. In CR detector techniques, the dual-sided read CR technique allows for great efficiency in relation to light collection through the reading of both sides of the screen, which results in an improved SNR. Moreover, the X-ray absorption efficiency is improved by the use of a needle-like phosphor (thicker phosphor) without any degradation to the spatial resolution [50].

DR system detectors use lower dose levels with respect to IQ than single-sided read CR systems. Moreover, the CsI-flat panel detector (CsI-FPD) provides better performance than the Se-flat panel detector (Se-FPD) with regard to the study of image quality as a function of spatial frequency. This is because the CsI has a high atomic number and density, which leads to the improved capture of latent X-ray images [50]. However, Se-based systems demonstrate less blurring of image signals at high spatial frequencies. For indirect conversion detectors, the decrease of light spreading in the scintillator at the needle structure of CsI allows for the application of thicker layers with greater efficiency. This enhances the image quality above what is possible with unstructured scintillators, such as regular CR systems and gadolinium oxysulphides (GOS) (Figure 2.14) [50].
2.8 Different methods for image quality evaluation

Several methods can be used to evaluate the digital radiography system, including CR and DR detector performance and image quality, as follows:

2.8.1 Detective quantum efficiency (DQE)

Detective quantum efficiency (DQE) is expressed as a function of object detail, and it is used to measure the combined effect of the contrast and the noise performance. DQE measurements depend on a combination of the functions of the modulation transfer function (MTF) and the noise power spectrum (NPS). The MTF is an imaging system’s ability to extract the contrast of an object as a function of object detail [50]. The DQE is used to describe the efficiency of a radiographic system in translating incident X-ray photons into
valuable signals relative to the noise inside an image [52]. Mainly, the DQE is applied when measurements of detectors’ physical performance are required [30].

The MTF is represented to be equal to the absolute value of the Fourier transformation of the line-spread function (LSF). This is obtained by reproducing a narrow slit (approximately 10 μm) or a sharp metal edge following derivation (Figure 2.15). When the MTF is measured directly from LSF data in digital radiography, the aliasing artefact will be formed. To overcome this phenomenon, a calculation of the MTF of analogue components and sampling apertures prior to digitization can be performed to evaluate the inherent resolution properties [30].

The NPS can be measured as the Fourier transformation of the autocorrelation function of the synthesised slit in the noise image. It is preferable to calculate the NPS directly from the Fourier transformation of two-dimensional noise images. In this case, the NPS is the second power of the absolute value of the Fourier transformation (Figure 2.15) [30]. The linearization of all images and the flat field images at different kinetic energies released in matter, or kinetic energy released in matter KERMA ($K_{air}$), at the entrance surface of the image detectors should be considered in calculating the performance of the digital image detectors. In every image, the central pixels are used (e.g., $1024^2$); then, this area is divided into a number of squares of smaller size (e.g., 64). At the end of this process, the NPS is calculated as the average of the NPSs of each of these small areas; hence, the NPS can be determined with greater precision [53].
Figure 2.15: The formation of detective quantum efficiency (DQE). In the DQE equation, $G$ is the gain factor, $X$ is the exposure at the detector associated with NPS measurements (µGy), $MTF(u)$ is the pre-sampling MTF, $q$ is the ideal $SNR^2$ (the number of incident X-ray quanta per unit area per µGy and $NPS(u)$ is the NPS of the output image [30].

The higher the DQE value, the better the SNR characteristics of the detectors is. A comparison of the DR and CR systems using the method recommended by the International Electrotechnical Commission (IEC) indicates the differences in DQE responses when using the same exposure conditions [30]. When applying an exposure level, which correlates to 4 µGy in the DR system, there is a wide range in performance with regard to resolution and noise components [30]. The DQE peak of 60 percent is obtained at spatial frequencies of 0 to 0.5 cycles/mm and normally decreases at lower spatial frequencies [30]. However, CR system results are affected by MTF and NPS measurements. Peak curves are achieved at 0.5
to 1 cycles/mm, and the peaks are 85 and 66 percent for the horizontal and vertical directions, respectively [30].

The DQE is a very valuable physical characteristic tool for digital radiography. However, it does not consider certain aspects, such as focal spot effects, magnification and anti-scatter grids, on the image quality obtained from clinical practice and scattered radiation. This consideration is important in computed tomography [4].

2.8.2 Information loss based on entropy

The concept of information entropy defines the amount of uncertainty or randomness in an image or signal and can describe the amount of information provided by a signal or image [54]. It is considered a quantitative measure of the information transmitted by the image. IQ can be evaluated when the transmitted information (TI) acquired by the image is known. According to the physical measurements perspective, the greater the information, the better the IQ will be [54].

The entropy method for evaluating the quality of radiographic images was first introduced by Uchida and Tsai in 1970 [54]. They used this method to evaluate the quality of tank-developed images for automatic processor-developed images. They illustrated that IQ is largely influenced by exposure factors, the film development process, the X-ray apparatus and intensifying screen film systems [55].
TI is a concept derived from information theory. When events like $S_1, \ldots, S_n$ occur with probabilities $p(S_1), \ldots, p(S_n)$, then the uncertainty average correlated with each event is defined by the Shannon entropy theorem, as follows [54]:

$$H(S) = - \sum_{i=1}^{n} p(S_i) \log_2 p(S_i) \quad (2.2)$$

If we consider two random variables (e.g., $x$ and $y$), which correspond to input and output variables, the entropies for the input and the output are represented as $H(x)$ and $H(y)$, respectively. These can be defined as follows [54]:

$$H(x,y) = H(x) + H_x(y) = H(y) + H_y(x) \quad (2.3)$$

Where $H_x(x)$ and $H_y(y)$ are conditional entropies representing the entropy of the output when the input is known and the entropy of the input when the output is known, respectively. Therefore, the TI, or $T(x;y)$ can be computed as follows [54]:

$$T(x;y) = H(x) - H_y(x) = H(y) - H_x(y) = H(x) + H(y) - H(x,y) \quad (2.4)$$
The relationships among entropies can be explained through a Venn diagram (Figure 2.16), which considers an experiment in which every input has a unique output belonging to one of several different output groups [54].

![Venn diagram](image)

Figure 2.16: Venn sketch depicting the relationship between input entropy $H(x)$ and output entropy $H(y)$. The transmitted pieces of information $T(x;y)$ and $H(x,y)$ are joined by the conditional entropies $H_x(y)$ and $H_y(x)$. The amount of information transmitted for each type of entropy is indicated by the respective area of the diagram [54].

Despite their usefulness, both TI and the information entropy method have some disadvantages. The TI values do not represent information about frequencies (e.g., MTF and NPS)[54]. Moreover, some noise contribution, such as structural and electronic noise, cannot be represented individually [54].
2.8.3 Receiver Operating Characteristic (ROC) analysis: Observer performance method

The receiver operating characteristic (ROC) analysis is a method for evaluating the quality or performance of a medical image and determining whether it is related to a normal anatomy or a pathological structures by incorporating human observers [17]. It consists of: 1) sensitivity, which is the number of actual positive cases or the number of true positive decisions, and 2) specificity, which is the number of actual negative cases or the number of true negative decisions [56]. The ROC is illustrated as a plot of test sensitivity with two coordinates (x,y). The y coordinate normally represents sensitivity, and the x coordinate represents the false positive rate (FPR) or a 1- specificity. Every discrete point (operating point) on the graph is generated through the utilization of varied cut-off levels for a positive test result. After that, the ROC curve can be created by connecting all points acquired at all possible cut-off levels. The resulting curve is called the empirical ROC curve, and it indicates the relationship between sensitivity and the FPR. This method can be used to evaluate the performance of a test independently of the decision threshold due to its ability to display all possible cut-off levels between the sensitivity and the FPR [57].

The area under the curve (AUC) can be used to measure the overall performance of the diagnostic test because it illustrates the average value of sensitivity for all possible values of specificity [17, 58, 59]. The AUC can represent values between 0 and 1, and higher values (close to 1) suggest a better overall performance of the diagnostic test [57]. The most successful way to complete the ROC study when comparing different modalities is to allow a number of readers to interpret the same cases for all modalities. However, this method has the disadvantage of being strongly reliant on the occurrence of a signal or disease. Moreover, the
performance of the observer (e.g., a radiologist) may differ when applying this method in clinical and experiment environments. Moreover, this approach requires a large number of cases to generate statistically significant results, which suggests low reliability [17].

2.8.4 Visual grading analysis (VGA): Observer performance method

The VGA technique is used to assess the image quality by grading the clarity of reproduction of pathological or anatomical structures [60, 61]. There are two main ways to perform a visual grading analysis (VGA): 1) a relative grading, which uses one or multiple images as a reference, and 2) absolute grading, which uses no reference. In the relative grading method, a comparison is done of the display quality of the target structures (of the test image) and the corresponding reference image landmark. This method uses a scale with different points (e.g. 3, 5 or 7) to classify the observer decision. For instance, a five-step scale comparing visibility can be demonstrated as: zero = equal to the reference image, -2 = much worse, -1 = slightly worse, +1= slightly better, +2 = much better. In the absolute grading method, the observer states his opinion regarding the appearance of a specific feature without using a reference image. This method uses an absolute scale of 4 to 7 points (table 2.2) [61].
<table>
<thead>
<tr>
<th>Absolute rating</th>
<th>Meaning</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Excellent image quality (no limitations for clinical use)</td>
</tr>
<tr>
<td>2</td>
<td>Good image quality (minimal limitations for clinical use)</td>
</tr>
<tr>
<td>3</td>
<td>Sufficient image quality (no substantial loss of information / moderate limitations for clinical use)</td>
</tr>
<tr>
<td>4</td>
<td>Restricted image quality (clear loss of information / relevant limitations for clinical use)</td>
</tr>
<tr>
<td>5</td>
<td>Poor image quality (loss of information / image must be repeated / image not usable)</td>
</tr>
</tbody>
</table>

Table 2.2: The absolute rating used to grade clinical images using VGA analysis [61].

The visual grading characteristic (VGC) analysis can be used to analyse a VGA data study [62]. In this method, the observer performs several scale steps to state his opinion concerning the IQ. In the first step, the two modality results are summarised separately in a table that includes a 2 x n frequency table, where n is the category numbers. In the second step, a calculation of the VGC data points is performed in a table. These points demonstrate the coordinate of the VGC curve with an origin of zero. The arrangement of the data points is done according to the cumulative or relative frequencies of the matching categories. The last point is represented by one, and it includes all decisions. The plotting of the VGC curve for calculating the area under the curve $\text{AUC}_{\text{VGC}}$ can be achieved using proper ROC software.
The AUCVGC is considered a tool that can be used to measure the variation in image quality between two modalities or settings [63].

The visual grading method has many advantages. First of all, the validity of such studies can be assumed to be high if the anatomical structures are selected based on their clinical relevance and if the observers are experienced radiologists. Moreover, in some cases, this method agrees with the outcomes of advanced calculations of physical image quality and human observers, which contribute to the detection of pathologies correlated to the reproduction of anatomy. Furthermore, the time consumption is relatively moderate (particularly for the observer), and it is easier to conduct in hospitals than the ROC method. The main disadvantage of this method (with regard to analysing the data) is related to the difficulties in interpreting the curves that cross the diagonal [61].

2.8.5 The Rose model method and the pixel SNR

The Rose model uses the pixel SNR, which is determined using the ratio of the mean signal of the object and the pixel’s standard background deviation [17]. It describes how the human observer detects a flat-topped sharp-edged signal of area in a uniform background containing uncorrelated Poisson noise. This method explains the SNR of an object with a size denoted by one pixel, as follows [17]:

\[
\text{SNR}_{\text{Rose}} = C \sqrt{\frac{A}{n_b}}
\]  

(2.5)
Where the signal contrast (C): \((\Delta n_s) / (n_b)\), where \((\Delta n_s)\) is the signal that contains extra photons per unit area. \(A\): signal of area. \((n_b)\): The count level in the background, which is the expected number of photons per unit area.

The Rose method seeks to establish a threshold value for the \(SNR_{Rose}\) for an object that is to be visualized by a human observer. Generally, the object to be detected requires a threshold value of five [17], and this threshold value corresponds well with the typical threshold value utilised by the human observer. For example, a number of disc-like objects of varied contrasts and sizes are allocated to a uniform background with uncorrelated Poisson noise. However, the pixel SNR does not consider the object size; hence, it has low correlation in relation to the human observer. Moreover, the pixel SNR is often used as a measurement for image quality when it does not meet the Rose model requirements. In addition, the human observer is not typically interested in single pixel values, and the fluctuations that occur from pixel to pixel are rarely noticeable. Therefore, the validity of pixel SNR as a valuable measurement for image quality is low, and it should be eliminated when comparing different image processing techniques or different imaging systems [17].

2.8.6 Low-contrast detail (LCD) detectability

It is a measure of the detection ability of imaging system under low subject contrast. LCD detectability is considered an image property that is related to image noise and contrast. Various measurements can be used to evaluate LCD detectability, such as the method of using a test phantom with low contrast detail [64, 65].
LCD detectability relies on various factors, such as the visual response of the observer, the noise, the spatial resolution, the radiation exposure level, etc. [66]. The most common method used to evaluate LCD detectability is the visual determination of threshold contrast or low-contrast resolution. This method utilises a test object that contains a series of disk-like details of different thickness (i.e., contrast). The observer’s task is to determine the faintest disk that s/he can distinguish in the image. The contrast detail detectability test is considered another variant of the measurement, and the threshold contrast is determined as a function of the detail size. This visual measurement appears simple; however, acquiring reproducible and accurate results could be difficult in practice due to the absence of an actual threshold. When the contrast of the detail is reduced, the observation of the disk can easily change from “clearly seen” to “not seen”, without any clear edge. To complete the measurements, the observer should adopt a standard above which s/he considers the details to be visible. However, it is hard to maintain, define and communicate such a standard in a consistent way. Thus, the large variations among results can be considered significant [64].

The CDRAD is a special phantom that can be used to test the low contrast details, depending on the disks size and contrast, at the threshold level of visibility (Figure 2.17). The most important feature of this phantom is the contrast detail curve that can be obtained by plotting the contrast ($C_i$) vs. the detail ($D_i$) for all rows from the analysis process of the acquired phantom images [more information about the two different approaches of analysing the LCD are discussed in (Chapter 2, Sections 2.8.6.1 and 2.8.6.2)]. [67]. This phantom is made of a poly methylmethacrylate (PMMA) plate containing holes of different depths and diameters, with dimensions of 26.5 cm by 26.5 cm and a thickness of 1 cm. The phantom must be located between additional PMMA layers to simulate different patient thicknesses (e.g., from 5 to 25 cm). The phantom is composed of 225 squares of equal size, and each square contains
one or two holes of identical dimensions. The first one is located at the centre, while the other occupies one of the four corners. The location of the corner hole is chosen in a random manner [4]. The depth of the holes in each row is altered logarithmically in 15 steps from 0.3 to 8.0 mm [18], and the diameter of the holes in every column is changed similarly. Therefore, in the vertical direction, the diameter detail is changed, while, in the horizontal direction, the contrast detail is varied [4].

The threshold contrast of low contrast detail depends on the noise of the imaging system and the detail size [4]. According to Rose, A [68], small details require high contrast, while large details can use lower contrast. This relationship can be expressed as follows:

$$ C \cdot D = k $$  \hspace{1cm} (2.6)

Such that the product of the detail sizes D and the contrast C at the visibility threshold is constant.

The parameter k can be used to quantify the image quality. For example, the lower the k value is, the better the imaging system is, since smaller details and a lower contrast can be seen in the image [4]. To compare the image qualities acquired with different exposure techniques or radiological equipment, an inversed image-quality Figure ($IQ_{\text{inv}}$) from the phantom image can be calculated using the following equation [67]:

$$ IQ_{\text{inv}} = \frac{100}{\sum_{j=1}^{15} c_i \cdot D_{ij}} $$  \hspace{1cm} (2.7)
Where \((D_{i,j})\) represents the threshold \((j)\) diameter in contrast column \((C_i)\). Better low-contrast visibility is represented by higher values of \(IQF_{inv}\).

There are two different approaches to analysing the LCD: human reading or observation (the subjective approach) and automated software analysis (the objective approach) [16].

![Figure 2.17: The main design for the CDRAD phantom [4].](image)

### 2.8.6.1 Analysing the LCD through human observation (subjective approach)

This approach is based on human perceptions and decision criteria. Different observers perform individually to detect the just-visible details or threshold details for each detail diameter available in an image. The final average score promotes quantitative information regarding the detected image quality by combining information on the visibility of small details and low contrast. The human decision criterion is essential and must be included in the
image chain for evaluating the IQ, due to its leading role in the medical diagnosis process [16].

Hendee and Ritenour [42] introduced three stages of visual perception: detection, recognition and interpretation. The detection of visual signals by a radiologist or observers is important because, if visual information is not detected, it cannot be added into the perceptual process to obtain a correct interpretation. Experiments reveal that 20 to 30 percent of the visual clues available in diagnostic images are missed by radiologists [42]. Moreover, inter-observer variations for different signals occurs in 10 to 20 percent of radiologists who investigate the same images [42]. Moreover, 5 to 10 percent of images incite different interpretations from the same radiologists across different readings[42]. These results highlight the tendency of radiologists or observers to miss vital visual information available in images [42].

In the second stage of visual perception, a problem of recognition occurs due to the tendency for the radiologist or the observer to dismiss information. Generally, the observer does not appreciate the importance of the information that appears in images. An experiment in eye tracking that was performed on radiologists reading diagnostic images illustrated that radiologists frequently detect visual clues that are essential to the diagnosis; however, they fail to include these clues in the interpretive process. This failure means that the visual clues are visualised, but not recognised [42].

In the third stage, the radiologist performs an incorrect diagnosis, despite detecting and recognizing the importance of the visual signals. This error in radiologist interpretations is
due to many factors, such as poor judgment, fatigue, inadequate training, inexperience, overlapping structures, etc. [69]. Because of the previous limitations of the subject approach, the application of another method for analysing LCDs (e.g., the objective approach) can enhance the outcomes of an LCD analyser.

2.8.6.2 Automated LCD analyser (objective method)

Automated LCD analysers are used to assess IQs based on calculations of image data (e.g., SNRs), which are not influenced by human perception [16]. Therefore, changes associated with the image are potentially more reproducible and reliable [70]. Mainly, it is the automated software of the CDRAD analyser that provides a statistical method to define whether a particular contrast-detail combination is detected or not. This method utilises the average pixel signal value and the standard deviation of both the image of the contrast-detail combination under evaluation and its background pixels. For example, the program uses the Welch Satterthwaite method (the Student t-test with a Welch correction) to define whether the average signal level in a specific square is higher than the average background level plus an analytical difference of means. Thus, a detail is detected when the difference between two signals is statistically significant at certain significance level [18].

Various studies that can benefit from the use of the observer method (compared to the subjective method). These include IQ assessment studies aimed at detecting drifts in equipment performance, such as routine quality control (QC) studies, and studies that contribute to improving technical parameters (e.g., kVp) that may affect image quality [16].
2.8.7 **Factors affecting detectability**

The factors affecting performance detectability include detector properties, exposure parameters and image processing technologies.

2.8.7.1 **Detector properties**

A comparison study of four different types of detectors (i.e., CR1, CR2, DR1 and DR2) conducted by Fernandez et al. [71] demonstrate the detectors’ performance in determining image quality and low-contrast detail (LCD) detectability. CR1 is a conventional AGFA CR compact plus system with AGFA MD40 plates and a scanning resolution of 10 pixels/mm, while the CR2 is a needle-based AGFA DXS system with AGFA CR HD 5.0 storage phosphor plates. The flat panel system, or DR1, is one-piece amorphous silicon panel with a caesium iodide scintillator (200 µm pixel size), and DR2 is the most recently developed DR detector, with a detector of amorphous silicon and caesium and a 143 µm pixel size [71]. By obtaining the IQF\textsubscript{inv} for all detectors (using a reference value of 0.3 mGy for the chest PA examination), CR2 and DR2 provide better IQF\textsubscript{inv} than CR1 and DR1 (though DR2 shows better trends with low doses). Moreover, the storage phosphor system promotes improved image quality, but the dose reduction is limited compared to the flat-panel detectors. Images acquired through the needle image plate/line scanner provide better low-contrast performance than images obtained using image plate/flying-spot scanners. As a result, a structured CR2 produces the best LCDs of the three detectors, as well as better image quality [71].
In CR detectors, a capture element blur (as well as noise) occurs due to the movement of the laser beam during the scanning process of the image plate. Consequently, there is a delay in the photostimulable emission as laser beam quickly scans the phosphor screen [72]. However, the capture element blur is neglected for direct flat-panel detectors because the electric field application eliminates the charge dissipation. In the phosphor-based detectors, the capture element blur is the main source of blur. Decreasing the thickness of the sensitivity layer is a valuable method for reducing the blur [72]. Veldkamp et al. [50] conducted a study evaluating such digital radiography systems as CR, Direct-DR and Indirect-DR and comparing their performances. The high and low attenuation regions in the dual readout CR were better than the those in the single readout CR [73]. The LCD detectability for IDR was better than that for the CR system [7]. A comparison study of IDR and DDR demonstrated that IDR has better SNR values and, hence, better LCD detectability than DDR. Moreover, IDR has the ability to compromise between the radiation dose and image quality [74]. The differences among these several detector types demonstrate the variations in their performance in relation to detecting LCD. They also highlight the detectors’ limitations [50].

2.8.7.2 Exposure factors

The main exposure parameters in radiographic images are the tube potential (kVp) and the tube current in mA. The beam quality is altered by the radiographer using kVp and mA according to patient conditions and the particular radiological study. Modifications to the exposure factors influence patient’s dose and image quality simultaneously [30]. For example, adjusting the tube current (mA) controls the beam quantity, and the penetrating power of the beam is controlled by adjusting the tube potential (kV). Changing the exposure parameter could result in better penetration of the primary beam (by kV) and enhance the quality of X-ray production. Therefore, the scattered radiation is reduced due to good beam
penetration, which lowers the dose absorbed by the patient. In general, a high tube potential decreases the patient dose without reducing the image contrast to an unacceptable level. Moreover, the image quality can be improved by lowering the exposure time; however, this may affect the effective dose and entrance skin dose (ESD). Thus, maintaining the same mA is an alternative method that involves increasing the mA and decreasing the exposure time (s). This will improve the image quality by limiting the patient’s motion blurring through a shorter exposure time [30].

Enhanced subject contrast requires a low kVp (tube voltage) due to increased X-ray attenuation. This results in optimised LCD detectability. Moreover, this leads to increased digital system SNRs, as well as increased DQE values for the detectors. In contrast, the decreased kVp results in image blurring and great exposure doses due to increased mA [5, 48].

The mA plays an important role in decreasing the radiation dose and enhancing image quality. There is a significant correlation between low radiation doses and noise production. Reducing the radiation dose ultimately degrades the SNR level, thereby increasing the potential for noise and the loss of important details in the radiographic image. Moreover, an overexposed image shows a very black image area, which is not easily recognised by radiologists. Therefore, LCD detectability is increased with great radiation exposure, and it is essential to balance the detectability performance with the patient radiation dose [48].
2.9.7.3 Image processing techniques

Image processing affects image outcomes, such that the benefits of different image processing techniques result in better image quality. Methods like multi-frequency processing algorithms and un-sharp mask filtering can improve image contrast [5]. Another method to improve image contrast is edge enhancement, which can reduce noise and change pixel values to improve contrast [75]. This method can cause misrepresentation to the structures; this can be considered its main drawback. Moreover, elimination of the image noise can be achieved by a smoothing processing technique. This uses a subtraction processing technique to provide a better anatomically structured image by removing the superimposed structures. However, this method may reduce spatial detail [75]. The application of image processing technology is difficult because improving one feature of image quality can create another image artefact. Therefore, the optimisation process must be utilised in parallel to the system characterisation in order to enhance the image quality without resulting in the loss of detail [75].
Chapter 3 Low contrast detectability in Computed Tomography (CT)

The CT scan is a method for obtaining and reconstructing an image from a thin cross section of an object [1]. It can be differentiated from conventional radiography projections in two respects: 1) the CT produces a cross-sectional image that prevents the anatomical structures from being superimposed, which happens in projection imaging (such as CR and DR systems) due to the compression of three-dimensional structures onto two-dimensional recording systems. 2) The CT scan’s sensitivity to subtle differences in X-ray attenuation is greater than that normally achieved by projection imaging by a factor of at least ten [6]. The CT scan is based on calculations of X-ray attenuation through the section, which are achieved by utilising many different projections at various angles [6]. The development of CT modalities over several years is based on data acquisition geometries, which are divided into three types: parallel beam geometries (i.e., first-generation scanners)[1, 76], fan beam geometry (i.e., second- to fourth-generation scanners)[1, 76] and CT scanners in spiral geometry, which are considered a recent development (i.e., fifth-generation scanners)[76]. Moreover, the revolution of multi-detector computed tomography (MDCT) in the mid-90s is considered a major evolutionary leap in CT technology [6]. This technique impacted the increase in CT procedures around the world because it provided high temporal resolutions with short scan times, thereby influencing the formation of a true three-dimensional imaging tool [6]. The MDCT was followed by the dual-source computed tomography (DSCT) technique, which provides high temporal resolutions in cardiac cases [6].

The image quality parameters of CT scan modalities are classified as spatial resolution (divided into in-plane and cross-plane spatial resolution), contrast resolution, temporal
resolution, image noise and image artefacts. Each parameter (e.g., artefacts and image noise) has a unique effect on the image quality of CT scans and can be reduced to enhance the image quality [24] [1]. Moreover, there are many factors that affect the radiation dose and image quality of CT scanners, such as mA, kV, patient size, slice thickness and pitch values [6]. Knowledge of these factors helps to optimize image quality (IQ) while minimising the radiation dose applied to patients [6]. Because the lesions in most CT cases depend on image contrast, it is vital to highlight the importance of low-contrast detail (LCD) detectability to increase the diagnosis efficiency. The LCD detectability utilises the Catphan phantom to evaluate CT scanner performance. The better the LCD detectability in a CT scanner is, the better the lesion detection is and, hence, the better the patient diagnosis is [24].

The purpose of this Chapter is to review the recent development of CT modalities and to discuss the IQ parameters of CT scanners and their influence on image quality. In addition, the Chapter explains the factors that affect radiation doses and image quality. This Chapter includes an illustration of the LCD, how it is measured and the factors that influence it.

### 3.1 Recent computed tomography (CT)

In the early 1990s, the introduction of helical (spiral) CT scanners was considered a major breakthrough for CT technology [1]. However, the introduction of multiple-row detector CT scanners (MDCTs) was considered an even more significant evolutionary leap in CT technology. It launched with dual-row detector spiral CT scanners, which have the ability to produce dual scans for each X-ray tube gantry rotation [6]. In these scanners, the detector is split into two columns that enable one to obtain dual scans. By late 1998, all major CT manufacturers launched MDCT scanners capable of at least four slices per X-ray tube
rotation (Figure 3.1) [24]. The MDCT technique has consistently improved as the number of row detectors has increased from 4 to 256 and even 320 row detectors by 2007 [6, 77].

Figure 3.1: Timelines showing the different development milestones in the field of CT [6]. From 2008 onwards, a large number of clinical CT installations is estimated.

3.1.1 The clinical advantages of MDCT

- Shorter scan time: The examination time for a standard protocol can be significantly decreased. This has immediate clinical benefits, especially for non-cooperative patients.
- Extended scan range: Wider scan ranges can be achieved within the time period of one patient breath-hold. This is beneficial in angiography cases and in oncological staging.
- Improved through-plane resolution: It is very important to examine a scan range of interest within a breath-hold using slices that are substantially thinner than those achieved in single-slice CTs. The significantly improved through-plane resolution is beneficial for all reconstructions (particularly 3D post-processing, which is required in medical examinations) [78].
However, adding more detector rows no longer translates into clinical benefits [6, 24]. Therefore, recent advances in CT, like dual source CT (DSCT) and CT systems with area detectors, have focused mainly on the remaining limitation: insufficient temporal resolutions for cardiac CT exams and limited scan ranges for dynamic examinations of entire organs [6, 24].

### 3.1.2 CT systems with area detectors

CT systems consist of scanners with 320 x 0.5 mm collimation and 0.35 second gantry rotation times. This technique is optimised for the acquisition of sequential scan data that covers entire organs, such as hearts or kidneys, with aero table feeds. The resulting volume of the reconstructed scan is cone-shaped [79]. Thus, when using a detector collimation of 320 X 0.5 mm, a longitudinal coverage of 16 cm is possible at the iso-centre. By appending axial scans shifted in the z-direction (stitching), larger scan volumes in the z-direction can be covered, at the expense of overlap scanning. This process can be used efficiently in cardiac CTAs and dynamic or perfusion CTs. However, increased scatter radiations due to larger detector z-coverage represent a challenge for perfusion scanning. These scattered radiations can affect CT number stability and cause artefacts. When these scatter radiations induce noise, they reduce the contrast to noise ratio (CNR) of CT images [79].

### 3.1.3 Dual source CTs (DSCTs)

DSCTs have two acquisition systems, which are mounted into one gantry with an angular offset of 90 to 95 degrees (Figure 3.2) [6]. Each acquisition system provides overlapping 0.6 mm slices using a z-flying focal spot technique. One system covers the full field of measurement (FOM = 50 cm in diameter), while the other is restricted to the centre of the
FOM (26-33 cm) [79]. It is possible to obtain 64 or 128 overlapping 0.6 mm slices with double z-sampling, such that the shortest gantry rotation time is 0.33 to 0.28 seconds [79]. The main advantage of DSCT is that it provides higher temporal resolutions without the need for faster gantry rotation. In general, 180 degrees of scan data are used for cardiac CT image reconstruction [79]. Using DSCT, these data can be divided into two 90-degree data segments, which can be acquired via the two DSCT acquisition systems for the same anatomical level and the same phase of the cardiac cycle [79]. Thus, the total data acquisition time per image is decreased to a quarter of the gantry rotation time. The resulting temporal resolution is 75 ms for a rotation time of 0.28 seconds (independent of the patient’s heart rate) [79].

Figure 3.2: A photograph of a DSCT. The gantry illustrates two X-ray tubes and two detectors positioned orthogonally. Detector assembly A is larger than detector assembly B [6].

3.2 Image quality parameters in CT scans

Generally, the IQ of a CT scanner can be described using a number of key performance parameters, such as resolution, noise and artefacts. These parameters are affected by the operator’s selection of protocols, such as mAs, kVp, reconstruction parameters, etc [24].
3.2.1 Spatial resolution

Spatial resolution is the ability to distinguish small and closely spaced objects in an image [33, 80]. The spatial resolution in a CT image is measured in two orthogonal directions: in-plane (x,y) and cross-plane (z) spatial resolution [81].

The variations between these two planes disappear dramatically due to the recent development of multi-slice scanners [24, 79].

3.2.1.1 In-plane spatial resolution (or high-contrast resolution)

The in-plane spatial resolution is a measure of a system’s ability to reproduce small features inside the image slice (x,y) planes [82]. It can be specified by the term of modulation transfer function (MTF). The MTF is the ratio of the output modulation to the input modulation, and it can calculate the response of the system to different frequencies. When the MTF curve is flat, this means that the system is ideal and that the system’s response is independent from the input frequency [81]. The modulation factor as a function of spatial frequency is represented by the MTF response. For instance, Figure 3.3 shows 1 mm bars and spaces with a spatial frequency of 5 line pairs percentimetre (lp/cm), since one centimetre contains five cycles of bar and space pairs. If the system transfer function at a spatial frequency of 5 lp/cm is greater than the original amplitude by a factor of 0.7, the modulation at this spatial resolution is 70 percent [82]. Consequently, only 70 percent of the amplitude of the input object at this spatial frequency is transferred to the image [82].
Figure 3.3: The spatial details of the resulting image are blurred by the spatial transfer function of the CT system. The spatial resolution is a calculation of the spatial transfer function, which is provided as an MTF system. The limit resolution occurs when the MTF curve approaches the first zero [82].

The spatial resolution is not symmetric; however, it changes as a function of position inside the scan field of view (SFOV). Moreover, the image resolution is influenced by several factors, which are related to the scanner design and the parameters that have been selected by the protocol operation. These factors are: 1) the detection function (i.e., the active width of the detectors pixels); 2) the focal distribution function (i.e., the shape and size of the focal spot); 3) the projection number per gantry revolution; 4) the sample spacing between rays; and 5) the reconstruction process (including algorithm selection) [81, 83].

The focal distribution function has a major impact on the spatial resolution of an image. A small focal size is required to maintain great resolution (Figure 3.5-a). Moreover, the best spatial resolution is achieved by the narrowest beam width. The spatial resolution analysis is
divided into two orthogonal directions: radial and azimuth [81]. The radial direction is the line that links the location of the point object to the iso-centre of the system. The azimuth resolution is tangential to the radial direction (Figure 3.4) [81]. The increment in the fan angle distance from the centre of the SFOV (iso-centre) results in an increase in the projected width of the focal spot. Therefore, the radial resolution of the image is reduced due to the increase in distance from the iso-centre [82]. The azimuthal resolution is decreased when the distance from the iso-centre is increased, according to the view numbers sampled during one gantry revolution. Moreover, a central SFOV provides a better resolution, thereby illustrating the importance of positioning a patient at the centre [81]. Reconstruction algorithms (or reconstruction filters or kernels) can either enhance or reduce the spatial resolution of an image’s edges. Thus, these can affect an image’s appearance, but they cannot change the inherent SNR or spatial resolution limit [81].

![Figure 3.4: Radial and azimuthal directions](image)

3.2.1.2 Cross-plane spatial resolution (or slice sensitivity profile)

The cross-plane spatial resolution refers to the spatial resolution in the longitudinal Z axis, and it is important in reformatted 3D image representations [1]. This spatial resolution is
referred as slice thickness, which is the full width at a half-maximum intensity of the slice sensitivity profile (SSP) [82]. The SSP defines a system’s response to a Dirac delta function $\delta (z)$ in $z$, and the actual system response curve to $\delta (z)$ is usually used to represent the SSP. This curve is replaced in many cases by two numbers: the full-width number at half-maximum and the full-width number at tenth-maximum (FWTM). The FWTM is the distance between two points on the SSP. Essentially, the Dirac delta function is often estimated by objects that have thicknesses smaller than the slice thickness of data acquisition (e.g., thin disc or small bead) [6].

In a multi-slice CT system, the ray enclosed in the Z-axis between the focal spot and a detector cell controls the Z-axis resolution limit. This resolution is affected by the projected length of the focal distribution function; hence, it varies across all detector rows. The Z-axis resolution is reduced in the detector rows located close to the cathode side because the focal spot is larger, thereby producing wider inherent slice sensitivity (Figure 3.5 C and D). The SSP can be affected by reconstruction processing. For example, the production of slices larger than those allowed by the inherent detector row aperture limitation may occur due to a combination of detector row data in helical scanning. The shape of the SSP is determined by the weightings of different detector row data, which control the Z-axis resolution [82]. Three-dimensional reconstruction methods decrease cone beam artefacts; however, they also increase the slice sensitivity profiles of the outer detector rows, as compared to the detector rows close to the centre of the detectors (since the X-ray samples are more perpendicular to the detector face). The contrast of objects inside the image is affected by the SSP; specifically, it is reduced if the contrast is smaller than the extent of the SSP [82]. The importance of Z-axis resolution has recently increased due to the resolution’s benefits in clinical settings, particularly with regard to coronal image quality [81].
3.2.2 Contrast resolution

The contrast resolution refers to the ability to discriminate between two regions of different attenuations that are located close to one another. The major limitation of contrast resolution is caused by the amount of quantum noise (mottle) in the reconstructed image [84]. The factors affecting this contrast resolution are: the photon flux, the slice thickness, the detector sensitivity, the image display and reconstruction algorithms [6].

In the photon flux effect, the shorter exposure time and smaller aperture size cause larger proportional fluctuations in the observed projection measurements. Because the mottle amount in the image is inversely proportional to the square root of the number of detected photons, the photon flux effect can be controlled [84].
The reconstruction of CT images using thicker slices will reduce the amount of noise and enhance the contrast resolution[85]. However, it will also decrease the spatial resolution and lead to partial volume effects [6]. The change in slice thickness controls the beam width entering each detector; hence, it influences the number of X-rays proportionally detected. For instance, a slice thickness of 10 mm doubles the detected number of X-rays entering the 5 mm slice thickness [80]. In an MDCT, the detector size is decreased, and the computational performance is increased. Therefore, the compromise between spatial resolution and contrast resolution is unnecessary because it provides high spatial resolution via thinner detectors and because its reconstruction process involves combining thin slices into thick slices to improve the contrast resolution [6].

The detector element is composed of a radiation-sensitive solid state material (e.g., cadmium tungstate), which transforms the absorbed X-rays into visible light. High detector sensitivity has great detection efficiency (i.e., a high atomic number) and a very short afterglow time to allow fast gantry rotation speeds, which can be used in ECG-gated cardiac imaging [86]. To compare the properties of different detectors, DQE, which calculates the different MTFs of the detectors, is used [17].

The image display’s contrast resolution depends on window width and window levels. The window width (WW) is the number of selected grey shades, while the window level (WL) is the mid-point of the selected grey scale. In low-contrast structures (e.g., the liver), a narrow WW and a lower WL are used to highlight structures’ details, which are associated with noise (relatively higher doses are required for structure delineation). For high-contrast structures (e.g., bones), a wider WW and WL are utilised; this can reduce the visual appearance of image noise (i.e., lower doses can be used for better display) [82].
Reconstruction algorithms (also known as reconstruction kernels or filter) are applied to reconstruct raw CT projection data; thus, these play an important role in the assessment of IQ in CT. The selection of reconstruction algorithms always contributes to a trade-off between contrast and spatial resolution. The type of visualisation needed for interpretation determines the reconstruction algorithm selection. For example, an operator should use smooth algorithms to detect low-contrast objects, like liver lesions, while sharp reconstruction algorithms are required for temporal lobe or lung detection [6]. However, in most cases, small, low-contrast objects are not significantly influenced by choice of reconstruction algorithm. This is due to the reduced frequency of both the object and the noise in a similar manner via the reconstruction algorithm process; hence, the SNR is not altered [82].

3.2.3 Temporal resolution

The temporal resolution refers to the length of time required to image an object. To achieve a high level of temporal resolution, an operator must obtain an image faster than the structure is moving in order to avoid motion artefacts [87]. Such resolution is vital in two main clinical applications: cardiac imaging and CT fluoroscopy [81]. Several factors affect the spatial resolution, such as the acquisition mode, the gantry rotation time, the pitch and the type of image reconstruction [6, 87].

In the acquisition mode, the MDCT system utilises two approaches: prospective electrocardiogram (ECG) triggering and retrospective electrocardiogram (ECG) gating [6]. The temporal resolution in the first approach ranges from 200 to 250 ms. Its main advantage is low radiation exposure, which is achieved by the acquisition of the projected data over short time periods. Prospective ECG triggering is used for structures with high CT numbers
In contrast, retrospective ECG gating provides temporal resolutions ranging from 80 to 250 ms. Despite the high radiation exposure in this method, it allows for better spatial resolution and continuous coverage in the patient’s longitudinal direction. Because the images can be reconstructed with overlapping slice increments, this method is mainly used for cardiac coronary artery imaging [6].

The temporal resolution is affected by the gantry rotation time; thus, the better the temporal resolution is, the faster the gantry rotation can be. The gantry rotation time is defined as the time required to complete one full rotation (or 360 degrees) of the X-ray tube or of the detectors around the subject. However, the enhanced time of rotation leads to an increased G-force in the heavy metallic instrument rotation, which makes optimising the gantry rotation time quite difficult [88].

In MDCT, the pitch is known as the longitudinal or Z-axis table increment per 360-degree gantry rotation / beam collimation. When the pitch is increased, the scan is finalized faster, resulting in fewer motion artefacts caused by voluntary or breathing motions. However, the spatial resolution can suffer due to greater image noise [87].

The reconstruction method involves partial scan reconstructions and multiple segment reconstructions. The partial volume reconstructed data are obtained by rotating the fan angles of the CT detectors and the X-ray by 180 degrees. The temporal resolution is limited to 260 to 280 ms, with a gantry rotation of 500 ms [6]. To improve the temporal resolution, CT manufacturers increase the gantry rotation time to around 300 ms; hence, the temporal
resolution can be high as 170 to 180 ms, which is associated with G-forces [6]. Multiple-segment reconstruction promotes better temporal resolution than partial-segment reconstruction. This is because, in the second one, the scan projection data need to perform a partial scan reconstruction, which is selected from different sequential heart cycles instead of from a single heart cycle. Because projection data may be selected from three or four varied heart cycles, the temporal resolution may be as low as 80 ms [6].

The use of multi-segment reconstruction, when accompanied by data scans of subsequent heart cycles, can significantly improve temporal resolutions. Recently, the DSCT technique has provided great temporal resolution without the need for faster gantry rotation[79]. It separates the 180 degrees of scan data into two 90-degree data segments acquired by the DSCT system during the same phase of a patient’s cardiac cycle and at the same anatomical level. Therefore, the total acquisition time per image is decreased to a quarter of the gantry rotation time. As a result, the temporal resolution is improved (to 75 ms) for a rotation time of 0.28 seconds, independent of the patient’s heart rate [79].

3.3 Image noise

Image noise refers to unwanted image details that obstruct the visualisation of a focal abnormality and inhibit image interpretation [72]. The CT number fluctuates (e.g., between -1 and +1) in a uniform phantom, like water. These random fluctuations appear as graininess on a CT image (image noise), which is caused by the limited number of photons that contribute to a CT image. The image noise is related to the X-rays involved in detector measurements (rather than individual pixels). Therefore, the CT noise is associated with the number of X-rays contributing to each detector measurement [80]. Generally, there are three types of
sources that influence image noise: the quantum mottle noise, the inherent physical limitations of the system and the image generation process [81].

In a CT scan, the energy source emits electromagnetic radiation, or photons, which naturally affect the image noise. The probability of photons being detected in a CT detector at any point during a specific interval of time is constant. The local random distribution of photon emissions causing variations in measured intensity is known as quantum mottle noise, and it can be described as a Poisson process [40]. In a Poisson process, the square root of the average signal ($\mu$) is equal to the standard deviation ($\sigma$) measurement of the signal for all identical sensor regions. This concept suggests that reducing the quantum mottle noise increases the signal energy flux relative to the total intensity [40]. For example, the image noise increases with the decrease in kV (which can be beneficial in cases of thinner patients). Moreover, the mAs are directly proportional to the radiation dose, such that higher mAs result in low image noise due to being inversely proportional to the square root of the mAs [88].

The second source that affects the image noise in a CT scan is the inherent physical limitations of the system. These limitations include scattered radiation, X-ray translucency of a scanned object, electronic noise in the data acquisition system [89], electron noise in a detector’s photodiode and many other factors. However, the noise elimination options available to CT operators are limited [81].

The image generation process is classified into different areas, such as reconstruction algorithms, reconstructions parameters and calibration effectiveness. The application of
specific reconstruction algorithms (e.g., filtered back projection (FBP) and iterative reconstruction (IR) can enhance noise performance [81]. IR algorithms have the ability to improve spatial resolution and image noise compared to the FBP algorithms. The reconstruction parameters include a selection of carious reconstruction filter kernels, image matrix sizes, reconstructions of FOVs and post-processing techniques [81]. Generally, noise levels are increased by high-resolution reconstruction kernels. This is due to the kernel enhancement of high-frequency content in projections, in which cases most of the noise is also present at high frequencies. The calibration techniques utilised in CTs to condition the collected data are not perfect. The residual errors, however, manifest as artefacts of small magnitude, and they often cannot be visually detected. They do impact the standard deviation measurement and, thus, should be considered part of the noise source [81].

### 3.4 Artefacts

Artefacts in medical imaging are misrepresentations of tissue structures. Because CT images are derived from a number of X-ray projections with thousands of measurements, these projections may suffer inaccuracies during the process of image reconstruction, ultimately appearing as artefacts. It is vital to correct these artefacts in medical imaging to avoid misdiagnoses. Table 3.1 demonstrates the most common artefacts that manifest in day-to-day clinical practice [6].
Table 3.1: The most common artefacts that appear in CT exams [6, 90]

<table>
<thead>
<tr>
<th>Artefact name</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Streaking artefacts</td>
<td>Appear as intense, straight lines across CT images, and often appear as bright or dark lines. These are caused by inconsistencies in individual calculations. Because they do not mimic tissue structures, they rarely cause misdiagnoses in patients. However, they may lead to image quality deteriorations and make images unreadable.</td>
</tr>
<tr>
<td>Ring or band artefacts</td>
<td>These appear as full or partial rings or bands that are superimposed on CT images. They are caused by errors in single or multiple DAS channels in detectors. Partial ring artefacts interfere with the diagnosis process more than full ring artefacts, as these can mimic tissue structures—and, hence, are difficult to identify.</td>
</tr>
<tr>
<td>Partial volume artefacts</td>
<td>These occur when an object is partially introduced into the scanning plane. For example, they can occur when scanning a plane with uniform density that contains a partial intrusion of a high-density object, causing shading artefacts to appear in the image. This type of artefact mainly occurs in thick slices. To eliminate such artefacts, thin slices (normally applied by MDCT) must be acquired. Such artefacts are completely different from the partial volume artefacts caused by errors in the average attenuation of the materials inside the image voxels.</td>
</tr>
<tr>
<td>Photon Starvation Artefacts</td>
<td>These are caused by deficits in X-ray photons in specific regions due to the technique being insufficient to penetrate the anatomy structure of a patient (particularly in cases of obese patients), resulting in few photons being received by the detectors. These artefacts can also form streak artefacts due to inconsistencies in image production. To avoid these artefacts, the operator can apply adoptive tube current modulation and filtration, which</td>
</tr>
</tbody>
</table>
promote adequate amounts of X-ray photons while scanning around thicker patients and keeping radiation doses as low as possible.

| **Patient motion artefacts** | During data acquisition, a patient’s motions may lead to inconsistencies in projection data, resulting in motion artefacts. To reduce or eliminate such artefacts, patient-immobilisation devices are used, particularly for paediatric patient. In rare cases, even sedation may be used. The most efficient method to eliminate motion artefacts is to instruct the patient to hold his or her breath or to stay still during data acquisition. |
| **Beam hardening artefacts** | These appear as streaks or cupping or as dark areas that appear between highly attenuated structures, such as bones. They are defined as increases in the mean energy of an X-ray beam as it passes through a patient. They induce variations in X-ray intensity: that is, the beam passes more forcefully through the centre of the focal structure than through the edges, producing cupping artefacts and causing misdiagnoses due to the artefact being similar in shape to some pathologies. The use of adequate beam filters, such as flat filters or beam-hardening correction software, helps to reduce these artefacts. |
| **Metallic artefacts** | This type of artefact is related to any metal worn by patients, such as jewellery or in vivo metal (e.g., prosthesis). An in vivo metallic object may cause streak artefacts as a result of beam hardening. Minimising these artefacts is possible through the use of partial scans or thin slices. Moreover, there are many metal artefact software correction algorithms that can reduce the effects of metallic artefacts. |

### 3.5 Factors affecting radiation dose and image quality

A trade-off has to be made between IQ and patient dose [91]. There are several factors that influence image quality: patient size, reconstruction algorithms, mA, kVp, etc. The trade-offs
among these factors are not straightforward; they depend mainly on the operator to improve image outcomes [6, 91].

3.5.1 Patient size, shape and anatomy

Patient size controls the technical factors set for a CT exam. For good penetration through large focal area (e.g., a large patient), it is necessary to select a high tube current and tube voltage [6]. It is also vital to provide the lowest dose capable of maintaining the appropriate image noise to achieve a good diagnostic. The patient size issue is resolved by CT manufacturers through the use of anatomic tube current modulation (ATCM) [82]. This can adapt to different patient sizes, increasing mA for larger patients and decreasing mA for smaller patients [82]. Moreover, the shape and anatomy of the patient influence the radiation dose attenuation and, hence, the IQ. For example, in highly attenuated regions, like the shoulder, the noise is great for X-rays taken in the direction of the high attenuation. These high attenuations are produced due to insufficient photon detection, and they also subsequently affect the electronic noise that can be eliminated through reconstruction algorithms. Patient attenuation explains how X-rays interact with matter. Different anatomical parts affect the attenuation and noise in an image. Because the lungs are less dense than the pelvis, when using the same technique, the noise in the lung is lower than that in the pelvis due to lower X-ray attenuation [82].

3.5.2 Tube current (mA) and peak tube voltage (kVp)

Increased tube currents lead to reduced CT image noise. The tube current is proportional to the patient’s dose and the number of photons in the image such that, if the tube current is increased by a factor of four, the image noise is reduced by one half (the inverse square root
of four) [82]. The product of the tube current and the scan time is measured in milli-amperes seconds (mAs) [6]. Modern CT scans use automatic exposure control (AEC) to reduce patients’ doses. AEC modulates the X-ray beam intensity according to the patient’s anatomy to generate the preferred image quality [88].

The peak tube voltage kVp is the peak potential variation between an anode and a cathode, which accelerates the electrons towards the anode for X-ray production [6]. The number of X-ray photons is largely influenced by kVp selection. The image noise is decreased through an increased number of photons; however, the patient’s dose is increased. Moreover, the contrast between human tissues (i.e., the HU value) decreases with increased kVp [82].

3.5.3 Pitch, scan time and slice thickness

The pitch is the ratio of the table feed per gantry rotation to the total X-ray beam width [6]. A patient’s dose and the number of X-ray photons are inversely related to the helical pitch [6, 82]. Thus, increasing the pitch value leads to a lower dose and, hence, increases the image noise [92]. Normally, in abdomen and pelvic cases, a pitch value of more than one is sufficient (P>1). However, in some cases that require higher resolutions (e.g., cardiac CTs), a pitch value lower than one is used (P<1) [6].

The scan time is the time required for the gantry to make one revolution. The patient dose and the number of photons in an image are proportional to the scan time. Increasing the scan time reduces the image noise and increases the potential for anatomic patient motion artefacts [82]. Another method for reducing image noise is to increase the slice thickness. The slice
thickness is similar to the tube current with regard to image noise because a wider slice thickness allows more X-ray photons to be included in the data to reconstruct the image. The noise reduction is correlated with the inverse square root of the slice thickness [82, 92].

3.5.4 Reconstruction algorithms (also called reconstruction filters or kernels)

The spatial frequency of an image’s noise is affected by the reconstruction algorithms used. Reconstruction algorithms have the ability to either boost or decrease the frequency of both noise and subject in similar manners; hence, they do not change the signal-to-noise-ratio [93]. For instance, a spatial frequency beyond 51 p/cm does not involve the information content of a 3 mm object [82]. Hence, the application of reconstruction algorithms helps to limit spatial frequencies, thereby enhancing the delineation of the objects size and become more visualised. The selection of reconstructed algorithms relies on experience and on the preference of the human viewer and the image task [82].

3.5.5 Filters, bow-tie filters and dose profiles for various patient sizes

Filters are absorbent materials that are placed between the X-ray tube and the patient to absorb low-energy X-rays or softened and hardened X-ray beams. Their low energy contributes disproportionally to the dose required for the patient (particularly with regard to the skin dose). Therefore, it is desirable to harden the X-ray beam [6]. Filters increase the average energy of the X-ray beam, and they are shaped as bow ties in the CT machine in order to modulate the X-ray beam to deliver a uniform X-ray intensity across the scanning
object, which results in a better uniform dose distribution to the scanning region. These filters are available for head, cardiac and body cases for different patient sizes [6, 92].

### 3.6 Evaluation methods for CT image quality

The information about a patient and the possible abnormalities found in medical imaging are transferred to radiologists by two major stages: 1) data acquisition and image formation and 2) processing and display [53, 94]. Stage one relies on the physical and technical properties of the equipment, while stage two focuses on the importance of the radiologist’s performance and his/her ability to detect and interpret image structures. Accordingly, the evaluation methods for image quality are classified into three categories: physical measurements, diagnostic performance measurements and psychophysical measurements [53, 94].

Physical measurements define the quality of the equipment itself. For instance, it is essential to evaluate new types of image detectors [17]. The most well-known measurement is the DQE measurement, which provides a physical evaluation of the detectors. DQEs describe how the system refines incoming X-ray photons in relation to contrast, noise and resolution. The DQE equation contains values of MTFs (describing the resolution features of a system), NPSs (describing the noise properties of a system) and SNRs (describing the energy or the photon-weighted difference of the incident beam) [17]. These terms were discussed in (Chapter 2, Section 2.8).

Diagnosis performance focuses mainly on the process of diagnosing patients in clinical cases, and it compromises almost all aspects of performance evaluation that represent simple preference studies [17]. Visual grading analyses (VGAs) and receiver operating
characteristics (ROCs) are good examples of this method (these terms were discussed at Chapter 2). A VGA is the perfect method to assess clinical image quality by means of grading anatomical structures[61]. ROC analyses are used to measure the accuracy of different types of examinations [57].

3.7 **LCD detectibility and CNR**

Low-contrast detail (LCD) sensitivity or detectability defines the potential to image small objects (typically with varied diameters in millimetres) with object contrasts of only a few Hounsfield units (HU)[95]. The low contrast (LC) resolution term describes the potential to image periodic patterns of LC objects. The ATS phantom is used to measure (LC) resolution which is proposed by the American Association of Physicists in Medicine (AAPM). This phantom is composed of circular rows, such that the diameter varies from row to row. These rows are situated inside plexiglass housing. These circular inserts have adjustable attenuation coefficient values, which are expressed in a Hounsfield scale, and the whole of the phantom’s inside is filled with water [76].

There is a large variety of data resulting from the absence of a general consensus regarding LC detectibility specifications[96]. For example, some scanner data sheets include values for LC sensitivity, others specify LC resolution and still others contain both. The comparison of the resolution data collected by ATS phantoms on different scanners is difficult due to the variation in contrast according to X-ray beam quality [76, 96]. Moreover, the diameter used in the ATS phantom varies around 16 cm, while, in Catphan, the diameter varies from 15 to 20 cm[96]. Different phantom dimensions inhibit the comparison of LC sensitivity data, since the image noise level at a constant dose varies significantly with object diameter. According
to Suess et al. [96], the Catphan section can only be used to measure LC detectability while the ATS phantom promotes inadequate performance, thereby preventing the object contrast from being reproducible or reliable [76, 96].

In psychophysical measurements, the observer responses to certain visual stimuli and gives a report of what he/she sees (e.g., LCD). The visual stimuli applied in such experiments are simple, such as disc-shaped objects. They are also usually associated with different contrasts and diameters to determine contrast-detail diagram. [53].

Figure 3.6 illustrates the earliest design of a phantom utilised to calculate low-contrast resolution. The metal rods are submerged in water and arranged in rows. Both the diameter $d_i$ and the distance between the rods $2d_i$ increase as the index $i$ of the row increases. The low-contrast resolution of the scanner is controlled by the smallest diameter of the, rod which is shown as a distinct element in the reconstructed image [24, 76].

Figure 3.6: An early phantom used to determine LCD [76].
LCD, or low-contrast performance, is defined as the smallest object that can be visualised at a
known contrast level and dose. Figure 3.7 illustrates a low-contrast phantom (Catphan). This
phantom contains three sets of discs with different contrasts (0.3%, 0.5% and 1%) and
different disc sizes (2 --9 mm and 15mm)[97].

Figure 3.7: An axial CT image of the CTP515 module in the Catphan 600 phantom. The contrast levels of these
groups are 1, 0.5 and 0.3, as shown in the Figure. The nine object sizes range from 2 to 15 mm [97].

The SNR of the image cannot reveal the noise effect because the visibility depends on the
contrast (the differences among signals). A highly overexposed radiograph may have a high
SNR, but contain no useful information about the imaged object [40]. CNR is very useful in
estimating the effect of noise in image information. The CNR can be represented by the
following equation [40]:

$$\text{CNR} = \frac{\mu_A - \mu_B}{\sigma_{BG}}$$  \hspace{1cm} (3.1)

$\sigma_{BG}$: The noise measured as the standard deviation of the background
The contrast described as the difference in intensity between an object and its background [40].

In a CT, the level of contrast is specified in terms of the percent of the linear attenuation coefficient. A one percent contrast means that the mean CT number of the object varies from its background by 10 HUs [24].

Figure 3.8 shows a series of nine circles of increasing contrast in relation to a mid-grey background. The contrast values range from 0.004 to 0.086, while the standard deviation of the noise is 0.173. This results in CNR values ranging from 0.02 to 0.50, thus the circle at the left end of the middle row is barely detectable (its CNR = 0.35) [40].

Figure 3.8: Demonstration of the effects of noise and contrast object visibility. The image contains nine circles of equal diameter, with contrast values ranging from 0.004 (top left) to 0.086 (bottom right). The standard
deviation of the noise is 17 percent of the possible intensity range. This results in CNR values ranging from 0.02 to 0.5. The circle at the left end of the middle row has a CNR of 0.35 (the red arrow) [40].

The ease of visual detection of an object in an image, or the conspicuity, relies on both the object’s size and its CNR. In the Catphan phantom, the largest circle is easy to see, while the smallest circle is effectively hard to visualise. The conspicuity or visibility of an object is proportional to its area. However, for the same area, more compact objects (e.g., circles) are more visible than less compact objects (e.g., stars) [40]. The CNR is affected by the slice thickness, such that thicker slices result in better CNRs, but poor spatial resolution [6]. An adequate CNR is required to detect tiny, low-contrast details, like non-calcified atherosclerosis plaques in the coronary artery [82].

IQ assessments for CNR follow one of two approaches: the objective method (using software) and the subjective method (using an observer) [98]. Thilander-Klang et al. [97] suggested that the use of a more advanced quantitative evaluation of the reproduction of low-contrast details, such as the automated calculation of theoretical contrast detail data supplied by commercial software and intended to be used with Catphan, like CT AutoQA Lite (IRIS QA, Fredrick, MD, USA), would improve performance compared to the simple method of analysing the pixel standard deviation which, in turn, was more reliable than the judgment of a human observer [97].

3.7.1 Factors affecting LCD

The LCD definition implies that the visibility of an object is highly affected by the presence of noise. There are many factors that influence the noise levels in reconstructed images and
the LCD. Some of these factors, like kVp, mA, the scan speed, the helical pitch and the slice thickness, are controlled by the operator, and some are controlled by the CT system (e.g., reconstruction algorithms)[24].

3.7.1.1 X-ray tube voltage (kVp) and tube current (mA)

A low kVp (~80 kVp) increases the tissue contrast however, this technique requires longer exposure times resulting in unacceptable image quality due to patient motion. A high kVp decreases the inherent subject contrast of the tissue in a transmitted X-ray beam [47]. For example, 140 kVp spectrum not only provides photons at increased energy, but also increases the photon intensity of all photon energies, hence the subject contrast is reduced. Noise is reduced in the image by increasing the number of photons, which increases the patient dose. It is very important to recall that the CT contrast (HU values) of human tissue relative to water is decreased at increased kVp settings. Therefore, despite the noise reduction achieved by higher kVp values, the CNR for features of interest may actually decrease for the dose used (especially for iodine contrast-enhancement tissues) [82]. In dual-energy MDCT imaging with low kVp, the arterial enhancement of the CT pulmonary artery (CTPA) is significantly increased in routine studies. Moreover, low kVp imaging permits a reduction in the contrast levels used for CTPA without reducing the image quality [99].

To explain the relationship between the mA and the LCD detectibility, we can consider a low-contrast portion of a GE quality assurance phantom that was scanned using two different tube currents: 200 mA and 50 mA (with all other parameters kept the same) [24]. In the 50 mA current, the noise is higher than that of the 200 mA case by a factor of two (which is one fourth of the dose). All four low-density holes are clearly identified, while the smallest holes
in the 50 mA scan are obscured by the noise (Figure 3.9). Thus, increasing the mA leads to noise reduction and better LCD detectability [24]. However, increasing the mA also results in increased radiation doses to the patient. In some CT scanners, this effect may cause the system to switch to a larger focal spot, which can reduce the spatial resolution, particularly in areas away from the iso-centre [24].

Figure 3.9: Demonstration of noise in LCD: A) a scan acquired with 200 mA and standard algorithms and B) a scan acquired with 50 mA and standard algorithms [24].

3.7.1.2 Helical pitch and slice thickness, scan speed

Increasing the helical pitch can decrease the number of X-ray photons, which can increase noise and decrease LCD detectability. The helical pitch is the ratio of a patient’s table travel per gantry revolution divided by the total Z-axis detection aperture at the iso-centre [6]. Increased slice thicknesses reduce noise and, hence, improve LCD. This is due to the great accumulation of X-ray photons in the data for the wider slice that can be used to reconstruct
the image [24]. Despite the noise reduction resulting from the great slice thickness, which occurs without a noticeable increase in patient dose, this approach also decreases the Z-axis resolution of the image and causes a reduction in CNR due to the volume averaging of small features. Von Falck et al. [100] recommended that it is possible to combine thin-section scanning with sliding thin slab-averaging during data post-processing and image display. This approach is useful for obtaining high through-plane spatial resolutions associated with small, minimal partial-volume averaging effects and good depictions of low-contrast lesions [100].

Reducing the noise increases the scan time. Consequently, the image quality and the LCD detectability of the CT images are enhanced. However, increasing the scan time may also increase the potential for patient motion during the CT exam. It is common to use milliamp seconds (mAs) as a relative indication of X-ray photon numbers [101].

3.7.1.3 Reconstruction algorithms

Reconstruction algorithms are computer software programs that are used to reconstruct raw CT projection data, and they play a vital role in CT image quality. Generally, the classification of these algorithms are: very smooth, smooth, medium smooth, sharp and ultra sharp, depending on the preferred image quality [6]. The choice of reconstruction algorithms involves trade-offs between the contrast resolution and the spatial resolution [6]. High-resolution CT images are provided by high-resolution reconstruction algorithms; however, these algorithms also create a great amount of image noise and edge-enhanced artefacts. Smooth algorithms decrease the image noise and improve the LCD; however, they also decrease the sharpness. Thus, these types of algorithms are useful for obese patients, in whom the image noise level is great because few X-ray photons reach the detectors [6]. In MDCT
scanners, increasing the computer capability allows the reconstruction of a multiple image set created by applying different reconstruction algorithms to one set of raw data. Eventually, it becomes possible to obtain data with the thinnest slice possible, to reconstruct high-quality 3D CT images and to simultaneously apply smooth algorithms to achieve less image noise. The choice of reconstruction algorithms relies on the type of visualisation required for the interpretation. For instance, the detection of low-contrast objects (e.g. small liver lesions) requires smooth algorithms (Figure 3.10), while, to detect abnormalities in the temporal bone or lung, sharp reconstruction algorithms are chosen[6].

Beister et al. [102] compared iterative reconstruction IR algorithms and the more common filter back projection FBP algorithms. The FBP algorithms are based on only single reconstructions, while IR algorithms use several repetitions, in which the current solution converges toward a better solution resulting higher computational demand. In some aspects, the IR algorithms are preferable to FBP algorithms because they permit the integration of various physical models, which can decrease image noise and result in different artefacts, depending on the model degree [102]. The main advantage of IR, besides reducing noise, is that it may reduce the dose to the patient which is considered very important in infant CT exams [102].
Figure 3.10: The reconstruction algorithms effects on the CT image quality is illustrated in abdominal images. The CT abdominal images acquired on 64-slice MDCT scanner were reconstructed with: A) very smooth, B) smooth, and C) ultra-sharp reconstruction algorithms. The noise of the image increases sharply with the use of the ultra-sharp reconstruction algorithms.
Chapter 4 represents the general aspects of materials and methods used in most of this thesis experimental work. Specific aspects of these materials and methods are later included in the results and discussion Chapters.
Chapter 4 Materials and Methods

This Chapter demonstrates the materials and methods used in this thesis. It includes:

- An illustration of the contrast details phantoms (CDPs) used in X-ray based modalities to test the low-contrast detail (LCD) detectability including 1) commercially available CDP (CDRAD 2.0) for conventional radiography and 2) a specially-designed CDP for CT modalities.
- An explanation of the application, correction procedure and results analysis of the CDRAD 2.0 used for conventional radiography.
- The employment of the information loss (IL) theory in the CDPs previously mentioned in the first point to evaluate the degradation of image quality (IQ).
- Grid applications techniques for enhancing the IQ.
- A demonstration of different tools (e.g. the computed tomography dose index CTDI, radiographic contrast media and radiochromic films).
- An introduction to the image J software for image analysis.

Explicit linkages are mentioned between the material/method and the Chapters in which they are used.

4.1 Contrast Detail Phantom (CDRAD 2.0)

The Medical diagnoses rely on more than just the image; they also depend on observers’ perceptions. Therefore, it is crucial to monitor the information contents of images. Contrast detail phantoms (CDPs) like CDRAD 2.0, aim to examine observers’ perceptions (used in Chapter 5, 6, 7 and 8). These phantoms have the ability to quantify the visibility of details for
different contrasts, as observed by radiologists. They can be used in different fields of diagnostic imaging systems, such as digital subtraction angiography and fluoroscopy [103].

4.1.1 The phantom’s description

The CDRAD phantom is composed of a Perspex tablet (a square of 265 X 265 mm) with a thickness of 10 mm, in which 225 cylindrical holes of various sizes and depths are drilled in the phantom medium. The diameter of the holes varies in size from 0.3 mm to 8 mm [3]. This range is equally distributed across 15 depths. These depths range from 8 mm (providing high contrast) to 0.3 mm (providing low contrast). The X-ray image depicts 225 squares organised in 15 columns and 15 rows [3]. In every square, either one or two spots are present that represent the images of the holes. The first three rows have only one spot, while the other rows have two identical spots in each square: one spot in the middle and the other in random corner. This allows the verification of the detection of each object. The CDRAD phantom design avoids patterns, leading to easy recognition [3].

4.1.2 The phantom’s application

To acquire an X-ray image, the CDRAD phantom is positioned on the patient table over the automatic exposure control (AEC) and the cassette. Individual experiments are determined by such exposure techniques as manual and automatic exposures, with grid and without grid, focal spot size and tube potential. Moreover, the phantom can provide measurements related to the following [103]:

- The comparison of IQ with different film-screen combinations.
- The use of varied densities to determine the optimum background density.
- The application of different tube potentials to determine the optimum exposure technique.
- The use of different object thicknesses for IQ comparisons (this is achieved by varying the Perspex amount at a fixed density).
- The impact of filtering by varying the filter thickness added.

To simulate different patient thicknesses, different Perspex thicknesses should be added both over and under the phantom [103].

### 4.1.2.1 The phantom image’s assessment

The CDRAD phantom must be assessed by at least three experienced observers to provide the best results. To improve the validity, three independent images made during the same sitting can be evaluated using a score from the CDRAD phantom. The evaluation of the image is focused on the area where the holes are just visible, which is indicated by the position at which the non-central holes can be seen. At least three fields, including at least one non-visible choice in each column or row, must be used to conform to the suggested correction scheme [103].

The indicated locations of the eccentric holes should be compared to the true hole positions in the phantom. There are specific rules for evaluating the results (e.g. consider the four nearest neighbours of the field under examination). The assessment of a specific field always refers to the original observations of the nearest neighbours [103].
4.1.2.2 Correction scheme

In each observation, there are three options for the observer [103]:

- T: the eccentric hole is indicated at the true position.
- F: the eccentric hole is indicated at the false position.
- N: the eccentric hole is not indicated at all.

There are also two main rules within the correction scheme [103]:

1. A True requires two or more correctly indicated closest neighbours to remain a True.
2. A non-indicated hole or a False can be considered True when it has three or four correctly indicated nearest neighbours.

However, there are two exceptions [103]:

1. A True that has only two nearest neighbours (at the edge of the phantom) requires only one correctly indicated nearest neighbour to remain True.
2. If both nearest neighbours are correctly indicated for a False or for a non-indicated hole that had only two nearest neighbours, it will be regarded as True.

4.1.2.3 The results depictions

There are two methods for presenting the results: 1) the use of formulas and 2) the use of the contrast detail (CD) curve[103].
4.1.2.3.1 Formulas

The CD curve is the curve through the threshold fields. The IQ can be illustrated by the ratio calculation of correctly identified hole positions to the total number of squares i.e. 225 squares inside the CDRAD phantom, as follows [103]:

\[
\text{Correct observation ratio} = \left( \frac{\text{Correct observation}}{\text{Total number of squares}} \right) \times 100 \% \tag{4.1}
\]

The IQF method can also be used to quantify the IQ. This method is defined in formula 4.2, as follows [103]:

\[
\text{IQF} = \sum_{i=1}^{15} C_i \cdot D_{i,\text{th}} \tag{4.2} \text{ (similar to equation 1.2)}
\]

Where \( D_{i,\text{th}} \) denotes the threshold \((th)\) diameter in the contrast column \( C_i \). A summation over all contrast columns yields the IQF.

Two extra rules are applied for calculation purposes [103]:

1) For a hole depth between 0.3 and 8 mm, a completely invisible column will result in a \( D_{i,\text{th}} \) of 10 mm.

2) For a hole depth between 0.3 and 8 mm, a completely visible column will result in a \( D_{i,\text{th}} \) of 0.3 mm.

The increase in the number of correctly identified hole positions reflects the IQ enhancement. The smaller the appearance of the contrast and the details, the better the system is. In this
case, because the values for the depths and diameters of the thresholds are smaller, the IQF will become smaller as well. However, as this could be misleading, an inverse measurement, IQFinv, is adopted [3]. IQFinv values increase as smaller diameter holes are observed. Formula 4.3 presents IQFinv approach for increasing the IQ, as follows [103]:

\[
\text{IQFinv} = \frac{100}{\sum_{i=1}^{15} c_i \cdot D_{i,th}}
\]  
(4.3) (similar to equation 2.7)

4.1.2.3.2 Contrast detail (CD) curve

The results can be demonstrated in a graph that depicts the hole depths plotted against the hole diameters. To compare the image performances of different systems, the phantom images should be produced for the same observer in identical conditions and at the same time. The smaller the appearance of the contrast and the details, the better the system is. This generates a shift of the CD curve to the lower-left part of the image. It also makes it possible to compare several performances [103].

4.1.2.4 Analysing of the CDRAD

Analysing the CDRAD involves analysing the images and providing a statistical method to determine whether a specific CD combination is observed or not. This statistical method utilises the standard deviation and the average pixel signal value in relation to the images of the CD combination under evaluation and its pixel background (or variables). It is vital to identify the locations of all 225 different CD combinations of the image phantom, which helps to correctly identify the two variables [103]. First, the programme identifies the location of the CD combinations. This achieved by applying a statistical method that
indicates CD combination recognition. Then, the programme depicts the user’s results. The assessment of a single CDRAD image can be broken into seven steps [103]:

1) Deciding the location of the phantom.
2) Deciding the centre of the 225 CD combination of the phantom.
3) Resolving the pattern type of the phantom.
4) Resolving the signal background.
5) Resolving the related CD signal.
6) Allocating True and False.
7) Calculating the contrast detail curve.

4.2 Computed tomography contrast-detail phantom (CTCDP)

This phantom is made of acrylic plates i.e. (Perspex: polymethyl methacrylate) with holes drilled in it, such that they vary diagonally in diameter from 1 mm to 12.5 mm (used in Chapter 9 and 10). They increase in diameter incrementally from the centre towards the edge of the cylindrical phantom. The diameters selected are; 1, 2.5, 5, 6, 7.5, 9, 10, 11, and 12.5 as displayed in Figure 4.1. These diameters are selected to test the contrast detection abilities of the CT scanners at different spatial resolution limits from low (1 mm) and up to high (12.5 mm). This enables the testing of a given CT scanner’s ability to detect contrast variations with large dynamic scale of spatial resolution.

The cylindrical phantom has a depth of 20 cm and a diameter of 30 cm. The phantom is scanned across its depth. The holes spread across about 4 cm along its depth in proximity to its middle section (see Figure 4.2). The holes can be filled with various concentrations of contrast media to represent different x-ray attenuation conditions depending on the conditions.
required for testing of the CT scanner’s abilities. Each radial line of holes extending from the centre to the edge of the phantom is filled with a given concentration of contrast media. These holes represent small and large objects to be imaged under different conditions and the concentrations of contrast media to water can be modified to assess very low and high subject contrasts. An imaging slice through this phantom across the hole’s area will evaluate a scanner’s ability to faithfully image objects of extremely low contrast with very low spatial resolution through to high spatial resolution in various contrast and resolution conditions.

Figure 4.1: The image depicts a section in the phantom showing the distribution of the holes in the Perspex. The blue inner circle highlights the central points that extend to the edge of the phantom. Each hole in a given ray will be filled to a certain concentration of contrast media. The red ellipse encompasses a ray of such holes indicating their size variations increasing outwardly up to 12.5 mm.
4.3 Employment of the information loss theory in CDP

The CDPs can be used to evaluate the degradation of IQ by quantifying un-detected holes, i.e. small diameters in the CDRAD phantom [7]. The IL concept can be applied to measure the amount of information that cannot be seen by the observers (used in Chapter 6, 7 and 10). IL quantification is based on information theory, which was established by Shannon in 1948 [25]. Entropy is the key measure, and the predicated random different-value uncertainty can be measured by it [25]. Information entropy is discussed in detail in (Chapter 2, Section 2.8.2). This theory can be applied to measure the IL in medical images using CDPs as follows [7]:

\[ IL = H(\text{no information loss}) - H(X) \]  

(4.4) (similar to equation 1.4)
The information content $H(x)$ is defined in the information theory as follows [7]:

$$H(x) = \sum p(x) \log_2 \left( \frac{1}{p(x)} \right)$$  \hspace{1cm} (4.5) \text{(similar to equation 1.3)}

Where: $p(x)$ is the probability. The probability $p(x)$ is discussed in detail in (Chapters 6, Sections 6.3.2.3 and 6.3.2.4).

$H(x)$: The information content.

The TIL assessed for different diameters of the phantom can be defined as follows:

$$\text{TIL} = \sum_{d}^{a} IL_{d}$$  \hspace{1cm} (4.6)

Where: $(a)$ is the largest diameter in the CDPs, $d$ is the smallest diameter.

$IL_{d}$: is the IL for each diameter.

The results from Niimi et al. [7] study showed that the IL increased as the diameter of the phantom holes decreased. Another study was done by the same group using the same method of measuring IL. In the latter experiment, however, they used a mammography CDP to obtain phantom images with a digital mammography system, and the results were the same. [26]. To quantify the performances of imaging modalities e.g. DR, CR and CT, the IL theory was employed to the CDRAD phantom and the specially-designed CTCDP in this research.
4.4 The grid applications in radiographic images

The main principle of X-ray projections depends on the travel of X-ray photons in straight lines. However, when X-ray scattering occurs in the patient, the resulting scattered X-rays are not aligned with the trajectory of the primary X-rays [2]. Therefore, the assumption that X-rays travel in straight lines is violated. These scattered radiations have no influence on an image if they do not reach the detectors. However, they may have an effect on the patient when the patient receives some of the scattered radiation. Moreover, the scattered radiation can cause image degradation if the image receptor detects this radiation. Thus, scattered radiation can decrease the contrast in a screen film radiograph by producing an unnecessary grey scale in an image [2]. In digital images, the contrast can be controlled and adjusted through the window or through levelling; hence, the scattered radiation can lower the signal to noise ratio SNR and act as a source of noise [2]. The scatter fraction, or the scatter-to-primary ratio (SPR), refers to the amount of scattered radiation detected in the image. It refers to the amount of energy deposited in a certain location in the image receptor by the scattered photons, divided by the amount of energy deposited by the non-scattered or primary photons for the same position, as follows [2]:

\[
SPR = \frac{S}{P}
\] (4.7)

For example, an SPR of 1 indicates that the amount of energy placed on the detector at the given location from scatter radiation represents 50 percent of the image information, making
the image useless. The amount of scatter can be calculated by the scatter fraction, as follows [2]:

\[ F = \frac{S}{P+S} \] \hspace{1cm} (4.8)

4.4.1 Anti-scatter grid

The most common method to decrease scatter radiation in radiography, is to use an anti-scatter grid, which is positioned between the patient and the detectors [2] (used in Chapter 7 and 8). This grid allows most of the primary radiation to pass through, while eliminating all of the scattered radiation. It has a simple geometric design that consists of grid septa and interspace regions; these components are aligned with the X-ray tube’s focal spot [2]. This alignment permits the primary radiation from the focal spot to proceed to the detector, while the scattered (or more oblique) photons are absorbed by the grid septa. The grid’s performance can be reduced, which may lead to the formation of artefacts when an error exists in the alignment. The anti-scatter grid is characterised by several parameters, such as the grid ratio, the interspace material, the grid frequency and the focal length [2].

The grid ratio is the ratio of the interspace material height to its width. The septa dimension has no effect on this ratio. Generally, in diagnostic radiology, the grid ratio values are 6, 8, 10, 12 or 14 [2]. In the grid constructor, the grid ratio is the most essential parameter, and the grid septa are mainly manufactured based on this lead. Even though the ideal interspace material is air, the lead septa demand a supportive structure. Thus, to maintain septa alignment, it is necessary to place a solid material in the typical linear grid used in general
radiography. Therefore, manufacturers use carbon fibres in the interspace, as carbon has a low atomic number that allows for great primary transmission due to its low density[2].

The grid frequency is the number of grid septa per centimetre. For example, in Figure 4.3, the width of the septa is 0.045 mm and the width of the interspace is 0.12 mm; hence, the line pattern space is 0.165 mm. Therefore, the frequency is 1/0.165 mm, which equals 6 lines/mm or 60 lines/cm. As an alternate to the grid motion for an imaging system with a discrete detector element, a stationary, high-frequency grid can be used. For instance, the detector element in chest radiography represents about 200 μm; hence, the grid septa, which are 45 μm wide, must be obscured, since the bars of the grid are much smaller than the spatial resolution of the detectors [2].

![Figure 4.3: The grid is manufactured from layers of septa material and interspace material. This image shows parallel grid septa; however, in reality, the septa and the interspace are slightly angled for focussing[2].](image-url)

The focal length is vital for focussing the grid to enhance the alignment between the X-ray source and the interspace regions of the anti-scatter grid. The focal length in most radiographic cases is 100 cm, while upright chest imaging has a focal length of 183 cm.
grid cut-off is produced when the grid is located at a different distance from the X-ray tube [2].

### 4.5 Computed Tomography Dose Index CTDI

Despite being designed as an index and not for patient dose assessment, the CTDI became the most accurate method for measuring doses following a modification to its original concept [2] (used in Chapter 11). Currently, CTDI-based dosimetry is the global standard for predicting patient doses in CT experiments. This method is specified by an “index” to differentiate the amounts of radiation doses absorbed by patients. The basic $CTDI_{100}$ uses a PMMA phantom that has a peripheral or centre hole. These holes are inserted through a 9 mm-diameter cylindrical (pencil) chamber with a length of 100 mm [2]. There are two types of standard PMMA phantoms that are used for dosimetry: the head phantom, which is 15 cm long, has a diameter of 16 cm and can be utilised as a paediatric torso phantom, and the body phantom, which is 15 cm long with a diameter of 32 cm (Figure 4.4). In the centre of the phantom, there is a pencil chamber, which is located in the z-dimension in the centre of the CT gantry, where a single axial CT scan is created. To generate accurate dose estimations using an ionization chamber, the total sensitive volume must be irradiated using an X-ray beam. Thus, the nominal beam width (or the total collimated beam width) for the partially irradiated 100 mm CT pencil chamber is exploited to achieve precise chamber reading for partial volume exposure. A chamber length of 100 mm is valuable for different slice thickness, from thin slices (e.g. 5 mm) to thicker beams (e.g. 40 mm). It is vital to correct the partial volume exposure [2, 104], as follows:

$$K_{corrected} = \frac{100 \text{ mm}}{B} K_{measured}$$  \hspace{1cm} (4.9)

Where $B$ is the total collimated beam width in mm (for an individual axial scan).
The B value is formed from the number of active detectors (n) and from the CT detector width projected for the iso-centre of the scanner (T). Hence, $B = nT$. For instance, if the channels of the CT scanner were $n = 64$ and if the measurement of detector channels was 0.625, $B = 64 \times 0.625 \text{ mm} = 40 \text{ mm}$. The equation for $CTDI_{100}$ is, therefore [2]:

$$CTDI_{100} = \frac{1}{nT} \int_{L=-50 \text{ mm}}^{+50 \text{ mm}} D(z)dz$$  \hspace{1cm} (4.10)

The previous equation depicts the calculation of the dose distribution from a single circular ($z$) axis associated with a normal beam width of $nT$. The scattered and primary radiation are measured over a 100 mm length, and the X-ray beam centre is set at $z = 0$ [105]. In most scanner machines, the $CTDI_{vol}$ can be displayed on the CT scanner console before the actual scan. This is because the measurement for $CTDI_{vol}$ is acquired by the CT manufacturers, and the calculation considers the kV ranges. Moreover, the kV values are scaled with the pitch and mAs of the machine; hence, they are displayed in the CT console. The $CTDI_{vol}$ product and the CT scan length along the Z-axis of patient (L) describe the dose length product (DLP), as follows [2]:

$$DLP = CTDI_{vol} \times L$$ \hspace{1cm} (4.11)

$CTDI_{vol}$ is a very useful method for comparing the doses delivered by different scan protocols, and it allows one to obtain a specific IQ level for a specific-sized patient. Prescribing the right dose for a specific patient size and for specific diagnostic tasks is
possible to achieve for $CTDI_{vol}$ using the technique chart and the diagnostic reference level. However, the DLP values ($CTDI_{vol}$ derivative) cannot estimate the potential cancer risks or effective doses for individuals [105].

Figure 4.4: The measurement of CTDI is acquired by a 16 cm- or 32 cm-diameter PMMA phantom, and the dosimeter is placed serially in both the centre hole and the peripheral hole [2].

### 4.6 Radiographic contrast media

Radiographic contrast media can be classified into positive and negative contrast agents. Positive contrast media provide better attenuation of X-rays than body soft tissue, and they can be divided into water-soluble iodine-based agents and non-water-soluble barium agents (used in Chapter 5, 9 and 10). The attenuation of the negative contrast agents to X-rays is low compared to soft body tissue, and it is not available commercially. In conventional radiography and CT, water-soluble iodine-based contrast agents diffuse throughout the extracellular space [106]. Blood vessels cannot be visualised without the administration of contrast agents. The enhancement of the attenuation in relation to the surrounding tissues is
obtained by introducing materials that are optimised through the energy dependence of photoelectric absorption, such as the K-edge of iodine at 33.2 keV [106].

However, the reduction of attenuation in relation to surrounding tissues is achieved through the use of a CO₂ gas (due to its low mass density) [4]. The iodine contrast agent is formed by a benzene ring, which is attached to three iodine atoms. A monomer consists of one tri-iodinated benzene ring, while a dimer contains two tri-iodinated benzene rings. The iodine contrast agents are divided into ionic and non-ionic agents according to their water solubility [106]. The ionic iodine has the ability to dissociate into negative and positive ions and then to attract the negative and positive poles of water molecules, while the non-ionic contrast reduces the water-soluble according to their polar OH groups. The main physical features of iodine contrast agents are osmolality and viscosity. The osmolality of contrast agents has a major impact on patients’ side effects, especially in cases higher than 800 mosm/kg [106]. The contrast agents are classified into high, low and iso-osmolar agents (Table 4.1). Their viscosities are a function of the molecular shape, solution concentration and the weak interactions between the contrast agents and the water molecules. When moving from ionic to non-ionic agents, the viscosity increases and the osmolality and toxicity decrease [106].
Table 4.1: Iodine-based contrast agents, including trade name, class and g-Iodine/ml [106].

<table>
<thead>
<tr>
<th>Contrast agent</th>
<th>Trade name</th>
<th>Structure</th>
<th>Charge</th>
<th>Class</th>
<th>Maximum g-Iodine/ml</th>
</tr>
</thead>
<tbody>
<tr>
<td>Diatrizoate</td>
<td>Renografin, Hypaque</td>
<td>Monomer</td>
<td>Ionic</td>
<td>HOCM</td>
<td>358-370</td>
</tr>
<tr>
<td>Amidotrizoate</td>
<td>Urografin</td>
<td>Monomer</td>
<td>Ionic</td>
<td>HOCM</td>
<td>300</td>
</tr>
<tr>
<td>Isothalamate</td>
<td>Conray</td>
<td>Monomer</td>
<td>Ionic</td>
<td>HOCM</td>
<td>370</td>
</tr>
<tr>
<td>Ioxithalamate</td>
<td>Telebrix</td>
<td>Monomer</td>
<td>Ionic</td>
<td>HOCM</td>
<td>350</td>
</tr>
<tr>
<td>Ioxaglate</td>
<td>Hexabrix</td>
<td>Dimer</td>
<td>Ionic</td>
<td>LOCM</td>
<td>320</td>
</tr>
<tr>
<td>Iopamidol</td>
<td>Iopamiro, Isovue</td>
<td>Monomer</td>
<td>Non-Ionic</td>
<td>LOCM</td>
<td>370</td>
</tr>
<tr>
<td>Iohexol</td>
<td>Omnipaque</td>
<td>Monomer</td>
<td>Non-Ionic</td>
<td>LOCM</td>
<td>350</td>
</tr>
<tr>
<td>Iomeprol</td>
<td>Iomeron, Imeron</td>
<td>Monomer</td>
<td>Non-Ionic</td>
<td>LOCM</td>
<td>400</td>
</tr>
<tr>
<td>Iopentol</td>
<td>Imagopaque</td>
<td>Monomer</td>
<td>Non-Ionic</td>
<td>LOCM</td>
<td>300</td>
</tr>
<tr>
<td>Ioxilan</td>
<td>Oxilan</td>
<td>Monomer</td>
<td>Non-Ionic</td>
<td>LOCM</td>
<td>350</td>
</tr>
<tr>
<td>Ioversol</td>
<td>Optiray</td>
<td>Monomer</td>
<td>Non-Ionic</td>
<td>LOCM</td>
<td>350</td>
</tr>
<tr>
<td>Iopromide</td>
<td>Ultravist</td>
<td>Monomer</td>
<td>Non-Ionic</td>
<td>LOCM</td>
<td>370</td>
</tr>
<tr>
<td>Iotrolan</td>
<td>Isovist</td>
<td>Dimer</td>
<td>Non-Ionic</td>
<td>IOCM</td>
<td>320</td>
</tr>
<tr>
<td>Iodixanol</td>
<td>Visipaque</td>
<td>Dimer</td>
<td>Non-Ionic</td>
<td>IOCM</td>
<td>320</td>
</tr>
</tbody>
</table>

*HOCM High-osmolar contrast media, LOCM Low-osmolar contrast media, IOCM Iso-osmolar contrast media*

### 4.7 Radiochromic films

Radiochromic films are composed of a thin polyester base saturated with radiation-sensitive organic microcrystal monomers (used in Chapter 11). In the presence of ionizing radiation, the film emulsion experiences a colour change within 24 hours due to either a chemical reaction or polymerization. Then, the radiation response signals can be read by measuring the optical density or absorbance change using a spectrometer at specific wavelengths. The polydiacetylene-based GafChromic films are currently the most popular; these are designed by international specialty products (ISPs) [107]. Lewis et al. [108] introduced the first GafChromic films, which were later developed by McLaughlin et al. [109] at the National Institute of Standards and Technology (NIST) in the United States. The following table illustrates some current commercial GafChromic films:
GafChromic films provide several advantages in medical radiography, such as accurate and precise dose measurements, outstanding spatial resolution (> 1200 lines/mm) [107], easy handling and weak energy dependence. Unlike silver-based film emulsions, GafChromic films have good energy dependence [107]. Butson et al. [110] investigated the energy dependence of GafChromic EBT2 films’ dose responses in the X-ray range of 50 kVp to 10 MVp and found a 6.5 ± 1 percent difference in the optical density to absorbed dose response.

In some medical applications, the energy independence of radiochromic films is beneficial, particularly when a broad range of radiation sources is being delivered [107]. Moreover, radiochromic films do not require subsequent chemical processing (unlike photographic films); hence, they can generate repetitive responses over a relatively long time period. However, environmental factors like humidity, temperature and UV exposure in irradiation and readout processing have potential impacts on radiochromic films. Thus, it is suggested that GafChromic films must be stored at a temperature of approximately 22°C and in a dark place to eliminate any environmental effects [107].
4.8 Image J software

Image J is software for digital imaging that has the ability to manipulate and process images for analysis purposes (used in Chapter 11). To manipulate images, the software offers a very convenient user interface, which is associated with a large number of readily available functions and tools for working with images interactively (Figure 4.5). The digital image processing promotes comprehensive and well-documented software libraries that assist in the implementation of new image-processing algorithms and working and prototype applications. Image J provides a set of ready-made tools for the interactive manipulating and viewing of images [111]. Moreover, the software can be easily extended by writing new software components in standard programming languages. The software can be run on any operating system (e.g. MacOS or Windows) because it is implemented completely in Java. Java’s dynamic execution model permits new modules, or “plugins”, to be written independently of Java codes. These plugins can be loaded, compiled and executed “on the fly” in the running system without restarting the Image J software. This makes Image J suitable for testing and developing new algorithms and image processing techniques (i.e. an ideal platform) [111]. Moreover, the software is freely available and requires no license for installation; hence, it is accessible to students and instructors. The utilisation of Image J involves serious research and application development, especially in medical imaging and biological studies. The key features of the Image J software can be summarised as follows [111]:

1. Interactive tools: A set of ready-to-use tools for editing, visualising, creating, analysing, processing and sorting images. The software can also support several common file formats. Image J can also provide 32-bit floating-point images and “deep” 16-bit integer images and image sequences (“stacks”).

2. A simple plugin mechanism can be used to extend the core functionality of Image J by writing small pieces of Java code.
3. Image J offers a macro language associated with the corresponding interpreter. This facilitates the implementation of larger processing blocks through the combination of existing functions without Java’s knowledge.

![Image J features diagram]

Figure 4.5: The main window of Image J displayed on a Windows operating system [111].

4.9 **Statistical Analyses**

The data presented in all chapters (mean ± standard deviation) was the result of three or ten independent measurements as indicated in each chapter. One way analysis of variance (ANOVA) or two-tailed student’s t-test were used to determine the significance of the difference between the control and experimental group. A difference was considered to be statistically significant when p<0.05.
Chapters 5 to 11 represent the individual experiments of the thesis including results and discussions.
Chapter 5 Low-Contrast Detail Phantom: Proof of Concept

5.1 Summary

The aim of this chapter is to investigate the concept of replacing the air gaps of the conventional contrast detail phantom (CDP) with various concentrations of contrast media, and to develop a variable level of attenuation level differential phantoms that could be more appropriate for contrast measurements in some radiology cases.

Images were acquired using the digital radiography system of the conventional CDP (Perspex-air hole phantom) and the novel form of CDP where the air holes were replaced with attenuating material. In this study, two different attenuating materials were introduced, water and a 30% concentration of iodine based contrast medium. Image quality was assessed using automated processing to calculate the image quality factor (IQF)_{inv}.

Phantom studies indicate that lower contrast levels are obtained when CDP holes are filled with water and a 30% concentration of iodine contrast media than those observed for air-Perspex or conventional CDP. As an example, when a 5 mAs beam is used the IQF_{inv} values are 5.32 in the case of air filling the holes; however, when these holes are filled with water under the same conditions, the value of the IQF_{inv} drops to
2.55, and to 2.83 when 30% of contrast media is used. Other concentrations were also tested. These results indicate that it is possible to extend the contrast scale in these phantoms to include ranges that are more realistic for a patient’s body than just air and tissue-equivalent material.

These findings indicate that the proposed extension of the contrast scales allows smaller changes in contrast to be discerned. This is due to the small attenuation differences of the subject materials (e.g. 30% contrast liquid and wax) from the conventional form of CDP (material/air). This suggests that the low form of the CDP may have a useful role in assessing image quality in planar radiology as an evaluation tool to better represent low subject contrast imaging requirements [112].

5.2 Introduction

The assessment of image quality in radiologic imaging is a vital process for ensuring that high quality images are provided, thereby enhancing diagnostic ability. Improvements in radiologic image quality commonly involve increasing patient and public dose. For example, in most cases, more X-ray photons are required to improve statistics and reduce the noise in an image, but this means an increased radiation dose [2, 12, 113]. In accordance with the As Low As Reasonably Achievable (ALARA) principle, to keep the radiation dose as small as possible while still providing adequate image quality to enable diagnosis, the dose should be optimised with image quality [9].
The assessment of image quality can use objective or subjective methods. Objective assessments of image quality associated with diagnostic imaging systems can be equipment based, such as noise analysis or modulation transfer functions [2]. These methods, while providing assessment of radiation detector performance, do not consider the effects of the image assessor typically a radiologist and the effects of the viewing system and conditions. Image quality can also be assessed subjectively. In this method, radiologists typically subjectively assess and judge the diagnostic images, with receiver operator characteristics used to compare the performance of various imaging systems [9]. Although subjective assessment of image quality using receiver operator characteristics analysis considers the whole imaging chain (equipment, scatter, image processing, and human observer), it is a time-consuming process and cannot be readily adopted as a quality assurance method in busy clinical practices. A commonly adopted alternative approach to assessing image quality is the use of contrast detail phantoms (CDPs) [9, 16, 18]. CDPs are designed to provide useful information on contrast detail detectability and have been shown to be one of the most reliable and commonly adopted phantoms for image quality assessments, especially in low-contrast conditions [114]. In fact, CDPs are commonly referred to as low contrast detail phantoms. Although there are many image quality reporting tools, one commercially available CDP, the CDRAD 2.0 (Artinis Medical Systems, Netherland) phantom, is the most commonly used CDP [16, 103, 114]. The CDRAD phantom is made of acrylic (Perspex; polymethyl methacrylate) 10 mm thick, in which 225 cylindrical holes of various sizes and depths are drilled in the phantom medium. The diameter of the holes varies in size from 0.3 mm to 8 mm. This range is equally distributed across 15 depths. These depths range from 8 mm (providing high contrast) to 0.3 mm (providing low contrast). Hence, the CDRAD phantom uses air–acrylic interface to create image contrast. However, the subject contrast of the CDRAD
phantom is relatively high because it represents attenuation differences between Perspex and air. This chapter aims to replace air by filling the holes with material of a slightly different attenuating ability than that of Perspex, representing a much lower contrast measuring phantom than the conventional CDP.

The rationale for this chapter stems from the potential of extending the application conditions of the current type of commercially available CDP. Air and acrylic in current CDPs are the only materials used to create various amounts of image contrast for the assessment of the contrast detectability of the imaging system. In this study, the air is replaced with water, a substance equivalent to tissue. The phantom structure modification introduced in this study represents the assessment of the ability of the imaging system to discern smaller subject contrast differences with water–acrylic rather than air-acrylic.

Different scales of contrast (compared with the current air based CDP) are also investigated in this study by filling the holes with various concentrations of contrast media. Hence, this study also examines the feasibility of including contrast media into water and using this phantom as a multiscale contrast measuring device. This chapter reports on the use of this new format for the CDP with images acquired using a commercially available flat panel detector digital radiography (DR) system.

### 5.3 Materials and Methods

#### 5.3.1 Materials

Modified CDPs were developed by adapting the commercially available CDRAD phantom:
the air in the holes was replaced with a medium that has absorption characteristics similar to the base material acrylic, hence reducing the subject contrast. First, the air filled holes within the CDRAD phantom were filled fully with distilled water. In this instance, replacing the air in the holes with water is the only physical modification of the CDRAD. By replacing air with distilled water, the CDP now represents the difference between water and Perspex rather than Perspex and air. Subject contrast in the phantom is reduced when the holes are filled with water instead of air. The 225 water-filled holes of depths and sizes varying from 0.3 mm to 8 mm will exhibit attenuation characteristics closer to that of acrylic and not exhibit the larger range of contrast in the air filled CDRAD phantom. This modified CDP now more closely replicates the low subject contrast commonly encountered in non-contrast radiology where attenuation between adjacent soft tissues in the human body is very similar. This modified CDP is essentially a “low-contrast” CDP.

Second, the air filled holes within the CDRAD phantom were filled with a 30% concentration of iodinated contrast media (Omnipaque 350; GE Healthcare) mixed with distilled water. The 30% concentration of iodinated contrast media was selected to simulate the common case of contrast injection into patients. The introduction of iodinated contrast media creates CDPs that now more closely replicate the attenuation encountered in contrast radiology.

The physical characteristics of the three compounds involved in this study (air, water, and Perspex) are listed in Table 5.1. An Agfa DX-D 600 digital flat panel system (Agfa HealthCare, Scoresby, Australia), a flat panel direct radiography system using cesium iodide DR
detector technology, was used to acquire images of the standard CDRAD phantom and the two novel CDPs.

5.3.2 Methods

To simulate clinical imaging, images were acquired with five sheets of scattering medium (Perspex square 265 x 265 mm with thickness of 10 mm) on top and another five sheets of scattering medium placed below the CDP [16]. The source-to-detector distance (SDD) was set to 100 cm. The X-ray tube voltage (kVp) was set to 70 kVp and X-ray tube current (mAs) values of 5, 10, and 20 were set for each experimental condition of exposure of the three CDPs and Perspex. The CDPs and Perspex sheets were positioned directly on the table top. The detector was placed in the table Bucky (grid 8:1) assembly. This condition was preferred because grids are recommended for use in DR systems with body part thickness 10 cm or greater [43]. Images acquired were processed using the clinical algorithm of abdomen, as this algorithm best represents low subject contrast radiology examination.

To compare image quality parameters of the standard CDRAD and the two modified CDPs, image quality factor (IQF) was determined (calculated) using CDRAD Analyzer software. The CDRAD Analyser was designed to determine the IQF for an air-Perspex test phantom. After communication with the supplier, the supplier provided a modified algorithm to fit the application for CDRAD when filled with water or diluted contrast media (M. Floor, personal communication, November 12, 2013).
The IQF is an algorithm developed by the CDRAD phantom manufacturer, and defined as:

\[
IQF = \sum_{i=1}^{15} C_i D_{i,j}
\] (5.1) (similar to equation 1.2)

Where \(D_{i,j}\) represents the threshold (j) diameter in contrast column “Ci”. The summation is over all the columns.

The values of the IQF (related to image quality) decrease as smaller diameter holes are observed. However, as this could be misleading, an inverse measurement, \(IQF_{inv}\), is adopted [16]. \(IQF_{inv}\) values increase as smaller diameter holes are observed. The \(IQF_{inv}\) is as follows:

\[
IQF_{inv} = \frac{100}{\sum_{i=1}^{15} C_i D_{i,j}}
\] (5.2)(similar to equation 2.7)

Table 5.1: Physical characteristics of the three compounds: water, Perspex and air, involved in this study.

<table>
<thead>
<tr>
<th></th>
<th>Water</th>
<th>Perspex</th>
<th>Air</th>
</tr>
</thead>
<tbody>
<tr>
<td>Density (g/cm³)</td>
<td>1.0</td>
<td>1.18</td>
<td>0.0012</td>
</tr>
<tr>
<td>Effective atomic number</td>
<td>7.4</td>
<td>6.6</td>
<td>7.67</td>
</tr>
</tbody>
</table>
5.4 Results and Discussion

Representative images acquired experimentally of the conventional CDP and the two modified CDPs are provided in Figures 5.1–5.3. The IQF_{inv} for each experimental condition is displayed in Table 5.2.

First, by comparing the conventional CDP (i.e. commercially available) (Figure 5.1) with the CDP with holes filled with water (Figure 5.2), it is seen both visually from the images and from IQF_{inv} values (Table 5.2) that lower contrast levels are obtained when CDP holes are filled with water than those observed for air-Perspex or conventional CDP. For example, when comparing the top right hand corner of the image, water filled holes are closer in brightness to the surrounding Perspex (Figure 5.2) than is observed between air filled holes and surrounding Perspex. This is also reflected in IQF_{inv} value for water CDP (Table 5.2) being consistently lower than that of air CDP. This is attributed to the smaller attenuation differences between water and acrylic than the air and acrylic.

The air and Perspex combination provides a higher contrast combination of objects. The subject contrast in the case of water-Perspex is lower than that of air-Perspex. The air-Perspex interface represents high subject contrast because air attenuates much less radiation compared with other media, including Perspex. We describe the air-Perspex format of the CDP used here as a conventional CDP. In contrast, by converting the conventional, commercially available CDP into a low subject contrast form by adding water into each hole, the phantom more closely resembles the real case of patients imaged in radiology. Although in practice it is easier to use the conventional CDP with air in its holes, this does not resemble the real patient’s body. Moreover, the larger difference between the contrasts of
Perspex compared with air limits its use to only relatively high contrast cases, although such phantoms are mistakenly termed low-contrast detail phantoms.

The scientific basis for the smaller differences in attenuation, and hence less contrast between liquid and acrylic compared with that between air and acrylic, is presented. The effective atomic numbers and the physical densities were selected since they influence the radiation absorption characteristics of these compounds.
Figure 5.1: (A) Conventional contrast-detail phantom with air in holes and (B) contrast detail score diagram and curve and IQF\textsuperscript{inv}. 
Figure 5.2: (A) Low contrast-detail phantom whereby the holes are filled with tissue equivalent medium “water” and (B) contrast detail score diagram and curve and $IQF_{inv}$.

Figure 5.3: CDRAD phantom image when holes filled of Iodine 30% concentration
The density of the free electrons does not vary significantly among these compounds. The probability of occurrence of the photoelectric effect depends on the effective atomic number strongly and linearly on the physical density of the target, whereas the occurrence probability for the Compton interaction depends on the density of the free electrons and the physical density of the targets. Accordingly, the likelihood of absorption and scatter of the radiation off these compounds is greatly influenced by their physical densities, as this parameter directly affects both photoelectric effect absorption and Compton scatter. In addition, because the effective atomic numbers of these compounds are only slightly different, the radiation effects will be governed mainly by the physical density of the compounds. It is therefore expected that there is much less absorption of X-rays through air in comparison to Perspex (a high-contrast combination), whereas there will be very close levels of absorption in Perspex and water (a low-contrast combination). If the aim is to use the phantom as a low contrast measuring device, it is better to fill the holes with water rather than air. It should be noted from Table 5.2 that the improvement in IQF value as mAs changes from 10 to 20 in the case of medium-filling holes (both water and 30% iodine) is slightly less than the change observed in case of air filled holes [3].

Table 5.2: IQF values for different media filling the holes of a CDRAD phantom obtained with 10 cm of scattering medium at 70 kVp for three mAs values. The data presented in this table (mean values ± standard deviation) was the result of three independent measurements (n=3). The data means of water and 30% concentration of iodine contrast medium are significantly different from the air phantom (**p<0.01, two-tailed Student’s t-test)
Another advantage of filling the holes with water to generate a low contrast detail phantom is that the scale of the contrast measurements can be varied and extended by adding different amounts (concentrations) of iodine based contrast media into the holes. This extends the low CDP so that it may be used as a specified CDP. Figure 5.3 displays images of the low CDP whereby a 30% concentration of contrast medium has replaced air in the holes. These modified versions of the CDP offer alternatives to the fixed contrast in the conventional CDP between the two media; that is, air-Perspex. Other alternatives that have been used in this project could be considered depending on the clinical conditions being evaluated (e.g. different concentrations of the iodine in the holes could be used).

In addition, the adaptation of the conventional, commercially available CDP into a modified CDP format enables this type of CDP to be used to test or validate the performance of new types of contrast media (e.g. gold nanoparticles, Bi2S3, and others) recently included or under investigation for inclusion[115, 116]. Such contrast media can

<table>
<thead>
<tr>
<th>Type of contrast media</th>
<th>IQF&lt;sub&gt;inv&lt;/sub&gt; values</th>
<th>5 mAs</th>
<th>10 mAs</th>
<th>20 mAs</th>
</tr>
</thead>
<tbody>
<tr>
<td>Air</td>
<td>5.32±0.78</td>
<td>6.97±0.40</td>
<td>7.57±0.54</td>
<td></td>
</tr>
<tr>
<td>Water</td>
<td>2.55±0.39**</td>
<td>3.01±0.35**</td>
<td>3.11±0.17**</td>
<td></td>
</tr>
<tr>
<td>30% concentration of iodine contrast medium</td>
<td>2.83±0.36**</td>
<td>3.56±0.06**</td>
<td>3.67±0.04**</td>
<td></td>
</tr>
</tbody>
</table>
also be used to fill the holes of the CDP. The phantom can be imaged under set conditions, and the IQF of the image measured. Comparing the IQF values of newer types of contrast media against traditional iodinated contrast media will be a clear indicator of subject contrast improvement.

5.5 Limitations
Introducing contrast medium into small holes within the CDP is problematic. As shown in Figure 5.3, contrast medium has not uniformly filled each hole. To overcome the technical issues associated with introducing a 30% concentration of iodine based contrast media, an alternate form of a novel CDP using three-dimensional printing to produce a phantom that has increasing thicknesses of attenuating material may be a more viable option, which researchers are currently investigating. Despite this acknowledged technical limitation, this study has provided proof of the concept that a novel form of the conventional CDP yields additional information for assessing image quality in planar radiology systems compared with the conventional CDP (air-Perspex).

5.6 Conclusion
The results presented in this study prove that the conventional form of the CDRAD can be extended to include various ranges of attenuating materials filling the holes instead of only using air. It has also been recently identified that conversion of a typical CDP into a low-contrast form has been considered highly valuable radiologically [117] . The findings from this chapter similarly suggest that using a modified form of the CDP for projection imaging systems will be a valuable addition to radiology departments. Extension of this line of research is to include the study of the effects of various concentrations of the
media filling holes and also the effects of kVp on the performance of CDP. The next stage of this research comprises steps of manufacturing further modified versions of CDP using three dimensional printing techniques.
Chapter 6 Information loss via visual assessment of radiology images using a modified version of the low-contrast detailed phantom in direct DR system

6.1 Summary

The previous chapter was clearly demonstrated the efficiency of using different contrast medium filling the air holes of the CDRAD phantom. This chapter further investigates the use of the modified CDRAD phantom with applying the information loss (IL) theory to evaluate the image quality. Quality in radiology images can be assessed by determining the levels of information retained or lost in an image. Information loss in images has been recently assessed via a method based on information theory and the employment of a contrast detail phantom (CDP). CDP is made of Perspex with holes of various sizes filled with air at atmospheric pressure. Such a phantom represents high-subject contrast due to the large difference in x-ray attenuation between the air and Perspex. In this Chapter, the conventional CDP (air-Perspex) and a modified CDP were used. In the modified CDP, the holes were filled with water (water-Perspex) to introduce an imaging condition where attenuation between subject materials is similar; i.e., there is low inherent subject contrast, which more realistically represents x-ray attenuation in human tissues such as the abdomen. The conventional and modified CDPs were used to determine the information loss in radiologic images.


6.2 Introduction

Digital radiography (DR) has been developed over the last decade by the advent of material science and information technology. DR imaging offers better spatial and contrast resolution compared to computed radiography (CR) imaging and, for this reason, has become the “new” standard in imaging institutions [5]. The flat panel detector in a DR system is characterised by a direct readout-matrix of electronic elements that are made of thin layers of amorphous silicon thin-film transistors (Si-TFT elements) placed on a sheet of glass. There are two types of DR detectors (depending on the material used):

1) Indirect conversion TFT detectors: These detectors use a scintillator, e.g. Caesium iodide (CsI-TFT), and light-sensitive photodiode. This technique is similar to the CR system, whereby absorbed radiation is transferred into light signals in the CsI layer. However, unlike CR, it provides minimal scattering of light along the silicon elements due to the needle type structure of the detector material. Applications of the CsI-TFT system include skeletal and chest radiography and is amenable to fluoroscopic procedures.

2) Direct conversion detectors: These detectors consist of condensator elements made up of amorphous selenium placed on the TFT array. The absorbed radiation energy is directly converted to a charge, avoiding the intermediate step of scintillation to promote conversion to visible light. This method is not amenable to fluoroscopic imaging because of the tendency to generate persistent latent images. Direct radiography is increasingly being used because patients are exposed to a lower radiation dose [5].

The superior spatial and contrast resolution provided by DR imaging enable accurate identification of small lesions, which may be concealed by other structures [2, 29]. When radiology imaging systems have similar imaging characteristics, quantification of image
quality provides additional information that can assist the selection of an appropriate system for clinical examinations. Many different methods are used to evaluate image quality in DR, including detective quantum efficiency (DQE)[50], information entropy [54], and the Rose model method [118]. The DQE determines the ability of the imaging system to transfer information from input stage to the output stage [60]. The information entropy is based on Shannon’s theory. It describes how much uncertainty or randomness there is in a signal or image [54]. The Rose model method attempts to define how the human observer can detect a flat-topped, sharp-edged signal of a specific area in uniform background containing Poisson noise [17]. In addition, some image quality evaluation methods rely on the observer, such as visual grading analysis (VGA) [61] and receiver operating characteristic (ROC) analysis [57]. VGA is a method that can be used to assess the fulfilment of the criteria in a scientific manner [60]. ROC measures the ability of the observer to detect lesions and possibly correctly interpret the visual signals in images [57].

Low contrast detail (LCD) detectability is used as a measure of image quality. The evaluation of radiology image quality contributes to the assessment of LCD [3]. There are two phantoms commonly used to detect LCD: Catphan phantom used in computed tomography and CDRAD phantom in DR systems [22, 119].

Contrast detail phantoms (CDPs) like CDRAD are used in the visual assessment of LCD and also can be used as a tool for maintenance/assessment of image quality [7]. This visual assessment is usually performed using an image quality factor (IQF). Information loss (IL) theory has also been applied to the visual assessment of CDP images. Niimi et al. [7] introduced the term total information loss (TIL), and demonstrated that TIL was lower when higher radiation dose was used, and when the diameters of the air-filled holes within the CDP
were larger. The CDP used in the study by Niimi et al. [7, 28] was air-Perspex which represents high subject contrast; ie, the x-ray attenuation difference between air and Perspex is relatively large. This relative large x-ray attenuation difference does not reflect low-subject contrast imaging that can occur in radiology such as abdomen and breast imaging. As such, using the conventional air-Perspex CDP may underestimate TIL for some clinical radiology examinations. The current study seeks to address this limitation by investigating TIL for low-subject contrast imaging phantom, Perspex-water, using the information loss theory to assess the image quality.

The aim of this Chapter was to employ the TIL to quantify the performance of two CDPs: one represents relatively high-subject contrast, air-Perspex CDP, and the other representing relatively low-subject contrast, water-Perspex CDP. As demonstrated by Geso et al.[3], use of a water-Perspex phantom better represents low-subject contrast, and is a valuable addition in radiology in assessing image quality.

6.3 Materials and Methods

6.3.1 Materials

The contrast-detail phantom CDRAD 2.0 (Artinis Medical Systems, Netherlands) was used in this study (Figure 6.3-a). This commercially available phantom is made of acrylic (Perspex; polymethyl methacrylate) of 10 mm thickness, in which 225 cylindrical holes of various sizes and at various depths is drilled into the phantom medium. The diameter of the holes varies in size from 0.3 mm to 8 mm. This range is equally distributed across fifteen depths. These depths range from 8 mm (providing high-subject contrast) to 0.3 mm (providing low-subject contrast). The CDRAD phantom is a conventional CDP as it uses an air-acrylic interface to
create image contrast [120]. The CDRAD phantom offers a difference in attenuation between phantom material and air only, which represents high-subject contrast. This form of CDP is unlikely to be representative of x-ray attenuation in the human body, except for lung fields. To overcome the inherent subject contrast limitation of the conventional CDP, Geso et al. [3] developed a modified low-contrast version of the commercially available CDP whereby the air filled holes within the CDRAD were filled with distilled water. By replacing air with distilled water, the phantom more closely replicated low contrast; i.e., less attenuation variations commonly encountered in non-contrast radiology where attenuation between adjacent soft tissues is very similar [12]. An Agfa DX-D 600 flat panel direct radiography system (Agfa Health-Care, Scoresby, Australia) using a caesium iodide DR detector technology, was used to acquire images of both the standard CDRAD phantom (air-Perspex) and the modified CDRAD (water-Perspex).

### 6.3.2 Methods:

#### 6.3.2.1 Test images

To simulate clinical imaging, images were acquired with 5 sheets of scattering medium (Perspex square 265×265 mm with thickness of 10mm) on top and another 5 sheets of scattering medium placed below the CDP [16]. The source to detector distance (SDD) was set to 100 cm. Using the Agfa DX-D 600 digital flat panel system, six phantom radiographs were acquired at 70 kVp and 20 mAs. Three x-ray images were acquired for each phantom.

In all testing conditions, the phantom and scattering medium (Perspex sheets) were positioned directly on the table top. The detector was placed in the table bucky (grid 8:1) assembly. This condition was adopted as grids are recommended for use with anatomical thickness of 10 cm or greater [43]. Images acquired were processed using the clinical
algorithm for imaging the abdomen, as this algorithm best represents low-subject contrast radiology examination.

6.3.2.2 Image scoring

Prior to data collection, informed consent was obtained from the participants through a certified human ethics approval document from RMIT University (ASEHAPP 62 see appendix VIII). For the sample forms of the invitation and consent form for participants and CDRAD phantom scoring sheet (see appendices IV, V and VII).

Nine radiographers, two medical physicists, and three radiologists were invited to participate in scoring of all acquired images. Each observer read the phantom images independently, using a medical grade monochrome LCD display. The room lights were turned off and all observers were asked to identify the faintest visible cylinder in the CDRAD image for each diameter, ie, the smallest cylinder that is visible on each column (see Figure 6.3a). Using the scoring method described by Niimi et al. [7], the faintest visible disc for each diameter was then used as an index of threshold contrast. The indices were then averaged for the 14 observers in this study and plotted against the different diameters of the CDRAD phantom. These plots are referred to as “Contrast-Detail (CD) diagrams”. An IQF was also calculated. It is given by:

\[
IQF = \sum_{i=1}^{n=15} C_i D_{i,j} \quad \text{(6.1) (similar to equation 1.2)}
\]
The calculated IQF evaluates the observer’s ability to detect the cylinders, where \( i \) is the row number (from 1–15 in the CDRAD phantom), \( D_i \) the cylinder diameter, and \( C_i \) is the threshold value of the cylinder length (contrast) in row \( (i) \), and \( (j) \) represents the column, and \( n \) is the number of the cylinders available in the CDRAD phantom.

6.3.2.3 Theoretical background

The commercially available CDRAD phantom consists of \( S \times A \) cylinders where \( S \) is the number of the column and \( A \) is the number of the rows. The \( (S) \) cylinders are available in each row and are made of different densities. The highest density is located on the left of Figure 6.1. The cylinder’s density is the same in each column and the cylinder’s diameter is decreased when moving down the phantom, whereas the largest diameter is positioned at the top column. The number of observers who can record that the \( d_{i,i} \) cylinders from the left are the faintest limit named \( V_{a,d} \). For instance, \( V_{a,S} \) is the number of observers who could identify all \( S \) cylinders on the \( j \)th row. \( V_{a,1} \) represents the number of observers who could read only the first cylinder in the left. The sum number of observers (F) can be given by \( \Sigma_{a}^{S} V_{a,d} \ (a = 0,1, \ldots, A) \).
Figure 6.1: Contrast detail phantom of S x A cylinder images. The observers were asked to record the faintest cylinder that he/she can detect from the background on ath row (a = 1 to A)[7].

The ensemble X is a random variable that has a set of possible outcomes and it can be calculated from the detection rates of cylinders with different densities; ie, \( C_x = [c_0, c_1, c_2, \ldots, c_S] \), which have the probabilities \([p_0, p_1, p_2, \ldots, p_t]\) in which the \( p(x = c_s) = p_s \), \( p_s \geq 0 \) and \( \sum p_s = 1 \). The ensemble \( V_a \) is a random variable has a set of possible outcomes, which were computed from the number of observers for the \( a \)th row, \( HV_a = [h_{a,0}, h_{a,1}, \ldots, h_{a,d}, \ldots, h_{a,S}] \), and having probability \([p_{a,0}, p_{a,1}, p_{a,d}, \ldots, p_{a,S}]\) in which:

\[
p(V_a = h_{a,d}) = p_{a,d} = \frac{V_{a,d}}{p}, \quad p_{a,d} \geq 0.0 \quad (6.2)
\]

And total \( \sum p_{a,d} = 1 \) (Figure 6.2-a)
Assuming the zeroth approximation for simplicity in the CDP readings [121]. The conditional probability for \( x = c_s \) for the selection of \( m \)th cylinders on the \( j \)th rows could be written as follows:

\[
p(V_a = h_{a,d} | x = c_s) = \begin{cases} 
\frac{1}{d}, & d \geq s > 0 \\
0, & s > d \geq 0
\end{cases} \tag{6.3}
\]

In which we assumed that when the observer selected the \( d \)th cylinder, the observer could identify all the \( s \)th cylinder \((d \geq s > 0)\) equally (zeroth approximation) (Figure 6.2b). When there was no cylinder chosen or when the observer could not find a cylinder, \( V_a = h_0 \), \( P(V_a = h_{a,0} | x = c_0) = 1 \), and \( p(V_a = h_{a,0} | x = c_s) = 0 \), for \( s > 0 \)

The joint probability is then given by:

\[
p(V_a = h_{a,d}, x = c_s) = p(V_a = h_{a,d} | x = c_s)p(V_a = h_{a,d}) \tag{6.4}
\]

And the probability \( p_f(x = c_s) \) for the \( a \)th cylinder will be given by:

\[
p_a(x = c_s) = \sum_{H_{a,d} \in H_y} p(x = c_s, V = H_{a,d}) \tag{6.5}
\]

And then the entropy is written as follows:

\[
W_a(x) = \sum_{x \in c_s} p_a(x = c_s) \log \frac{1}{p_a(x = c_s)} \tag{6.6}
\]
6.3.2.4 Information loss calculations

Information content $H(x)$ is defined in the information theory as [7, 28]:

$$H(x) = \sum p(x) \log_2 \left( \frac{1}{p(x)} \right)$$

(6.7) (similar to equation 1.3)
Where \( p(x) \) is the probability of observing the object. When there is no information loss, i.e. all the observers could discriminate all the columns, \( p_i = 1/(\text{number of columns}) \), which is equal to 1/15 in the present case.

\[
H(\text{no information loss}) = \log_2(15) \quad (6.8)
\]

Information losses (IL) is then given by,

\[
IL = H(\text{no information loss}) - H(x) \quad (6.9) \text{ (similar to equation 1.4)}
\]

The total information loss is evaluated for different diameters of the phantom could be defined as

\[
TIL = \sum_{d=0.3}^{8} IL_d \quad (6.10) \text{ (similar to equation 4.6)}
\]

### 6.4 Results

Figures 6.3a and 6.3b are representative radiographic images of the air-Perspex and the water-Perspex CDPs, respectively. The air-Perspex phantom shows significantly better/sharper, i.e., short-scale contrast compared to the water-Perspex phantom for the same exposure factors (70 kVp and 20 mAs) [3]. This is due to the higher x-ray attenuation difference between air-Perspex compared to water-Perspex. Air-Perspex has a higher inherent difference in attenuation of the x-ray beam; i.e., it has higher subject contrast, and this is reflected in higher radiographic contrast in Figure 6.3a compared to Figure 6.3b.
Figure (6.3-a): A radiograph of contrast-detail phantom with air in holes.
Figure (6.3-b): A radiograph of contrast-detail phantom whereby the holes are filled with tissue equivalent medium “water”.

From Table 6.1, in the case of the air-Perspex phantom, most of the observers could not recognise the image of the hole that was 0.3 mm in diameter. In contrast, diameters up to 0.6 mm were not recognised by observers in the water-Perspex phantom. For both categories (air and water), the information loss (IL) (bits) is correlated inversely to the hole diameter of the CDP. The smallest detected diameter (0.4 mm) presents the greatest IL (54.75) and the largest diameter value (8 mm) has the lowest IL value (8.29). In addition, the water-Perspex phantom showed that the highest value of the IL (60), which was associated with the smallest detected diameter of 0.8 mm. Using the same exposure factors (70 kVp and 20 mAs), the
water-Perspex phantom demonstrates significantly higher information loss than the air-
Perspex phantom (see Table 6.1). As the water-Perspex phantom more closely resembles
human tissue x-ray attenuation, this suggests that the results from this study demonstrate a
range of information loss that more closely reflects imaging in clinical radiology.

The dependence of the information loss on the hole’s diameter, for both the air-Perspex and
water-Perspex phantom, is presented in Figure 6.4. It is shown that the levels of IL decrease
as the diameter size increases. It is also observed that at low values of IL, i.e. for larger size
holes (≥5mm), both versions of the phantom (air-Perspex, water-Perspex) gave the same
results. However, both phantoms showed significant differences in IL values for all other
sized holes. Larger IL values are obtained with the water-Perspex phantom, due to the fact
that the attenuation differences between the holes and the surrounding Perspex are very small
i.e. low inherent subject contrast. The total information loss for all the holes and the image
quality factor are displayed in (Table 6.2). Both TIL and IQF are higher (almost double) in
case of the water-filled holes.
Figure 6.4: Information loss (bits) versus Cylinder diameters (mm) for air-Perspex and water-Perspex CDPs. Images acquired using an Agfa DX-D 600 flat panel direct radiography system at kVp 70 mAs 20. The data presented in this figure (mean values ± standard deviation) was the result of three independent measurements (n=3) for fourteen participants (N=14). Correlation coefficient parameters of each fitted line are shown in the inset of the graph.

### Table 6.1

<table>
<thead>
<tr>
<th>Cylinder diameters (mm)</th>
<th>(Average)Air-Perspex KVP 70 mAs 20</th>
<th>(Average)Water-Perspex KVP 70 mAs 20</th>
<th>Poly. ((Average)Air-Perspex KVP 70 mAs 20)</th>
<th>Poly. ((Average)Water-Perspex KVP 70 mAs 20)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0.3</td>
<td>y = 0.265x² - 8.0721x + 70.298</td>
<td>R² = 0.9949</td>
<td></td>
<td></td>
</tr>
<tr>
<td>0.4</td>
<td>y = -0.0954x² - 4.1442x + 86.474</td>
<td>R² = 0.9559</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 6.1: Information loss values (bits) for each diameter (mm) in both conventional (air-Perspex) and modified (water-Perspex) CDPs. The data presented in this table (mean values ± standard deviation) was the result of three independent measurements (n=3) for fourteen participants (N=14). The data means of the water-Perspex are significantly different from the air-Perspex CDPs (*p<0.05, **p<0.01, two-tailed Student’s t-test).
Table 6.2: TIL (bits) and IQF values for both conventional (air-Perspex) and modified (water-Perspex) CDPs under the same exposure factors. The data presented in this table was the result of three independent measurements (n=3) for fourteen participants (N=14). The IQF values (mean values ± standard deviation) and the TIL values of water-Perspex are significantly different from the air-Perspex phantom (**p<0.01, two-tailed Student’s t-test).

<table>
<thead>
<tr>
<th>Diameter (mm)</th>
<th>kVp 70 mAs 20</th>
<th>Information loss (bits)</th>
<th>IQF</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>with Air</td>
<td>with distilled water</td>
<td></td>
</tr>
<tr>
<td>0.3</td>
<td>No holes detected</td>
<td>No holes detected</td>
<td></td>
</tr>
<tr>
<td>0.4</td>
<td>54.75±4.3</td>
<td>No holes detected</td>
<td></td>
</tr>
<tr>
<td>0.5</td>
<td>48.66±5.06</td>
<td>No holes detected</td>
<td></td>
</tr>
<tr>
<td>0.6</td>
<td>44.33±6.53</td>
<td>No holes detected</td>
<td></td>
</tr>
<tr>
<td>0.8</td>
<td>36.28±7.71</td>
<td>60.52±5.25**</td>
<td></td>
</tr>
<tr>
<td>1</td>
<td>30.18±6.09</td>
<td>57.27±6.09**</td>
<td></td>
</tr>
<tr>
<td>1.3</td>
<td>26.25±5.48</td>
<td>54.01±5.92**</td>
<td></td>
</tr>
<tr>
<td>1.6</td>
<td>21.29±6.22</td>
<td>49.45±5.27**</td>
<td></td>
</tr>
<tr>
<td>2</td>
<td>18.1±6.22</td>
<td>45.91±5.28**</td>
<td></td>
</tr>
<tr>
<td>2.5</td>
<td>16.9±6.04</td>
<td>41.44±4.74**</td>
<td></td>
</tr>
<tr>
<td>3.2</td>
<td>15.56±5.71</td>
<td>25.03±3.18**</td>
<td></td>
</tr>
<tr>
<td>4</td>
<td>12.74±5.41</td>
<td>17.41±5.4*</td>
<td></td>
</tr>
<tr>
<td>5</td>
<td>9.76±5.34</td>
<td>10.98±3.27</td>
<td></td>
</tr>
<tr>
<td>6.3</td>
<td>8.79±4.85</td>
<td>9.02±3.67</td>
<td></td>
</tr>
<tr>
<td>8</td>
<td>8.29±4.85</td>
<td>8.77±3.69</td>
<td></td>
</tr>
</tbody>
</table>

6.5 Discussion

The information loss values are varied between the two phantom types (holes filled with air and water) when using the same DR system with a grid and exposure factors (70 kVp and 20 mAs). According to Tanaka et al. [27], a higher grid ratio in the digital radiography system...
can provide an improvement in the detectability of low-contrast signals without increasing the patient dose. In this chapter, a comparison between two different phantoms using the same grid ratio (8:1) shows different results when using the same exposure factors. The IQF of the air-Perspex phantom (31) shows superior results compared to the water-Perspex phantom (53) in detecting low-contrast details (note; lower IQF value means better image quality). By comparing the results from diameters 0.8 mm to 8 mm (Table 6.1) the water-Perspex phantom shows higher IL values compared to the conventional phantom. These differences can be attributed to the fact that the attenuation differences between water and Perspex are less than those between air and Perspex for the same x-ray beam quality. This finding highlights the importance of the CDP material when assessing the information loss.

All the results indicate a direct relationship between the diameter of the holes and the levels of IL as determined by the observers’ responses. The level of IL is higher, under the same conditions, for the water-Perspex phantom than the air-Perspex phantom. This can be explained using the differences in attenuation of the x-ray beam for the two subject conditions. In the case of water-Perspex phantom the difference in the levels of x-ray attenuation between the two materials (water and Perspex) is low; i.e. there is low inherent subject contrast compared to the air-Perspex phantom. The IL is higher when imaging areas with low inherent subject contrast.

The IQF quantifies the image quality based on the observers’ detection capability [122, 123]. The IQF represents the area under the contrast-detail (CD) curve, has the dimension of square of length, and is given by the sum of products of averaged length and diameter of visible holes [123]. The TIL is dimensionless, does not rely on the cylinder diameter or the thickness
of the phantom, and is applicable to an all-or-nothing decision [124]. The TIL method, therefore, can be used to compare the performance of any two imaging modalities or techniques. However, the assessment of image quality by using TIL does not always agree with that of IQF [28]. This is due to the TIL calculation being obtained from the probability of conveyance of the image information, while the IQF is acquired from the distribution of image recognition (the average). Moreover, the IQF is less sensitive to individual decisions due to the fact that it represents the average of all observers’ outcomes. In contrast, the TIL can increase or decrease when a member of the group could recognise a cylinder, which other members failed to recognise.

In this project, the TIL and IQF of the water-Perspex was higher than that of the air-Perspex phantom by 175 bits and 22, respectively, for the same exposure factors (70kVp and 20 mAs). As shown in Figure 6.4, small diameter holes are more visible, when the air-Perspex phantom is used in comparison to the water-Perspex phantom. These results imply that the TIL values can provide a better indicator in comparing the performance between two different mediums (air/water) than the IQF values alone. According to Niimi et al. [7], the TIL can also provide more detailed discrimination between two different modalities e.g. CR and DR than IQF alone.

This study may be expanded to include other materials for the CDP such as contrast media. It would also be of value to investigate the efficiency of applying different grids and detecting the low-contrast details and information loss in DR and CR systems.
6.6 Conclusion

This chapter builds on the notion of employing the information theory, which stems from entropy. It was expanded by Shannon [25] to assess the levels of information loss in radiology imaging using images of contrast-detail phantoms assessed by observers. In this study, a modified version of the contrast-detail phantom was used [3]. In this modified CDP, the empty (air) holes of the CDP were filled with water to reduce the subject contrast and hopefully mimic human tissue more closely. Our results demonstrate that the material within the CDP influences TIL and IQF measurements. It was shown that the modified CDP, water-Perspex, with its lower inherent subject contrast that more closely represents soft tissue imaging in radiology, had higher TIL and IQF measurements. These higher measurements provide a more realistic account of TIL and IQF for soft tissue radiology imaging.
Chapter 7 Study Based on Information Loss of the Grid Effects in CR Imaging Using CDRAD Phantom

7.1 Summary

This chapter applies the information loss (IL) theory to quantify the image quality (IQ) of the computed radiography (CR) system when using anti-scatter grid. IQ in radiology is assessed by determining the levels of information retained or lost in an image. This assessment is made using a method based on IL and the employment of a contrast detail phantom (CDP). The CDPs are made of Perspex with holes of various sizes filled with air at atmospheric pressure. In this Chapter, the conventional CDP (air-Perspex) is used with and without a grid in CR to determine the information loss in radiologic images. Because it is necessary to set up the exposure usually to fit the grid used.

7.2 Introduction

The CR systems are based on photo stimulated phosphor (PSP) image detectors named image plates. These systems are widely used in radiology departments because they are a cost-effective means of transitioning from conventional film based imaging into digital imaging [46]. In emergency bedside radiographs, the wide latitude of the CR improves the good quality consistency. For the large regions of the body such as the chest, the CR IQ is higher than digital radiography which is more suitable for smaller body areas such as the extremities [46]. Even though digital radiography (DR) is becoming more commonplace, CR systems will continue to play an important role in emergency situations. The CR is also advantageous over the digital radiography when cost analysis is carried out [46].
Additionally, the CR system is more flexible in image plate positioning for difficult X-ray views compared to the DR system [46]. The processing of the image plate was discussed in details in (Chapter 2, Section 2.4). Whilst CR systems offer wider dynamic range of flexibility in positioning [46], scatter radiation negatively impacts image quality. This effect reduces IQ due to the interaction of the X-ray photons with the object being imaged [27]. Scattered radiation can be decreased by placing an anti-scatter grid between examined part and the imaging plate (the cassette). The grid selectively absorbs a large amount of scattered radiation and hence improves IQ [2]. Anti-scatter grid was discussed in (Chapter 4, Section 4.4). The aim of this experiment was to evaluate the effect of applying anti-scatter grid in the CR system. The IQ assessment methods used in this experiment were the total information loss (TIL) and image quality factor (IQF) obtained from conventional CDP (air-Perspex) with and without a grid at CR system.

7.3 Materials and Methods

7.3.1 Materials

The contrast-detail phantom CDRAD 2.0 (Artinis Medical Systems, Netherlands) was used in this study (Figure 7.1). This commercially available phantom is made of acrylic (Perspex; polymethyl methacrylate) of 10 mm thickness, in which 225 cylindrical holes of various sizes and at various depths have been drilled into the phantom medium. The diameter of the holes varies in size from 0.3 mm to 8 mm. This range is equally distributed across fifteen depths. These depths range from 8 mm (providing high-subject contrast) to 0.3 mm (providing low-subject contrast). The CDRAD phantom is a conventional CDP as it uses an air-acrylic interface to create image contrast [3].
The Shimadzu (R-20) CR system was used to acquire images of a commercially available conventional CDP (CDRAD 2.0) with and without a grid.

7.3.2 Methods
7.3.2.1 Test images
To have full scatter conditions as it is normally encountered in real patient imaging, images of the CDP were acquired with 5 sheets of scattering medium (Perspex square 265×265 mm with thickness of 10 mm) placed on top and another 5 sheets of scattering medium placed below. The source to detector distance (SDD) was set to 100 cm.

In the selected CR system, six phantom radiographs were acquired at 93 kVp and 1.2 mAs for gridded and non-gridded conventional contrast detail phantom. Three X-ray images were acquired for each condition. Grid 8:1 is placed on the cassette directly on the table top. Images acquired were processed using the clinical algorithm for the chest.
In all testing conditions, the phantom and scattering medium (Perspex sheets) were positioned directly on the table top. The grid is used because it is recommended for use with anatomical thickness of 10 cm or greater [43].

The image scoring of this experiment, the theoretical background of the IQF calculation and the information loss calculation used in this Chapter are identical to those outlined in the previous Chapter (Chapter 6 at the material and method section).

7.4 Results
Figures 7.1 and 7.2 are representative radiographic images of the two experimental conditions, using conventional CDP (air-Perspex) for the same exposure factors (93 kVp and
1.2 mAs) acquired without and with the grid, respectively. Applying grid in case of imaging the CDP leads to an improvement in the image contrast (as can be seen Figure 7.2 in comparison with Figure 7.1). This effect can be visualised more clearly in the right upper regions in both Figure 7.1 and 7.2 and is evident due to the increased number of air-filled holes being visible in Figure 7.2 where a grid is used.

Table 7.1 shows that hole detection is easier for the case CDP imaged with the grid compared to the CDP without the grid. The two smallest diameters of cylindrical holes in the CDP, that is, those between 0.3 - 0.6 mm, were not visualized by any of the observers who participated in this experiment when the grid was not in place. In contrast, only the 0.3 mm diameter hole in the CDP was undetectable with the anti-scatter grid (Table 7.1).

The comparison of the information loss (IL) between images taken with and without the grid is displayed in Figure 7.3. This Figure illustrates that the IL curve of the non-grid technique extends from 0.8 to 8 mm diameter (upper red curve) while the detected IL in anti-scatter grid technique were between 0.4 to 8 mm in diameter (lower blue curve) i.e resolution improved by almost 100% from 0.4 to 0.8 mm. This means that without a grid the minimum size observable hole is 0.8 mm while under same conditions with a grid the minimum size hole that can be detected is half of that 0.4 mm. These information loss curves indicate that the IL values were higher in the non-grid technique compared to the anti-scatter grid technique and the curve trends showing that IL reduces as the diameter of the holes increases as it can be seen at Figure 7.3. Moreover, Table 7.1 confirmed these results and showed higher IL values for the CDP without grid (56.42 bits) compared to with grid (47.58 bits) for the same diameter (0.8 mm). The IL values for the largest diameter (8 mm) were 14 bits for the anti-
scatter grid technique and 23 bits for the non-grid technique. For such a high subject contrast and high spatial resolution (largest hole) clearly the information loss is higher (23-14=9 bits ~ 60%) in case of without the grid.

In Table 7.2, the total information loss (TIL) and image quality factor (IQF) were calculated in both techniques for the same exposure factor (93 kVp and 1.2 mAs). The TIL and IQF for the conventional CDP with grid were 300 bit and 46 and the TIL and the IQF without the grid were 403 bit and 68. These results indicate that the images of the CDP obtained using the grid have a higher IQ because the grid allows more primary beam to get to the imaging plate while eliminating most of the secondary (scattered) photons (increases the ratio of primary to scatter), hence promoting both low TIL values and better IQ. Conversely, the CDP images obtained without the anti-scatter grid in place are affected by a higher incidence of scatter radiation interacting with the imaging plats, which causes greater TIL and poor image contrast.
Figure 7.1: A radiograph of conventional contrast-detail phantom acquired without a grid.

Figure 7.2: A radiograph of conventional contrast-detail phantom acquired with a grid.
Figure 7.3: Information loss (bits) versus Cylinder diameters (mm) for both with and without the grid techniques. Images acquired using Shimadzu (R-20) computed radiography system at kVp 93 mAs 1.2. The data presented in this figure (mean values ± standard deviation) was the result of three independent measurements (n=3) for fourteen participants (N=14). Correlation coefficient parameters of each fitted line are shown in the inset of the graph.
Table 7.1: Information loss values (bits) for each diameter (mm) in both with and without the grid techniques. The data presented in this table (mean values ± standard deviation) was the result of three independent measurements (n=3) for fourteen participants (N=14). The data means of the gridded are significantly different from the non-gridded techniques (**p<0.01, two-tailed Student’s t-test).

<table>
<thead>
<tr>
<th>Diameter (mm)</th>
<th>Information loss (bits)</th>
<th>KVP 93 mAs 1.2</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>without Grid</td>
<td>with Grid</td>
</tr>
<tr>
<td>0.3</td>
<td>No holes detected</td>
<td>No holes detected</td>
</tr>
<tr>
<td>0.4</td>
<td>No holes detected</td>
<td>57.2±0</td>
</tr>
<tr>
<td>0.5</td>
<td>No holes detected</td>
<td>54.61±3.67</td>
</tr>
<tr>
<td>0.6</td>
<td>No holes detected</td>
<td>53.27±6.05</td>
</tr>
<tr>
<td>0.8</td>
<td>56.42±1.79</td>
<td>47.58±5.17**</td>
</tr>
<tr>
<td>1</td>
<td>52.54±4.14</td>
<td>40.84±4.22**</td>
</tr>
<tr>
<td>1.3</td>
<td>46.21±5.22</td>
<td>35.4±5.12**</td>
</tr>
<tr>
<td>1.6</td>
<td>41.55±4.88</td>
<td>31.31±4.89**</td>
</tr>
<tr>
<td>2</td>
<td>37.54±5.18</td>
<td>27.21±3.39**</td>
</tr>
<tr>
<td>2.5</td>
<td>34.54±4.93</td>
<td>24.85±3.81**</td>
</tr>
<tr>
<td>3.2</td>
<td>32.21±5.59</td>
<td>23.32±3.56**</td>
</tr>
<tr>
<td>4</td>
<td>28.97±5.28</td>
<td>22.22±3.6**</td>
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<tr>
<td>5</td>
<td>25.93±6.87</td>
<td>17.4±5.35**</td>
</tr>
<tr>
<td>6.3</td>
<td>24.23±6.82</td>
<td>16.25±5.35**</td>
</tr>
<tr>
<td>8</td>
<td>23.8±5.04</td>
<td>14.38±6.27**</td>
</tr>
</tbody>
</table>
Table 7.2: TIL (bits) and IQF values for both with and without grid techniques under the same exposure factors. The data presented in this table was the result of three independent measurements (n=3) for fourteen participants (N=14). The IQF values (mean values ± standard deviation) and the TIL values of the gridded are significantly different from the non-gridded techniques (**p<0.01, two-tailed Student’s t-test)

<table>
<thead>
<tr>
<th>Exposure factors/ Technique</th>
<th>TIL (bit)</th>
<th>IQF</th>
</tr>
</thead>
<tbody>
<tr>
<td>CR system kVp 93 mAs 1.2 (Air holes without grid)</td>
<td>403.95</td>
<td>68.69±15.17</td>
</tr>
<tr>
<td>CR system kVp 93 mAs 1.2 (Air holes with grid)</td>
<td>300.82**</td>
<td>46±8.08**</td>
</tr>
</tbody>
</table>

7.5 Discussion

The calculations of the TIL and the IQF were performed on images taken using the conventional CDP with an anti-scatter grid and non-grid techniques by the CR system and same exposure factors of 93 kVp and 1.2 mAs. The scattered radiation is found to cause degradation to the image contrast which can be reduced by employing an anti-scatter grid. The results in Table 7.1 shows that the variation of the IL when using conventional CDP (air-Perspex) with and without a grid in a CR system. The information loss is greater with the small diameter cylindrical holes and it is smaller for the larger diameter cylindrical holes in both experimental conditions with and without a grid. Under the grid condition the IL ranged from a high of 57.2 bits (linked to the holes of diameter 0.4 mm) to a low of 14.38 bits (linked to 8 mm diameter). However, the IL value corresponding to the 8 mm diameter hole in images taken without the grid was 23.8 bits, a value of 39 percent higher than that obtained by the anti-scatter technique. This means a low level of contrast detectability encountered, hence low IQ in the non-grid technique. The results in Table 7.2 summarise the differences in the performance between the gridded and non-gridded techniques of CDP at CR systems for the same exposure factors (kVp 93 mAs 1.2). Without the anti-scatter grid, the CDP images
have higher TIL and IQF values, which are 25 percent and 33 percent higher than those obtained with the grid, respectively. These results of the grid performance agreed with the results obtained by Tanaka et al. [27] study. They found that the scattered radiation was high when images were acquired without the grid [27].

The visual assessment of the upper right corner of the CDP under both conditions (with and without grid) concurred with the results obtained from Table 7.2. The holes located in the upper right region in Figure 7.1 (representative of large holes) for the conventional (air-Perspex) phantom without using the grid appeared unclear and were associated with significantly high TIL and IQF values (403.95 bits and 68.69) respectively. However, the holes located at the same area with anti-scatter grid (Figure 7.2) had better visualisation; the holes were more visible and their edges were sharp. These were associated with lower TIL values and IQF (300 bit and 46) respectively.

The results presented in Table 7.2 shows that both methods of IQ determination i.e. TIL and IQF indicates an improvement in the image quality by inclusion of the grid. As discussed in Chapter 6, the calculation using TIL is considered to be more sensitive method compared to the IQF method because the IL differs depending on the distribution of detection rate. Also, the TIL calculation allows the evaluation of the variation between two techniques in bits for multiple observers [28]. This implies that TIL is more reliable indicator of the performance of the improvement to IQ that use of the anti-scatter grid yields (in this experiment CR system) and for same exposure factors (93 kVp and 1.2 mAs).
All the results demonstrate that the grid significantly improves IQ by reducing the scattered radiations reaching the imaging receptor, and there is a significant relationship between the levels of IL and the CDP hole diameters assessed by the observers.

7.6 Conclusion

The image quality is shown to improve under techniques of employing grid to reduce the scatter reaching the image receptor as obtained via the IQF and/or using IL. Determining the performance difference between the gridded and non-gridded techniques of CDP in CR systems for the same exposure factors is best achieved by TIL in comparison with the IQF. The calculation using TIL is considered to be more sensitive method compared to the IQF method because the IL differs depending on the distribution of detection rate [28]. The TIL of non-grid CDP was higher by 103 bits compared to the CDP using anti-scatter grid results, implying that using grid in CR system can significantly improve the efficiency of detecting the low contrast details and reduces the amount of IL, hence increasing the IQ.
Chapter 8 The role of the anti-scatter grid in flat-panel direct digital radiography investigated using IQF\textsubscript{inv}

8.1 Summary

The previous chapter investigated the impact of the anti-scatter grids on the image quality of computed radiography (CR) system using the information loss (IL) theory method. This chapter continues the investigation of the anti-scatter grid on the direct DR system using the inverse of image quality factor (IQF\textsubscript{inv}). The anti-scatter grids are used in radiologic imaging to reduce the level of scatter radiation reaching the image receptors. However, because they reduce the intensity of the primary beam getting to the receptors, they can indirectly increase noise in the image and hence reduce the signal to noise ratio. For this same reason, they could require a higher primary beam intensity, which would increase the patient dose [8]. Direct digital radiography (DR) is based on using detectors that are technically made to reject most of the scatter; this reduces the demand for the grids. Hence, the applicability of grids in this imaging modality has been debatable. This Chapter investigates the influence of such grids on the image quality formed by direct DR when using a contrast detail phantom. Three commercial systems were investigated: Agfa digital flat panel system (Agfa DX-D 600), Philips ProGrade DR retrofitted to existing Philips Optimus 65 X-ray machine and Shimadzu, RAD speed radiography system. The greatest IQF\textsubscript{inv} values for 109 kVp and 5 mAs with the
application of the anti-scatter grid technique were as follows: Agfa DR system (5.97), Philip DR system (7.02) and Shimadzu DR system (6.76). The improvement in image quality was shown to vary among the DR systems, with improvement being most significant for the Agfa DR system compared to the other two DR systems.

8.2 Introduction

A digital radiographic (DR) system with flat panel detectors is most commonly used in two-dimensional radiologic imaging to replace the screen/film system [1-4]. These detectors provide a dynamic range that optimises the image contrast and brightness independently. Consequently, the deterioration of image quality due to inadequate image processing and the reduction in diagnostic information are decreased [48].

In diagnostic X-ray modalities, contrast degradation is considered to be the key factor that influences image quality. Such degradation is caused by the scattered radiation resulting from the interaction of the X-ray photons with the imaged object. The most common technique for eliminating the scattered radiation before arriving at the image receptors is the insertion of an anti-scatter grid between the image receptor and the targeted body part of the patient for both the digital and the analogue systems. The grid consists of two metal foils that show low absorption and high absorption of X-rays; this combination decreases the scattered radiation dose and enhances the image quality [29]. The grid improves the image contrast by selectively absorbing a large amount of the scattered radiation compared to the primary beams. One of the disadvantages of the grid technique is the required increased radiation exposure to the patient. The performance of the grid properties can be addressed by two terms: the Bucky factor (BF) and the contrast improvement factor (CIF) [125-127]. The
incident radiation is increased by the BF, but the film density in the analogue system remains the same when using the anti-scatter grid. The CIF is the ratio of the radiographic contrast with the grid as compared to that without the grid [127].

The aim of this study was to investigate the impact of the anti-scatter grid on different commercial brands of digital radiography systems. The rationale for this study stems from the debate among different studies on the effect of applying anti-scatter grids with DR systems [27, 29]. Because the detector’s areas in the DR system are vulnerable to the effects of scattered radiation, the application of anti-scatter grids for DR systems could be similar to those in computed radiography (CR) systems, particularly for thicker body parts such as the upright chest radiography, in order to improve the image quality [48]. However, one study that included the peak signal to noise ratio showed no significant difference in the image quality between the images obtained by grids or without grids [29].

In this study, the evaluation of the image quality was achieved by utilising contrast detail phantoms (CDPs). These phantoms are designed to provide useful information on contrast detail detectability and are considered as the most reliable phantoms in assessing image quality, especially in low-contrast conditions [3, 19]. To quantify the overall detection performance, the inverse of image quality factor (IQF_{inv}) was calculated from the CDP data for three commercial DR systems: 1) Agfa digital flat panel system (Agfa DX-D 600), 2) Philips ProGrade DR retrofitted to existing Philips Optimus 65 X-ray machine and 3) Shimadzu, RAD speed radiography system. The IQF_{inv} values for the Shimazdu, Philips and Agfa DR systems with double mAs values were significantly greater with the grid than
without the grid, suggesting an improvement in image quality when using an anti-scatter grid with a DR system.

### 8.3 Materials and Methods

#### 8.3.1 Materials

An Agfa digital flat panel system (Agfa DX-D 600), a Philips ProGrade DR retrofitted to an existing Philips Optimus 65 X-ray machine and a Shimadzu RAD speed radiography system were used to acquire images of commercially available CDPs (CDRAD 2.0 (Artinis Medical Systems, Netherland). All these systems are flat-panel direct radiography systems utilising an amorphous selinum (a-Se) scintillator layer as the DR detector.

#### 8.3.2 Methods

To simulate clinical imaging, images were acquired with five sheets of scattering medium (Perspex square 265×265 mm with thickness of 10 mm) on top and another five sheets of scattering medium below the CDP. The source to detector distance (SDD) was set to 100 cm for the three systems. The kVp value was set to 109 kVp and the mAs values were set to 0.64 mAs, 1.25 mAs, 2.5 mAs, and 5 mAs for the three DR systems. All previously mentioned exposure factors were used in each experimental condition (with and without the grid). The phantom and scattering medium (Perspex sheets) were positioned directly on the table top. The detector was placed in the table bucky (i.e. the suitable grid within the distance 100 cm) assembly. This condition was preferred since grids are recommended for use in DR systems with a body part thickness of 10 cm or greater [43]. Images acquired were processed using the clinical algorithm of the supine anterior–posterior (AP) chest.
To compare image quality parameters of the CDP, the IQF was calculated using CDRAD analyser software (CDRAD phantom manufacturer). This software determines IQF as

$$IQF = \sum_{i=1}^{15} C_i \cdot D_{i,j}$$  \hspace{1cm} (8.1) (similar to equation 1.2)

Where $D_{i,j}$ represents the threshold (j) diameter in contrast column $C_i$, and the summation is over all columns.

The values of the IQF (related to image quality) decrease as smaller diameter holes are observed. However, as this could be misleading, an inverse measurement, IQF$_{inv}$, has been adopted [16]. IQF$_{inv}$ values increase as smaller diameter holes are observed IQF$_{inv}$ is calculated as follows:

$$IQF_{inv} = \frac{100}{\sum_{i=1}^{15} C_i \cdot D_{i,j}}$$  \hspace{1cm} (8.2)(similar to equation 2.7)

### 8.4 Results

Tables 8.1, 8.2 and 8.3 show the IQF$_{inv}$ values for three different DR systems. Each table represents the correlation between mAs values and the IQF$_{inv}$ with the application of the anti-scattered grid and non-grid techniques. These tables also compare the percentages improvement in the DR systems between the two techniques.
The lowest mAs value (0.64) showed no improvement by using an anti-scatter grid, the values were –6.38% and –5.99% for the Shimadzu and Agfa systems, respectively. However, for 1.25, 2.5 and 5 mAs, the IQF$_{inv}$ values were enhanced for the three DR systems. For the Shimadzu system, the IQF$_{inv}$ for the non-grid condition was 4.11, 5.39 and 6.37; these increased with the anti-scatter grid application to 5.47, 5.62 and 6.76 for 1.25, 2.5 and 5 mAs, respectively. These improvements were consistent for the Philips and Agfa DR Systems when using the anti-scatter grid. The enhancements with fixed mAs values (1.25, 2.5 and 5) for the Philips and Agfa DR systems were 24.16%, 18.55% and 18.78% and 27.19%, 23.27% and 39.16%, respectively.

When the mAs value was doubled (1.25, 2.5 and 5 mAs), the IQF$_{inv}$ indices significantly improved with the anti-scatter grid for the three DR systems. In the Philips DR system, increasing the mAs value from 0.64 to 1.25 mAs resulted in an improvement in the IQF$_{inv}$ from 3.47 with no grid to 5.55 with the anti-scatter grid; in addition, the improvement from doubling the mAs was about twice the improvement of the fixed mAs. The improvements with the fixed mAs (1.25, 2.5 and 5) were 24.16%, 18.55% and 18.78%, while the improvements with doubled mAs were 59.94%, 38.70% and 34.23%, respectively.

Figures 8.1–8.3 show different CDRAD phantom images acquired by utilising the anti-scatter grid. Each Figure demonstrates two CDRAD images: the exposed CDRAD phantom and the measured IQF$_{inv}$ curve of the CDRAD phantom calculated by the software analyser. A visual assessment of the three DR systems indicates that the centric and eccentric holes of the CDRAD phantom are clearly detectable at the right upper corner of the CDRAD phantom.
Table 8.1: $IQF_{inv}$ values for non-grid and grid techniques for the Shimadzu DR system. The data presented in this table (mean values ± standard deviation) was the result of three independent measurements (n=3).

<table>
<thead>
<tr>
<th>Exposure factors</th>
<th>Shimadzu, RAD speed radiography system</th>
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<tr>
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<td></td>
</tr>
<tr>
<td>109</td>
<td>0.64</td>
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</table>

Table 8.2: $IQF_{inv}$ values for non-grid and grid techniques for the Philips DR system. The data presented in this table (mean values ± standard deviation) was the result of three independent measurements (n=3).

<table>
<thead>
<tr>
<th>Exposure factors</th>
<th>Philips system</th>
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<tr>
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</tr>
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<td></td>
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<tr>
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<td>2.5</td>
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</tbody>
</table>

Table 8.3: $IQF_{inv}$ values for non-grid and grid techniques for the Agfa DR system. The data presented in this table (mean values ± standard deviation) was the result of three independent measurements (n=3).

<table>
<thead>
<tr>
<th>Exposure factors</th>
<th>Agfa system</th>
</tr>
</thead>
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<tr>
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<td>1.25</td>
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<td></td>
<td>2.5</td>
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</table>
Figure 8.1 (a) A radiograph of the contrast-detail phantom at 1.25 mAs with grid for the Shimadzu DR system.
(b) Contrast-detail score diagram and IQF$_{inv}$ curve.
Figure 8.2 (a) A radiograph of the contrast-detail phantom at 1.25 mAs with grid for the Philips DR system. (b) Contrast-detail score diagram and IQF$_{inv}$ curve.
Figure 8.3 (a) A radiograph of the contrast-detail phantom at 1.25 mAs with grid for the Agfa DR system. (b) Contrast-detail score diagram and IQF<sub>in</sub> curve.
8.5 Discussion

Scattered radiation travels in random directions and is not distributed equally at the image receptor (IR); this produces fog or useless signals to the IR and affects the useful primary radiation, which is not scattered. For small body part radiography, like hands, scattered radiation heads sideways or backwards rather than toward the IR, and this small amount of the scattered radiation has no effect on image quality. In contrast, large body parts, like the chest, produce a large amount of scattered radiation. In addition, the use of a high peak voltage in large body parts can lead to a large amount of scattered radiation. Anti-scatter grid has been used to eliminate or reduce the scattered radiation effects. It is composed of two metal foils that have low absorption and high absorption of X-rays, preventing scattered radiation (by selectively absorbing a large amount of scattered radiation compared to the primary beam) and enhancing image contrast.

Most of the exposure parameters for kVp and mAs that were used in this experiment were similar to those used clinically for chest exams. These two parameters have a major impact on image contrast. Modifications to the exposure factors influence a patient’s dose and image quality. For example, adjusting the tube current (mAs) controls the beam quantity, and the penetrating power of the beam is controlled by adjusting the tube potential (kVp). As the peak voltage increases, the image contrast increases due to increased attenuation of the X-ray. The mAs plays an important role in decreasing the radiation dose and enhancing image quality. There is a significant correlation between low radiation doses and noise production [48].
The IQF<sub>inv</sub> value is a good method for image quality evaluation; as the IQF<sub>inv</sub> value increases, the image quality improves. We found a significant, although variable, enhancement in the IQF<sub>inv</sub> values when using the anti-scatter grid technique in the Shimadzu, Philips and Agfa DR systems. At the lowest value of 0.64 mAs, the image quality Philips DR system improved with the anti-scatter grid by 27.67% compared to the other two DR systems (Shimadzu DR system –6.38; Agfa DR system –5.99). The highest improvement for 1.25 mAs was demonstrated by the Shimadzu system (33.09%) compared to the Philips (24.16%) and Agfa (27.19%) DR system. At 2.5 mAs and 5 mAs, the improvements of the Philips (18.55% and 18.78%, respectively) and Agfa (23.27% and 39.16%, respectively) DR systems were largely greater than the Shimadzu DR system (4.27% and 6.12%, respectively).

Increasing the mAs twofold showed a significant improvement compared with a fixed amount, particularly for the Shimadzu and Philips DR systems. In the Shimadzu DR system at 2.5 and 5 mAs, improvement with a fixed value increased by 4.27% and 6.12%, while doubling led to 36.74% and 25.42%. The improvement in the Philips DR system with the double mAs was almost twice that of the fixed mAs. A significant finding was that as the mAs doubled, the grid performance became more efficient.

This experiment clearly showed that the anti-scatter grid can provide good results when it is utilised in a DR system, particularly when doubling the mAs values, despite the fact that DR system manufacturers have attempted to improve image quality without using grids. We showed that the anti-scatter grid is a vital component in DR systems.
8.6 Conclusion

The IQF$_{inv}$, as a method for measuring image quality, increased for the same or twice the amount of mAs values when using a grid compared to the non-grid technique for three commercial DR systems. Utilising the anti-scatter grid technique in the DR system is vital for removing scattered radiation. Our results clearly show the importance of anti-scatter grids in DR systems to improve image quality by reducing the level of scattered radiation reaching the image receptors.
Chapter 9 A special contrast-detail phantom to quantify the Low Contrasts Detectability “LCD” of CT

9.1 Summary

The detection of low-contrast detail (LCD) in conventional radiography was investigated in the previous chapters through the use of the CDRAD phantom. This chapter further investigates the low-contrast detail (LCD) detectability with a specially-designed contrast detail phantom for the computed tomography (CT) modality. Usage of this modality is increasing due to its ability to produce and display images with information on adjacent tissues with detailed subject contrast. To establish that, CT is highly efficient in LCD detectability, an assessment of its capability in the low range contrast is performed using special-type phantoms, with the insertion of various high- and low-density materials. However, the limited variability in the density and atomic numbers of the materials in the commercially available CT low contrast detail detectability phantom limits the dynamic range of assessment.

This Chapter is not based on the Catphan phantom and includes a description of the design of another LCDP that was fabricated for this study. This new phantom is made of cylindrical Perspex, with holes drilled in evenly spaced radial lines, extending from the centre to the
edge. The holes gradually increase in diameter from 1.0–12.5 mm. Each line of holes is filled with contrast media of a certain concentration. This design mimics the low-contrast detail phantom used in conventional radiography, with the concentration of the contrast media replacing the depth of the holes. The concentration of the contrast media can be made as low as a small fraction of a milli-Molar (approximately 0.2 mmol).

9.2 Introduction

The technological development of CT modalities has enhanced the diagnostic ability of CT, resulting in an increase in the number of scans being performed per year. The high radiation doses delivered in this modality remains a concern, especially for specific clinical cases such as pregnant women or infants [12, 128]. Manufacturers of CT scanners have introduced various methods to reduce the patient dose relating, achieving dose reduction in vulnerable cases without affecting diagnostic sensitivity or reducing the amount of information retrieved [102]. Optimising CT protocol is essential and needs to be adapted to differing patient conditions. Selecting the correct tube potential (kVp), depending on patient size, is considered to be one possible strategy for the optimisation of CT protocol [128-130]. The kVp influences the object contrast because the attenuation coefficients of the materials are dependent on X-ray beam energy. Tube current modulation is another example [1]. The utilisation of intravenous contrast agents, like iodine, during an intravenous clinical CT examination, can increase the CT number at a lower kVp [131]. The effects of selecting a low kVp for the detectability of low-contrast objects has been discussed in many studies [22, 132, 133].

Low-contrast detail (LCD) detectability is a vital parameter in CT image quality as it is considered a low contrast imaging modality. The LCD detectability in CT can be defined as
the ability of the CT scanner to differentiate between tissues that have similar radiation attenuation characteristics, such as various types of soft tissue. The detection of small objects can be affected by noise, particularly if the contrast is low. The measurement of LCD is obtained by using phantom images, e.g. Catphan phantom, which contains objects of varied contrast and sizes [97]. These measurements are carried out with human observers scoring the images (radiologists) to determine the smallest object of the lowest contrast that they are able to visualise. The Catphan phantom consists of three series of nine cylindrical rods each, with diameter ranging from 2 to 15 mm, with three contrast levels (1.0, 0.5 and 0.3%). These contrast levels are expressed as percentages and defined as the variation in Hounsfield units (HU) between the mean pixel value measured on a ROI placed in the 15 mm object and a closed background region of equivalent size, divided by 10 (Figure 9.1). The Catphan phantom cylindrical rods are easily observable with increasing the diameter and object contrast level. For instance at 1%, most of the rods are detected, while at 0.3% only the large diameter rods can be detected [22].

Figure 9.1: An axial computed tomography image of the CTP515 module in the Catphan 600 phantom, showing the distribution of low-contrast objects in the computed tomography image [97].
The Catphan phantoms are therefore limited to measuring contrast variations between only a few set objects of different subject contrast. These phantoms are restricted to set contrast levels e.g. 0.3%, 5% and 1% [22]. Consequently, there is a strong demand to create a new phantom with ability to include various sized diameters, and which can accommodate any desired contrast level. This Chapter suggests extending the levels of contrast detail detectability to much more than that by using a number of cylindrical holes that can be filled with various concentrations of contrast media. The differences in the concentrations will represent different subject contrasts.

A phantom with the capacity to represent any tissue in the body will be of great value for the quality assurance (QA) of the CT, especially when assessing a very small variation in tissue contrast. Contrast detail phantoms (CDPs) are commonly used to assess image quality in conventional radiography, including digital radiography (DR) and computed radiography (CR) systems. The CDRAD phantom is one of the most commonly known CDP, and is used to detect the low-contrast detail detectability limits of these systems. It is made of acrylic (Perspex; polymethyl methacrylate), with 10 mm thickness, in a square shape. It has 225 cylindrical holes of different sizes and depths. The hole diameter ranges in size from 0.3–8.0 mm [3]. Because of the conventional radiography modalities, the images are acquired in two dimensions. These radiographic images generate information without the depth as it relies on the “x” and “y” axes only [2]. The CDRAD phantom is not suitable for CT image quality assessment as it is square in shape and its shallow depth makes it difficult to obtain slices from it. Moreover, the air filled holes compared to the Perspex (phantom material) represent very high subject contrast for CT, hence not suitable to be used in CT low contrast detail detectability phantom.
The CDP imaging design, made to detect low-contrast details of the CT modality, will improve the QA outcomes of today’s sophisticated CT scanners. Unlike the CDRAD in use in conventional radiography, the CDP for CT fabricated for this study accommodates the geometrical shape of large body parts like the chest and abdomen, provides in-depth information via the “z” axis and has the ability to accommodate a wide dynamic range of different contrast media fill.

This Chapter provides information on this specially designed CT phantom, called the CTCDP, which can be used to detect LCD for QA purposes at medical institutions. This phantom was validated using the image quality factor (IQF) method by 10 participants prior to being used for further experiments. This chapter explores the idea of employing a modified version of the CDRAD phantom for CT QA procedures, to be extended and modified further for use involving most CT imaging conditions.

9.3 Materials and method

9.3.1 Materials

9.3.1.1 Computed tomography contrast detail phantom (CTCDP)

This phantom is made of acrylic plates i.e (Perspex; polymethyl methacrylate) with holes drilled in it such that they vary diagonally in diameter from 1 mm to 12.5 mm. They increase in diameter incrementally from the centre towards the edge of the cylindrical phantom. The diameters selected are; 1, 2.5, 5, 6, 7.5, 9, 10, 11, and 12.5 as displayed in Figure 9.2. These diameters are selected to test the contrast detail detection abilities of the CT scanners at different spatial resolution limits from low (1 mm) and up to high (12.5 mm). This enables
the testing of a given CT scanner’s ability to detect contrast variations with large dynamic range of spatial resolution.

The cylindrical phantom has a length of 20 cm and a diameter of 30 cm. The phantom is scanned across its length. The holes spread across about 4 cm along its depth in proximity to its middle section [see Figure 9.3(B)]. The holes can be filled with various concentrations of contrast media to represent different x-ray attenuation conditions depending on the conditions required for testing of the CT scanner’s abilities. Each radial line of holes extending from the centre to the edge of the phantom is filled with a given concentration of contrast media. In this testing phantom, the concentrations selected ranged from 2.1 to 4 percent iodine contrast media in water. These holes represent small and large objects to be imaged under different conditions and the concentrations of contrast media to water can be modified to assess very low and high subject contrasts. An imaging slice through this phantom across the hole’s area will evaluate a scanner’s ability to faithfully image objects of extremely low contrast with very low spatial resolution through to high spatial resolution in various contrast and resolution conditions.
Figure 9.2: The image depicts a section in the phantom showing the distribution of the holes in the Perspex. The blue inner circle highlights the central points that extend to the edge of the phantom. Each hole in a given ray will be filled to a certain concentration of contrast media. The red ellipse encompasses a ray of such holes indicating their size variations increasing outwardly up to 12.5 mm.

9.3.2 Method

9.3.2.1 Image acquisition

CT scanner, Discovery CT590 RT, (GE Healthcare) was used to scan the phantom at Alfred Hospital – Melbourne - Australia. Phantom multi-slice images were acquired at 140 kV and 13 mAs by helical mode. Pelvis protocol was adopted to scan the phantom, with a field of view (FOV) of 512 X 512 mm, and slice thickness of about 1.25 mm. The acquired images were processed using the reconstruction algorithm based on filtered backprojection. The field
size for the reconstruction was selected to be 50 cm. The used beam filter was a combination of cooper and aluminium.

The phantom was positioned on the table top of the CT system, as shown in Figure 9.3(A). The laser light was aligned with the middle part of the phantom, where the contrast-filled holes are located. The holes of varying diameters in each ray extending from the centre to the edge of the phantom were filled with different concentrations of iodine contrast media, specifically 2.1%, 2.2%, 2.3%, 2.4%, 2.6%, 2.8%, 2.9%, 3.2%, 3.7% and 4.0%. A typical slice from the whole phantom scan was selected among those used in the scoring process and is displayed in Figure 9.4.
Figure 9.3: A) The scanning position of the computed tomography contrast-detail phantom CTCDP. B) The image demonstrates the real position for the CTCDP section that was filled with different contrast media concentrations.

### 9.3.2.2 The scoring process

Prior to data collection, informed consent was obtained from the participants through a certified human ethics approval document from RMIT University (ASEHAPP 62 see appendix VIII). For the sample forms of the invitation and consent form for participants and CTCDP scoring sheet (see appendices IV, VI and VII).

Similar to the CDRAD phantom in conventional radiography, the CTCDP can be assessed subjectively by radiologists. However, radiographers and medical physicists are also included as radiation technologists and scientists. To test this phantom, 10 participants (six radiographers, two medical physicists and two radiologists) were invited to contribute to scoring the phantom CT images in this study. Statistically, the sample size was low, but the
chief aim of this research was to establish proof of principle and it was not intended to constitute full-image QA focused research.

Each observer read the phantom images independently, using a medical grade monochrome LCD display. The room light was turned off. Observers indicated the faintest distinguishable disk images for each fixed diameter, i.e., the lowest-contrast cylinder visible on each column. The lowest-contrast visible cylinder for each fixed diameter was used as an index. These indices were averaged for each observer, and then used in the calculation of the IQF as follow:

\[ IQF = \sum_{i=1}^{n=9} C_i D_{i,j} \]  

(9.1) (similar to equation 1.2)

The IQF was calculated to evaluate the observers’ detectability limits, where “i” was the row number (from 1-9 in the CTCDP), “D_i” the cylinder diameter, and “C_i” the concentration of the contrast media in rows “i”, and “j” representing the column (the red eclipse in Figure 9.2), and n is the number of the different size cylinders available in the CTCDP. Equation 9.1 closely resembles the IQF equation employed with image observations when CDRAD is employed in conventional radiographical images. This equation was modified because the IQF calculation for conventional radiography takes into account the cylinder diameter and its corresponding depth. In the case of CTCDP the depth was replaced by the concentration of the contrast media. Higher IQF value means a good detectability level for the holes of that diameter, hence a high image quality. As the diameter increased, so did the IQF value.
9.4 Results and Discussion

The IQF average was calculated and recorded the faintest hole image as visualised by each observer. In other words, each observer reported the smallest and faintest hole that she/he can certainly visualise. An example of one observer’s data for detecting various diameter holes with different contrast media concentrations is displayed in Table 9.1. The averages obtained for all 10 observers are shown in Table 9.2.
Table 9.1: The IQF was derived from an average of the image scores recorded by individual participants

<table>
<thead>
<tr>
<th>Diameters</th>
<th>Image 1</th>
<th>Diameters × iodine concentrations</th>
<th>Image 1</th>
<th>Diameters × iodine concentrations</th>
<th>Image 1</th>
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<tbody>
<tr>
<td>1.0</td>
<td>No hole detected</td>
<td>0.000</td>
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</tbody>
</table>
All participants could not detect the smallest diameter of 1 mm, which clearly indicates the spatial resolution limits for this CT scanner. As the diameter increased, so did the IQF value. The detectability of the holes with a larger diameter was an indication of the effects of the subject contrast, as imaged by the CT. The concentration of the iodine could enhance the detectability of the holes, but it was correlated to the diameter size, as can be seen in Table 9.1. For instance, the IQF for the 10 mm diameter hole was higher than the IQF for the 9 mm diameter hole, using the same iodine concentration. The IQF average of the 10 participants varied between 1.61 and 1.75, and demonstrates good distribution with low standard deviation (Table 9.2). This new method for calculating the IQF for CT imaging and hence it is listed per hole also.

Table 9.2: The image quality factor (IQF) average values for each individual participant and the average for ten participants (N=10). The data presented in this table (mean values ± standard deviation) was the result of three independent measurements (n=3).

<table>
<thead>
<tr>
<th>Participants</th>
<th>IQF averages</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>1.61 ± 0.01</td>
</tr>
<tr>
<td>2</td>
<td>1.65 ± 0.09</td>
</tr>
<tr>
<td>3</td>
<td>1.62 ± 0.03</td>
</tr>
<tr>
<td>4</td>
<td>1.63 ± 0.05</td>
</tr>
<tr>
<td>5</td>
<td>1.69 ± 0.04</td>
</tr>
<tr>
<td>6</td>
<td>1.73 ± 0.01</td>
</tr>
<tr>
<td>7</td>
<td>1.71 ± 0.06</td>
</tr>
<tr>
<td>8</td>
<td>1.71 ± 0.02</td>
</tr>
<tr>
<td>9</td>
<td>1.73 ± 0.01</td>
</tr>
<tr>
<td>10</td>
<td>1.75 ± 0.00</td>
</tr>
<tr>
<td>Average of all the participants</td>
<td>1.69 ± 0.05</td>
</tr>
</tbody>
</table>
9.5 Conclusion

A new phantom design to complement the existing Catphan phantom used in CT image quality assessment which was tested has been demonstrated to be valuable for extending the contrast ranges to lower limits and increase the number of measuring points. The IQF method was applied to validate this newly designed CTCDP to quantify the LCD of objects using the CT imaging modality. The IQF scores recorded by all participants demonstrated efficiency with regard to the detection of low-subject contrast (small diameter) and high-subject contrast (large diameter). This phantom can be used to test the CT scanner’s abilities to distinguish very small objects with low subject contrast. Hence, it is well suited for the assessment of both the low contrast detectability and the spatial resolution of a given CT scanner. The information loss theory will be applied to the specially designed CTCDP and investigated in the next Chapter.
Chapter 10 Information loss in CTCDP slices

10.1 Summary

The previous chapter was discussed the experimental validation of the CTCDP in detecting low-contrast subjects. This chapter further investigates the advantage of using this phantom by employing the information loss (IL) theory. Visual assessment of the CTCDP was obtained by 10 medical radiation staff (six radiographers, two medical physicists and two radiologists). Thereafter, the information loss theory was applied to quantify the performance of the CTCDP mainly to employ it for determination of information loss in a slice as scanned by a CT scanner.

10.2 Introduction

The CT modality facilitates the detection of small lesions in the human body. It forms the images by using three dimensions, “x”, “y” and “z”, and produces different slices to allow the detection of small lesions or low-contrast detail (LCD) objects [1]. Image quality is assessed visually by the radiologists and sometimes the radiographers in hospitals, where high image quality is required [7]. This visual assessment is performed with CDPs, using the IQF factor. CDPs are usually used to provide information on contrast-detail detectability in conventional radiographical systems such as the DRs and the CRs. The contrast detail is usually evaluated using the Catphan series for the CT modality. However, the Catphan phantom is limited by the relatively small number of different materials with varying subject contrast that can be selected. This makes the measuring scale short as mentioned in Chapter 9. Moreover, as Giron et al. [22] observed nearby objects could be included inside the samples.
The CTCDP used to detect the contrast details for the CT modality introduced and validated using the IQF factor in Chapter 9 was specially designed to exceed these limitations. The aim of this study was to determine the information loss in a slice obtained from the CT modality by employing the information loss parameter in LCD objects of CTCDP.

10.3 Materials and Method

The materials and method are explained in detail in (Chapter 9, Section 9.3) including the CTCDP description, image acquisition and the scoring process. The information loss theory was applied to the acquired images as follows;

10.3.1 Information loss calculation

Information content $H(x)$ was defined in the information theory as

$$H(x) = \sum p(x) \log_2 \left( \frac{1}{p(x)} \right) \quad (10.1) \text{ (similar to equation 1.3)}$$

Where “$p(x)$” was the probability. When there was no information loss, i.e., all the observers could discriminate all the holes of the same diameter size, “$p_i$” = 1/(number of holes), which was equal to 1/10 in the present case.

$$H(\text{no information loss}) = \log_2(10) \quad (10.2)$$

Information loss was then given by: $IL = H(\text{no information loss}) - H(x) \quad (10.3) \text{ (similar to equation 1.4)}$
The total information loss evaluated for the different diameters of the phantom could be defined as:

$$TIL = \sum_{d=1}^{12.5} IL_d \quad (10.4)$$

(similar to equation 4.6)

### 10.4 Results

The CTCDP scanning made of large number of slices and only one slice was selected for scoring process. All the scoring results were obtained from slice number 205, from the middle object containing different concentrations of contrast media. The measured information loss for the different diameters of the CTCDP was recorded in Table 10.1. It can be seen that, as the diameters increased, the information loss values decreased. The smallest diameter (1 mm) was not visible to all the observers (Table 10.1). The second smallest diameter (2.5 mm) was detected and assigned to the highest information loss value (36.38 bits). The information loss value was reduced from a diameter of 5 mm (information loss of 30.32 bits) to reach the lowest information loss value at the greatest diameter size of 12.5 mm (information loss of 6.55 bits). This is represented in Figure 10.2, where the inverse relationship between the information loss and the diameter sizes is shown.

Regarding the different concentrations (Figure 10.1), the contrast media were not visualised at concentrations of 2.1% and 2.2%. With an increment in the contrast media concentration from 2.3% to 2.9%, the diameter of the CTCDP became more easily visualised, particularly at the largest diameter. Visual assessment of the diameter from 2.5–12.5 mm was enhanced in the same phantom due to the increase in the contrast concentration (from 3.2% to 4.0%). The total information loss (TIL) for all the different diameters in the CTCDP is 158.37 bits as shown in Table 10.1.
Figure 10.1: A computed tomography slice of the designed computed tomography contrast-detail phantom. More holes are observable at high concentrations (4-2.8 %) and with much difficulty to not observable at all at lower concentrations.

Table 10.1: Information loss (bits) versus the cylinder diameter (mm) and the total information loss (TIL) for the computed tomography contrast detail phantom CTCDP filled with different contrast media concentrations ranged from 2.1 to 4 percent. The data presented in this table (mean values ± standard deviation) was the result of three independent measurements (n=3) for ten participants (N=10).
<table>
<thead>
<tr>
<th>Diameter (mm)</th>
<th>Information loss (bit)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.0</td>
<td>No holes detected</td>
</tr>
<tr>
<td>2.5</td>
<td>36.38±0.50</td>
</tr>
<tr>
<td>5.0</td>
<td>30.32±1.24</td>
</tr>
<tr>
<td>6.0</td>
<td>25.71±1.08</td>
</tr>
<tr>
<td>7.5</td>
<td>21.57±1.31</td>
</tr>
<tr>
<td>9.0</td>
<td>16.25±2.47</td>
</tr>
<tr>
<td>10.0</td>
<td>12.60±2.18</td>
</tr>
<tr>
<td>11.0</td>
<td>8.95±1.68</td>
</tr>
<tr>
<td>12.5</td>
<td>6.55±1.29</td>
</tr>
</tbody>
</table>

Total information loss 158.37

Figure 10.2: The curve depicts the relationship between the information loss values in bits and the cylinder diameters in millimetres.
The data presented in this figure (mean values ± standard deviation) was the result of three independent measurements (n=3) for ten participants (N=10). Correlation coefficient parameters of the fitted line are shown in inset of the graph.

10.5 Discussion

Remarkable progress has been made with regard to the use of contrast and the spatial resolution of CT images, enabling the depiction of extremely small and subtle lesions. The ability to obtain a high-quality images and proficient perception by the observer are essential for the detection of such lesions, particularly in deep tissue [7]. The LCD assessment is known to be the most efficient evaluation method for imaging modalities, and relies upon subjective analysis, specifically that of the radiologist or radiographer in a medical institution. LCD detectability is a vital factor in abdomen cases, especially in hepatic CT. Hepatic tumours in CT images are recognised by the attenuation differences between the tumour and the hepatic parenchyma, termed tumour to liver contrast [132]. The aim of this study was to investigate the reliability of a LCD assessment of the number of visible objects of different contrast media concentrations of CT images extracted from CTCDP. Such an assessment is required as part of quality control programmes in several radiology departments worldwide.

From the varied concentration in our special designed CTCDP (Figure 10.2), the contrast of the phantom object was gradually enhanced. Distribution of the different concentrations in CTCDP was designed to closely mimic the CT examinations of large body parts where different tissues have different range of contrast levels, such as real abdomen cases. Also, the diameters of CTCDP holes vary in size and were designed in this way to mimic real CT examination during the detection of low contrast detail lesions, like hepatocellular carcinoma tumours, especially if the tumour size is ≤ 2 cm in the arterial enhancement phase [78].
Applying the information loss (IL) theory using CTCDP with respect to the CT modality allowed for a better evaluation of the image quality by calculating the information loss in imaged slice. The utilisation of total information losses (TIL) renders the image quality assessment more feasible because variations in the units (bits) between observers can be evaluated. The usefulness of total information losses was shown in another study in which a comparison was made between two different group of observers, however using non CT imaging systems [7]. In that study, the total information loss results of an expert radiographer (who had similar criteria to the radiologist in term of detection decisions) was 0.9 bit larger than that of the radiologist for the same exposure dose level (1.4 mGy in the flat panel detector FPD system).

The determination of the IQF is considered to be a valuable method when quantifying image quality, provided by the sum of products of the diameters and the average length of the visible cylinders. Total information loss was given in units of bits, and utilised information theory, while the IQF used the area under the contrast-detail curve. The total information loss was exhibited to be more sensitive to the observers’ distribution of image reading. For example, if two observers drew a different conclusion, , if one selected R+1, and the other the K-1, column, from the Rth column, the IQF would not change as a result, but the total information loss would decrease. This is because every pi (i < R) is increased by the amount 2/RN (R^2 – 1), where pi is the probability and N is the number of observers [7]. This method permits an assessment of the degradation of image quality as information loss. Also, the calculation of total information loss can be adapted in order to compare the different level of performance between varied modalities, for example, between CT scan and magnetic resonance technology, in detecting a solitary or several lesions [7].
In this study, we applied the information loss (IL) theory to CTCDP to measure the degradation of the image quality of the acquired CT images. Information loss (in bits) is inversely proportional to the diameter of the CTCDP holes (Table 10.1). The smallest visible diameter (2.5 mm) is found to be with the greatest information loss value (36.38 bits), while the largest diameter (12.5 mm) is with the lowest information loss (6.55 bits). This outcome demonstrates that greater information loss occurs when measuring small lesions or small diameters using the CTCDP i.e. under conditions requiring highest spatial resolution. Given the number of small diameter holes that have been incorporated into CTCDP, it is clear that this phantom is a very useful tool in detecting LCD and hence suitable for the performance of QA routine in medical institutions.

10.6 Conclusion

This type CT phantom introduced in this thesis facilitates the use of different contrast media concentrations, e.g. from 2.1–4.0% of iodine, within varied diameter sizes, making it suitable for the detection of low-contrast details using CT modality. It allows for LCD detectability to be tested and for information loss on acquired CT images to be measured. The application of the total information loss method provides a good indication of this phantom’s performance since it allows the amount of information loss for each diameter size in the CTCDP to be calculated.
Chapter 11 CTDI formula modified to include dose enhancement introduced by inclusion of the contrast media

11.1 Summary

The contrast media used in both projection imaging and the computed tomography (CT) scanning is introduced for only one purpose and that is to improve the subject contrast of the target. This is achieved by inclusion of the relatively high atomic number atoms of Iodine (making the contrast media) into the target prior imaging. Then the higher attenuation of the iodine atoms will result into clear image contrast which can be visualised hence showing clearly the target details. However, it has been established that these high atomic number atoms also enhances radiation effects i.e. the dose. Therefore they act as dual effects agent which enhances the image contrast but at the same time enhances radiation doses.

The primary aim of this chapter is to include the effects of the contrast media on the value of the computed tomography dose index (CTDI). The introduction of the contrast media into the target area not only enhances the contrast, it also enhances the absorption of the x-rays hence leading to dose increase. The levels of dose enhancements inflicted by the contrast media in the (CT) imaging is estimated, measured and related to the average CT numbers of the target and included in the CT dose index. The dependence of this factor on the concentration of the contrast media and the beam energy are presented.
A term is included in the CTDI equation representing the levels of dose enhancement as a function of the concentration of the contrast media and the beam energy. This factor is then related to the average CT numbers in the target. The levels of dose enhancements under normal CT conditions are estimated using mass energy absorption coefficient based equation and are also obtained and validated against those deduced from the CT numbers. The levels of contrast and dose enhancements are also determined experimentally under typical conditions for clinical applications of the CT scanners. The CT numbers in a well filled with contrast media in a phantom are compared to those of pure water filled same size wells and in the same phantom to determine the variation in the CT number values in the two wells when scanned by a CT scanner. From these data the dose enhancement values for a set concentration of the contrast media are obtained and the dose-enhancement values are validated against those obtained using mass energy absorption equation. In addition, the levels of dose enhancements are determined using CT gafchromic type film dipped in a reservoir of contrast media and another dipped in a water reservoir. The relative dose difference as obtained by scanning the two films represents the dose difference and hence dose enhancement.

During CT scans when contrast media is used the dose delivered should include the enhancement inflicted by the high density and atomic number of the contrast media atoms depending on its concentration in the target and the beam energy. For instance if the media concentration is about 10% of the target by weight it will enhance the dose by about 5% while if the concentration increases to about 20% the dose enhancement will rise to almost 20% of the total dose. These levels of dose enhancements are obtained when determined using mass-energy absorption equation and from the measurements based on contrast enhancements or using gafchromic films.
Dose enhancements caused by insertion of contrast media in the CT target should be included as a factor in the CTDI equation. Attempts of replacing current type of contrast media (iodine compounds) by heavy atoms in the forms of nanoparticles will further increase these CT levels of dose enhancements.

11.2 Introduction

Computed tomography is increasingly becoming a highly reliable imaging modality in radiology. Whilst computed tomography techniques had limited use during 70s and 80s, due to the increased scanning speed and improved image quality CT usage has increased dramatically since then. Today large numbers of patients are imaged by CT techniques covering almost every part of the body. According to the ICRU report about 250 million CT scans are performed per year [134]. Even though projectional imaging such as chest radiographs remains the dominant radiological imaging modality, CT is recognized as the modality that delivers the highest radiation dose to the public. As an example, according to Kalender et al. [1] and für Strahlenschutz et al. [135] whilst only about 7% of total x-ray imaging in Germany was done by CT, 60% of the total dose to the public from all the x-ray imaging modalities was attributed to the CT scans.

Whilst radiation dose delivered during CT examination is an important consideration, it is not measured directly but deduces indirectly through a dose indicator. This is because the distribution of the dose delivered by CT imaging through slices is not uniform and it cannot be measured directly. Shope et al. [136] introduced the definition of computed tomography dose index/indicator (CTDI) in 1981 as a parameter to indicate the level of doses delivered by CT scanners. Since then many modifications of the CTDI basic form has been introduced to
accommodate new conditions and new instrumentations introduced in this field such as the pitch value. Full meaning and description of this parameter (CTDI) is discussed by McCollough et al. [105] and also the references cited in this chapter provide comprehensive review of the changes introduced to this parameter over the years. For example, the dose estimated over volume and the difference between the dose-levels at the centre of the slice as compared to that on the periphery are currently included as corrections. Full details of employing CTDI to estimate dose delivered by a CT scanner with all the corrections can be found in Kalender et al. [1]. The IAEA code of practice for dosimetry in diagnostic radiology “TRS457” recognises the concept of CTDI as defined by the IEC [137] and to be measured by 100 mm pencil ionization chamber. CTDI is a dose indicator and all of its modified forms such as Dose Length Product ‘DLP’ are also dose indicators. CTDI is defined for sequential CT. For volume scanning based on the helical principle the pitch is included and the CTDI is then termed volume CTDI, ‘CTDI<sub>vol</sub>’. The DLP is the product of the CTDI<sub>vol</sub> and the scan length and it is utilised for multislice and multi detector CT. DLP can be converted crudely to measure the effective dose through the use of some conversion factors. Seibert et al. [138] recently introduced the effects of patient size on the CTDI<sub>vol</sub> and DLP values.

Contrast media, such as iodine-based compounds, enhance tissue contrast due to its higher atomic number and physical density. These characteristics dramatically increases the probability for the photoelectric effects and this leads to enhancing the CT image contrast [10]. Though CT images show far superior details of the target compared to projection imaging i.e. it is of much higher contrast, in some cases still further details are required clinically therefore contrast media of the type normally used in projection imaging (iodine compound) is required and used to enhance the image details by increasing subject contrast. The contrast media is used in CT imaging to show anatomic structures (usually blood vessels)
that would be indistinguishable from background tissue, in terms of their x-ray attenuating properties. A phantom based study by Jackson et al. [10] showed the levels of contrast enhancement by CT scan when various types of contrast media are used.

It has also been proven, through radiobiological studies, that inclusion of the contrast media in the tissue of interest (target) enhances the dose locally (or radio-sensitises the cells) at the target [11, 139-143]. Generation of copious secondary electrons i.e. free radicals from the interaction of x-ray photons with the higher density and particularly higher atomic number of the iodine atoms causes/leads to the increase in the photoelectric effect probability of interaction and this is the main reason underlying dose enhancement. However, the effect of contrast media on the dose and hence on the CTDI value is currently not factored into the equation for the determination of the CTDI value. This means that the CTDI value is calculated to be the same whether there is contrast media in the target or not. This chapter is based on the evaluation of the effects of the contrast media on the CTDI value. The results comprise both theoretical calculations and experimental measurements. Moreover this factor is derived in terms of the average CT number at the target. The contribution of contrast media to the CTDI value depends on the concentration of the media in the target and also on the kVp value. This makes it easy to determine the levels of dose enhancement directly from the CT numbers of the imaged target.

Paul et al. [144] conducted a study on patients CT data to determine the effects of the contrast media on the image noise and the dose delivered. The effects on dose were inferred from the CTDI$_{vol}$ value changes between conditions of non-contrast with the existence of the contrast media to the same targets. Their results showed an increase in CTDI$_{vol}$ values
between 3 to 13 percent. A clinical study by Amato et al. [145] also identified that inclusion of iodine contrast media in CT scans results in dose enhancement to various organs ranging from 20% to 70%. This study was based on patient scanned images for radiotherapy treatment. In addition, a recent Monte Carlo simulation based study conducted by He et al. [146] also demonstrated an increase in dose caused by micro-spheres containing contrast media in a modelled phantom. Whilst these studies indicate (or suggest) that contrast media does impact on CT dose, there is a need to theoretically and experimentally examine the effect of contrast media on CTDI.

This chapter aims to theoretically and experimentally examine the effects of the contrast media on the CTDI value and to relate it to the CT numbers. In particular, this chapter includes theoretical and experimental dose enhancement measurements/determinations following the inclusion of certain concentrations of iodine based contrast media in the target just prior scanning, in the CTDI equation and this chapter also relates dose enhancement of contrast media to the CT number. The dose enhancement at these levels of low doses are calculated according to the method introduced by Corde et al. [11] for radiotherapy beams which agrees well with the dose enhancement values obtained from the CT numbers.

11.3 Materials and Methods
11.3.1 Materials
A purpose built phantom was designed to investigate the effect of contrast media on CT contrast enhancement and dose. The phantom was made of Perspex and cubical in shape of dimensions about 3.8cm×3cm×5.9cm. Two holes were drilled in the phantom to be filled with various concentrations of the contrast media ranging from 0 to 100%. Two identical
holes were drilled in this phantom. The phantom holes’ diameter and depths were 1.25 cm and 4.5 cm respectively. One of the holes was filled with water and the other one with either 10% or 20% iodine based contrast media (Omnipaque 350, GE Healthcare) as shown in Figure 11.2. This phantom was designed to be used for the investigation of the contrast media effect on the CT dose. The computed tomography scanner used in this research to investigate the contrast and dose enhancement at Alfred Hospital Victoria Australia is a 64 slice type Discovery CT590 RT, GE Healthcare. CT type gafchromic films (XR-CT2) were dipped in various concentrations of contrast media then scanned at the same conditions in computed tomography then scanned using Image J programme. The image regional optical density can then be related to dose via a calibration curve or used directly as relative dose since these films have a very linear dose response [147] (see appendix IX for Gafochromic film calibration curve).

11.3.2 Methods

This section describes the methods used for the three components of this study.

The phantoms were axially scanned by computed tomography system (Discovery CT590 RT, GE Healthcare). The CT exposure parameters were set to 140 kV, 13 mAs and 1.25 mm slice thickness.

Image J software was used to quantify both contrast and dose enhancement. For contrast enhancement quantification, ten pixel value measurements were selected in the phantom and the average of these pixel values used as indicator of the dose to the central region of the CT acquired images (Figure 11.2) taken for the distilled water, iodine contrast media holes and
Perspex medium pixels. These points were selected inside the 10%, 20% of iodine contrast media and the water hole and also outside in Perspex medium. The image J software was also used to analyse the dose enhancement by taking the pixel value from 1cm long area on the gafchromic film (Figure 11.3).

Firstly, theoretical examination of the effect of contrast media on CTDI was performed using the CTDI general formula. Derivations of the CTDI general formula containing the dose enhancement factor “DEF(c,E) as a function of the concentration of the contrast media and the beam energy are introduced. The DEF is also related to average CT number in the target, using the definition of the dose enhancement based on the mass-energy absorption coefficient. The values obtained are then displayed graphically as a function of the concentrations of the contrast media and beam energy.

Secondly, contrast enhancement caused by the inclusion of the contrast media using the simple phantom was determined by scanning holes filled with various concentrations of contrast media and comparing the average CT numbers in the holes’ images to represent the contrast enhancement relative to the Perspex phantom material. The method is based on Jackson et al. [10, 116] procedure for contrast enhancement by contrast media.

Thirdly, dose enhancement due to the inclusion of the contrast media was also measured using CT type gafchromic film. The films were scanned in various media concentrations in reservoirs as depicted in Figure 11.3. The optical density at any region on the film is related to dose linearly.
Therefore the measured optical density obtained from film scanning represents relative dose. The gafchromic films were scanned using EPSON PERFECTION V700 PHOTO with the transmission mode and with the PTW software. The protocol followed for gafchromic film scanning can be found in the AAPM report and also some aspects of the scanning are outlined by PTW [147].

11.3.3 Derivations

The dose enhancement factor “DEF” will be linked to the CT number “N” and then the DEF will be included in the CTDI equation.

The basic formula representing the CTDI showing its value as an integral for the dose along the z direction which runs along patients’ longitudinal axis and covers 7 slices on each side is displayed in equation 11.1 below;

\[
CTDI = \frac{1}{nT} \int_{-7T}^{7T} D(z)dz
\]

Where T is the slice thickness and n is the number of slices.

However, when the contrast media is added then the whole dose will be enhanced due to the inclusion of the relatively high density and atomic number of the iodine compound. This dose enhancement should therefore be factored into the above equation. The levels of dose enhancement have been previously determined based on cells study and termed as dose enhancement factor DEF(c,E), where c stands for the concentration of the media and E the beam energy which directly depends on the applied kilovolt on the x-ray tube. The DEF can be written in terms of the mass energy absorption coefficient \(\mu_{en}/\rho\) [11] as follows:
\[
DEF(c, E) = \frac{\omega_t \left[ \frac{\mu_{en}}{\rho} \right]^T_E + (1 - \omega_t) \left[ \frac{\mu_{en}}{\rho} \right]^H,O_E}{\left[ \frac{\mu_{en}}{\rho} \right]^H,O_E E}
\] (11.2)

\(\omega_t\) represents the concentration of the iodine in the target by weight.

Whenever \(\omega_t\) value approaches zero then equation 11.2 will be reduced to equation 11.1 meaning that there is no dose enhancement or \(\text{DEF}=1\).

Now including dose enhancement caused by the existence of the iodine based contrast agent in the target in the CTDI equation will result in;

\[
CTDI = \frac{DE\text{F}(c, E)}{nT} \int_{-T}^{T} D(z) dz
\] (11.3)

However, \([\mu_{en}/\rho]\) is mass-energy absorption coefficient which is related to the linear attenuation coefficient in the following way (7);

\[
\mu_{en} = \mu_{tr} (1 - g)
\]

Where \(\mu_{tr}\) is the transfer energy factor and it is directly related to the linear attenuation coefficient ‘\(\mu\)’ as follows;

\[
\mu_{tr} = \frac{E_{tr}}{k \nu} \mu, \text{ where } E_{tr} \text{ represents average energy transfer}
\]

Therefore \(\mu_{en} = \frac{E_{tr}}{k \nu} \mu (1 - g)\)
Since transfer energy ‘\(E_{tr}\)’ will be almost the same for both cases of tissue equivalent material with and without the iodine contrast agent. The g factor representing the energy generated inside the target and deposited outside and not balanced by the electron equilibrium and assuming that the g factor is also the same in both cases then \(\mu_{en} = Y\mu\)

Where \(Y = \frac{E_{tr}}{h\nu}(1 - g)\)

Inserting the above relationship in equation 11.2 will be;

\[
DEF(c,E) = \frac{\omega_i Y \mu_i + (1 - \omega_i) Y \mu_o}{Y \mu_o}
\]  

(11.4)

From the definition of the CT number ‘\(N\)’ then

\[
DEF(\omega_i,E) = \omega_i N/1000 + 1
\]  

(11.5)

This equation implies that when there is no contrast media in the target then \(\omega_i\) will be zero and hence the DEF will be one. This means that when no contrast media is in the target area, there is no dose increase. However, when contrast media is added to the target with a \(\omega_i\) concentration then DEF will be greater than one depending on the concentration of the Iodine agent in the target at specific CT number “\(N\)”. 

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\[ CTDI = \frac{(\omega_f \cdot N/1000 + 1)}{nT} \int_{-T}^{T} D(z) \, dz \]  

(11.6)

Hence the CTDI becomes dependent on the concentration of the contrast media in the target.

In the case of no media in the target then \( \omega \) becomes zero and the above equation reduces to the basic definition of the CTDI presented in equation 11.1 above.

### 11.4 Results

This section describes the results for the three components of this study.

Firstly the DEF is formulated in terms of the mass-energy absorption coefficient. The relationship between the DEF and the beam-energy at various concentrations of the contrast media is displayed graphically. The DEF is also related to the average CT number of the target.

Secondly, the relationships between DEF and the CT numbers are validated experimentally by determining the contrast enhancement in relation to the CT numbers. The contrast enhancement i.e. CT number change is used to estimate the dose enhancement levels caused by the inclusion of the contrast media in the target.

Thirdly, the dose enhancements due to the inclusion of the contrast media and at the CT scan levels are determined experimentally using CT type gafchromic films.
11.4.1 Dose-Enhancement by Mass-Energy absorption

One way of obtaining dose enhancement factor is through substituting the mass energy absorption coefficient for both iodine compound with its concentration, and that for water, as tissue equivalent material, in equation (11.2). Dose enhancement values, as predicted by equation (11.2) above, depends on the concentration of the contrast media in the target and also it is influenced by the exposure especially the kVp value (beam energy). Graphical representation of this dependence is displayed in Figure 11.1. The dose enhancement values increase with beam energy up to the value of about 55 keV and then drops sharply till it drops to zero at energy ranges beyond the scale of the CT type x-ray tubes.

![Graph showing dose enhancement factor vs. beam energy for different iodine concentrations.](image)

Figure 11.1: Calculated dose enhancement factor from mass energy absorption coefficient for different beam energies and at different concentration of iodine contrast agent in the target.

The dose and contrast enhancements are also determined experimentally as described below;
11.4.2 Experimental Measurements of Contrast Enhancement

A typical tomographic slice of the phantom after irradiation is displayed in (Figure 11.2). It is observed from this Figure that the image of the wells of water and contrast media are clearly displayed. Image of the well containing the contrast media has a brighter display representing greater x-rays absorption in this region. The average value of the pixels in the centre of the wells compared to those outside represents the x-rays attenuation variation i.e the contrast. Table 11.1 data shows about 5 times more absorption of x-rays through the well containing iodine contrast media (20% concentration) in comparison to the absorption through the Perspex material. While a negative contrast of about 4 times is produced by water well. This result augments the usage of these contrast media in some cases of CT scanning where further contrast enhancement is required.

The CNR ratio are also presented in Table 11.1. From the definition of contrast C in digital images [2] or contrast to noise “σ” ratio;

Contrast to noise ratio “CNR” was obtained by determining the contrast value for the CT digital image using equation (11.7)

$$\text{CNR} = \frac{P_I - P_p}{\sigma}$$  \hspace{1cm} (11.7)

Where $P_I$ represents the average pixel value in the centre of the iodine image and $P_p$ is the pixel value at the phantom region other than the wells (the Perspex) and $\sigma$ represents the noise in the phantom part of the image which is just the standard deviation of ten pixels.
Table 11.1: CT numbers pixel values of the phantom holes filled with two concentrations of contrast media and water. The data presented in this table (mean values ± standard deviation) was the result of ten independent measurements (n=10). The data means of the holes filled with two concentrations of contrast media and water are significantly different from Perspex phantom medium (**p<0.01, one-way ANOVA).

<table>
<thead>
<tr>
<th>Medium</th>
<th>Pixel Values</th>
<th>σ Values</th>
<th>CNR Values</th>
</tr>
</thead>
<tbody>
<tr>
<td>Iodine Contrast Media 20%</td>
<td>893.3 ± 17**</td>
<td>17</td>
<td>42</td>
</tr>
<tr>
<td>Iodine Contrast Media 10%</td>
<td>512±11**</td>
<td>11</td>
<td>31</td>
</tr>
<tr>
<td>Distilled Water</td>
<td>46.2±5**</td>
<td>5</td>
<td>-29</td>
</tr>
<tr>
<td>Perspex Phantom</td>
<td>176.6±21</td>
<td>21</td>
<td></td>
</tr>
</tbody>
</table>

**11.4.3 Experimental Measurement of Dose Enhancement**

Dose enhancement was determined by measurements using CT type gafchromic films as described in the method section. The average pixel value in the centre of the film scanned by the CT scanner at 140 kV immersed in various concentrations of contrast media are presented in Table 11.2.

Table 11.2: Pixel values of CT gafchromic film for distilled water and three different concentrations. The data presented in this table (mean values ± standard deviation) was the result of three independent measurements (n=3). The data means of CT gafchromic film for three different concentrations are significantly different from distilled water medium (**p<0.01, one-way ANOVA).
<table>
<thead>
<tr>
<th>Medium</th>
<th>Pixel values</th>
<th>Dose enhancement percentage from (equation 11.5)</th>
</tr>
</thead>
<tbody>
<tr>
<td>distilled water</td>
<td>3428.1±2.76</td>
<td>-----</td>
</tr>
<tr>
<td>10% Iodine Contrast Media</td>
<td>3656.5±3.50**</td>
<td>6.6%</td>
</tr>
<tr>
<td>20% Iodine Contrast Media</td>
<td>4013.6±2.99**</td>
<td>17%</td>
</tr>
<tr>
<td>50% Iodine Contrast Media</td>
<td>5035.1±3.84**</td>
<td>46.8%</td>
</tr>
<tr>
<td>100% Iodine Contrast Media</td>
<td>5649.9±3.08**</td>
<td>64.8%</td>
</tr>
</tbody>
</table>

The higher the measured pixels value of the gafchromic film the higher the x-ray absorption. The differences between the average pixel values of the films immersed in reservoirs with iodine contrast media compared to that of the film in water reservoir represents the levels of dose enhancement. This method is acceptable as these films show a linear dose response[147] i.e. a linear relationship exists between optical density on the film and absorbed dose. Hence, at 10% the dose enhancement is about 7% while at 20% the dose enhancement approximates 20%. These values are in agreement within the experimental limits to those deduced from the CT numbers using equation 11.5 and also with those deduced from mass-energy absorption equation. This means that we are estimating between 5 to 20% dose increase due to the inclusion of 10 to 20% by weight contrast media in the CT scanning targets.
Figure 11.2: The area of the ten pixel values measurement from the contrast enhancement phantom at different medium: iodine contrast media 20%, iodine contrast media 10%, distilled water and Perspex phantom.
11.5 Discussion

The results of this chapter clearly indicate that around 40 to 50 keV (about 120 to 140 kV in average) of beam energy results in the largest dose enhancement to the target at all concentrations as it is clear from Figure 11.1. This energy is around the k-edge value of the iodine atom. For instance from Figure 11.1, the blue curve represents 10% concentration and the corresponding dose enhancement at k-edge value is approximately 15%. This further shows that clinically it will be better to have the kV values at the higher scale (>140) to reduce the dose dramatically in cases of imaging with the contrast media in the target.

From equation 11.5 when the media concentration is about 10% the dose enhancement is also around 5% approximately in agreement with the mass-energy absorption coefficients determination i.e. Figure 11.1. Likewise, using equation 11.5 and Figure 11.1 dose
enhancement at other concentrations such as 20% and 50% can be validated. This variation between the dose enhancement values obtained using equation 11.5 (and directly measured by the gafchromic films) with those obtained using mass energy absorption coefficient can be attributed to the fact that mass-energy absorption coefficients are not linear at this range of energies. These coefficients are more reliable at megavoltage range of energies which are much higher than the energies employed in CT imaging.

Therefore, using either through equation 11.5 or from the mass-energy absorption coefficients i.e. Figure 11.1, the dose enhancement can be determined at all iodine concentrations and at any kV adopted in practice. Moreover, equation 11.5 also relates the dose enhancement to the CT number of the target tissue type which makes the procedure of dose enhancement determination easy and all factors required for calculation are obtained in the CT data.

This dose enhancement is attributed to the higher absorption rate of x-rays by iodine atoms in the contrast media leading to increase in the generation of secondary electrons i.e. free radicals including Auger electrons. More details on the principles of dose enhancement caused by metallic nanoparticles and gold in particular can be found in literature [148].

The results displayed in Table 11.1 clearly shows contrast enhancement caused by the inclusion of the contrast media. This contrast enhancement is expected as it is well known that the iodine compounds of higher attenuation abilities will increase the subject and then the image contrast. From data in Table 11.1 for 20% contrast media concentration with pixel
average values of N=893 then using equation 11.5 the dose enhancement will be about 18% which matches approximately that anticipated from the mass energy absorption equation as displayed in Figure 11.1. Similarly for 10% the dose enhancement according to equation 11.5 will be about 6% which validates the dose enhancement values from the graphs for 40 to 50 keV (which is about equivalent for 140 kV). In general, dose enhancement as determined by equation 11.5 and linked to the CT number agrees with the values obtained from mass-energy absorption formula.

In this research the levels of dose enhancements as predicted by the mass-energy absorption equation, from the formula linking the dose enhancement to the CT number and also from direct measurements using CT type gafchromic films are approximately matched. This indicates that some level of dose enhancement occurs when contrast media is introduced into target tissues. It is therefore vital to include a term into the basic CTDI formula which accounts for contrast media introduction and reflects the level of dose enhancement.

11.6 Conclusion

Inclusion of the contrast media in the imaged target, not only affects image contrast, it also affects the dose delivered by the CT scanning. This effect has been introduced into the CTDI formula theoretically and validated experimentally. We believe that this more general form of the CTDI should be used in estimating the dose delivered by the CT scanners to the patients and public. The efforts of replacing iodine based contrast agents, in some special cases, with gold nanoparticles will enhance the dose even further more. In addition, the outcomes of this chapter will lead to further research that can examine the feasibility of optimising the levels
of contrast media used and also the exposure conditions (particularly the kVp) selected for CT scanning under the conditions of contrast media injection.
Chapter 12 summarises the conclusions of the thesis.
Chapter 12 Conclusions, Limitations and Future Directions

12.1 Conclusions

The focus of this thesis research was to investigate and improve the image quality (IQ) assessment methods of two X-ray-based modalities: conventional radiography [including computed radiography (CR) and digital radiography (DR) systems] and computed tomography (CT). The assessment approaches of these modalities include two lines of investigation: objective and subjective [as discussed in detail in (Chapter 2, Section 2.8)]. It should be noted that the end point of a radiological image assessment in the clinical environment is performed by the radiographers and image assessors (radiologists). However, an imaging system’s abilities and limitations in generating acceptable images are determined via the IQ assessment of these systems. One commonly adopted approach used to assess the IQ of a projection imaging system is the employment of contrast detail phantoms (CDPs) [9, 16, 18]. CDPs are designed to provide useful information about the contrast detail detectability, and have been shown to be one of the most reliable and commonly adopted phantoms for IQ assessments, especially in low-contrast conditions [114]. In fact, CDPs are commonly referred to as low-contrast detail (LCD) phantoms.

This thesis has explored the feasibility of extending and improving the most commonly used IQ evaluation methods by utilising a modified version of the CDRAD (for the conventional radiography system) and specially designed CTCDP (for the CT imaging system), filled with various contrast materials and concentrations for detecting the LCDs of these imaging modalities. In addition, this research was extended to include the use of the information loss (IL) theory and image quality factor (IQF) to validate the phantom’s performance, and
provide a better method to quantify the potential missing information. “It should be noted that more information is transmitted (or less information loss), the better the image quality is” [54].

In the case of the CT scan modality, the image quality is one issue and the radiation dose delivered to the public and/or patient is another. It is well documented that this imaging modality delivers the highest dose of radiation to the public when compared to all other modalities [12]. However, the determination of the dose per slice is not an easy task, and it has been well established that an indicator should to be used to determine the level of the dose delivered by the CT scan. This indicator is called the CT dose index (CTDI); however, it has always been used in all imaging situations which include the use of contrast media when it is administered to the patient. As explained in detail in Chapter 11, this research, for the first time, has addressed the effects of the contrast media on the total dose delivered by including a new factor in the CTDI equation. This thesis has also discussed a derivation of the general formula of the CTDI equation to measure the dose enhancement of the contrast media.

The main findings of this thesis can be demonstrated as follows:

- The results presented in this thesis have demonstrated that the conventional form of the CDRAD (air-filled holes) can be extended to better assess the LCD detectability by filling the holes with a range of attenuating materials, instead of only air. It has also been recently determined that the conversion of a conventional CDP into a low-contrast form is highly valuable radiologically [3]. The findings of this thesis have similarly suggested that using a modified form of the CDP for projection imaging systems could be a valuable addition to any radiology department.
• The air and Perspex combination provided a higher subject contrast. The subject contrast in the case of the water-Perspex was lower than that of the air-Perspex, since the attenuation difference between the two materials was much smaller than that of the air-Perspex. The air-Perspex interface represented a high subject contrast because air attenuates much less radiation when compared with other media, including Perspex. In contrast, by converting the conventional, commercially available CDP into a low-subject contrast form by adding water to each hole, the phantom more closely resembled the real case of patients imaged in radiology.

• The CR system experiment demonstrated the validity of utilising the information loss calculation in assessing the phantom’s performance, which can be used in the evaluation of clinical radiographs. The anti-scatter grid technique for the CR system maintained better detail and information when compared to the non-grid technique. The amount of IL was larger for the CR system when it was used without the grid with the same exposure factors conditions. This emphasises the importance of the grid application for controlling the amount of the primary radiation (increased ratio of primary to scatter) delivered to the image receptor, which can influence the IQ. It can reduce the amount of information lost and improve the efficiency of detecting the LCD.

• The impact of the grid application on the DR experiment showed that the anti-scatter grid technique can provide good results in improving the IQ when it is utilised in a DR system, particularly when doubling the mAs values. The IQF inv, as a factor for measuring the IQ, was increased for the same or twice the amount of the mAs value when using a grid, when compared to the non-grid technique, for the three commercial DR systems. Utilising the anti-scatter grid technique in a DR system is vital for removing scattered
radiation. The results clearly showed the importance of the anti-scatter grid in the DR systems to improve image quality by reducing the level of scattered radiation reaching the image receptors.

- Both of experiments using the conventional radiography systems (CR and flat-panel direct DR) indicated that the IL method was better than the IQF method in terms of the comparison between the different modalities. This is because the IQF is less sensitive to individual decisions, since it represents the average of all of the observers’ outcomes. In contrast, the total information loss (TIL) can increase or decrease when one member of the group is able to recognise the cylinder that the other members failed to recognise.

- For detecting the LCD, a specially designed CT phantom was created to assess the low contrast detectability. This phantom was made of cylindrical Perspex, with holes drilled in lines extending from the centre to the edge. The holes gradually increased in diameter from 1.0 mm to 12.5 mm. Each line of holes was filled with contrast media of a certain concentration, which mimicked the design of the LCD phantom used in conventional radiography, with the concentration of the contrast media replacing the depth of the holes. The phantom performance was validated on its ability to quantify the LCD of objects using the CT modality, by utilising the IQF and IL methods. The IQF scores recorded by all of the participants demonstrated efficiency with regard to the detection of the LCD (small diameter) and high-subject contrast (large diameter). The application of the TIL method provided a good indication of the phantom’s performance, and allowed the amount of information lost with regard to each diameter size in the CTCDP to be calculated. Accordingly, the applications of this phantom may include routine quality assurance checks in medical institutions.
The inclusion of the contrast media in the imaged target not only enhanced the image contrast, but also enhanced the dose delivered by the CT scanning. Although the radiation dose delivered during a CT examination is an important consideration, it is not measured directly, but deduced indirectly, through a dose indicator called (CTDI). In this thesis, the effect of the dose enhancement caused by the iodine contrast media was introduced into the CTDI formula theoretically, and validated experimentally. This derived formula including the dose enhancement can be used for estimating the dose delivered to the public and patients. With the inclusion of the contrast media effect, the dose enhancement caused by the contrast media was also linked, for the first time, to the CT number. These results strongly recommend the integration of the dose enhancement into the CTDI general formula.

This thesis attempted to develop an approach that positively enhanced the efficiency of previously mentioned modalities in detecting the low contrast differences that are encountered in clinical X-ray examinations. This was achieved by: 1) modifying the typical form of the CDP (air-Perspex) for conventional radiography systems into a low-contrast form that included various ranges of attenuating materials filling the phantom holes, 2) investigating the role of the anti-scatter grid in conventional radiography, including CR and direct DR systems, 3) designing a special CT contrast detail phantom to evaluate low contrast detectability, and 4) applying the IL and IQF methods in all phantom-acquired images. This work will improve patient outcomes by providing an enhanced evaluation process for the IQ and LCD that will enable imaging equipment to be more accurately calibrated, and minimise the dose delivered to the patients. In addition, in order to more accurately calculate the radiation dose of the CT scans, this thesis experimentally investigated a modified formula of the CTDI equation to calculate the estimated dose enhancement caused by the contrast media.
12.2 Limitations

Introducing contrast medium into the small holes within the CDP is technically very challenging. As demonstrated in Chapter 5, the contrast medium does not fill each hole uniformly. To overcome the technical issues associated with introducing a 30% concentration of iodine-based contrast media, an alternate form of a novel CDP may be a more viable option. Researchers are currently investigating the use of 3D printing to produce a phantom with increasing thicknesses of the attenuating material. Despite this acknowledged technical limitation, this study has provided proof of concept that a novel form of the conventional CDP yields additional information for assessing image quality in planar radiology systems, when compared with the conventional CDP (air-Perspex).

To overcome this issue, a prototype including filled holes is currently being produced utilising a 3D printing technique (Figure 12.1). However, the proper materials with the desired level of attenuation required for this experiment are not available in the 3D printing facility at RMIT University in Australia. Therefore, the experiment requires further investigation by our research team, which is beyond the time limits for this thesis.

In Figure 12.1-A, the prototype phantom is shown with two samples that were made from two different materials in the 3D printing lab. The holes in these samples are already filled (totally closed) with different materials to the bases of the prototype samples. According to the radiographic image (Figure 12.1-B), the attenuation thicknesses of these holes’ samples were inappropriate for the experiment, although these were the only materials available in the 3D printing lab.
Figure 12.1: A) Prototype samples of the contrast detail phantom showing two different filling materials. B) Radiographic image of the prototype sample.

12.3 Future directions

Further study is needed to include the impact of the exposure factors (e.g. kVp and mAs) on the IQ and LCD performance of various modalities. In addition, some CT parameters should be included when investigating the LCD, such as the algorithms used for the image reconstruction and slice thickness. Despite the good performance of the specially designed CTCDP phantom with regard to the IL and IQF methods, this phantom requires
accompanying software for calculating the $IQF_{\text{inv}}$ automatically. This will provide faster results with better performance in busy daily clinical situations.
References


New York
London: Springer.


Appendices:

Appendix I: Publication Arising From This Work

Low-Contrast Detail Phantom: Proof of Concept

Moshi Geo, PhD, Madeleine Shanahan, PhD, Salem Saed Alghamdi, MSc, Rob Davidson, PhD and Somayah Alghamdi, BSc

ABSTRACT

Purpose: To investigate the concept of filling the air gaps of the conventional contrast detail phantoms (CDP) with various concentrations of contrast media, and to develop a variable level of autoregressive level differential phantom that could be more appropriate for contrast measurements in some radiology cases.

Methods: Images were acquired using the digital radiography system of the modified CDP (Low Contrast Detail phantom) and the standard CDP where air at the holes were replaced with increasing material. In this study, two different contrast media were used: iodine, water and 2% and 3% concentration of iodinated contrast medium. Image quality test was used using automated processing to calculate the image quality factor (IQF).

Results and Discussion: Phantom studies indicate that lower contrast media are suited when CDP holes are filled with water and a 3% concentration of iodinated contrast media does these observed for air/Foam or conventional CDP. As an example, when a 5% NaCl solution is used the IQF values are 5.3, in the case of air filling the holes, however, when these holes are filled with water solution, the same condition, the value of the IQF values drop to 2.25, and to 2.81 when 3% of contrast media is used. Other concentrations were also tested. These results indicate that it is possible to instead of the contrast media to these phantoms to introduce agents that are more suited for a patient’s body than just air and some equilibrium material.

Conclusions: The results indicate that the proposed variation of the contrast media allows smaller changes in contrast to be observed. This is due to the small attenuation differences of the subject materials (e.g. 3% contrast liquid and water) with the conventional forms of CDP (unmodified). This suggests that the levels of the CDP may have a small role in reducing contrast in phantom radiology as an evaluative tool to better represent low-contrast detail imaging experiences.

RéSUMé

But : Étudier le concept de remplissage des espaces d’an air des formes de contraste des phagômes (CDP) avec différents concentrations d’agents de contraste et développer des formes d’insertion différentielles pouvant une variété d’imagerie appropriées, qui pourrait être plus appropriées pour la mesure du contraste dans certains radiographies.

Méthodologie : Des images ont été acquises à l’aide d’un système de radiographie numérique de la version modifiée du CDP (Low Contrast Detail phantom) et la version standard CDP dans lequel l’air a remplacé par du matériau dense. Dans cette étude, deux solutions d’iode étaient utilisées, l’une est un agent de contraste à 2% et 3% de concentration et l’autre est un agent de contraste à 30% de concentration. L’image a été testée à l’aide d’un procédé automatisé pour calculer le facteur de qualité d’image (IQF).

Résultats et discussion : Les études de phantoms indiquent que des concentrations d’iode plus faibles sont observées à l’aide des formes d’iode ou des CDP conventionnels. Par exemple, une utilisation d’un agent de contraste à 30%, les valeurs IQF sont de 5.3, lorsque les avantages d’iode remplacent l’air, les valeurs IQF sont de 2.25. Les résultats suggèrent que la concentration d’iode est aussi utile. Ceux-ci indiquent qu’il est possible de remplacer l’iode de contraste de 25% en utilisant des solutions plus légères pour éviter le surcroît de charges qui peuvent entraîner un surdosage au sein des particules de contraste.

Conclusions : Ces résultats indiquent que l’excitation appropriée des phantoms de contraste peut donner des variations de contraste plus faibles, sans entraîner de perte d’information sur les concentrations spécifiques (par exemple, le liquide à 30% de concentration) de sorte que ce qui est possible avec la forme conventionnelle...
Appendix II: Copy Rights for Using Published Work

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Appendix III: publication arising from this work
Appendix IV: Participants consent form

CONSENT FORM

1. I have had the project explained to me, and I have read the information sheet
2. I agree to participate in the research project as described
3. I agree:
   - to score the CD phantom images and complete the scoring sheet
4. I acknowledge that:
   (a) I understand that my participation is voluntary and that I am free to withdraw from the project at any time and to withdraw any unprocessed data previously supplied (unless follow-up is needed for safety).
   (b) The project is for the purpose of research. It may not be of direct benefit to me.
   (c) The privacy of the personal information I provide will be safeguarded and only disclosed where I have consented to the disclosure or as required by law.
   (d) The security of the research data will be protected during and after completion of the study. The data collected during the study may be published, and a report of the project outcomes will be provided to Dr Moshi Geso. Any information which will identify me will not be used.

Participant’s Consent

Participant: ___________________________ Date: ___________________________

(Signature)

Participants should be given a photocopy of this PICF after it has been signed.
Appendix V: CDRAD phantom Scoring Sheet
Appendix VI: CTCDP Scoring Sheet
Appendix VII: Invitation for participants

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Discipline of Medical Radiations
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Plenty Road, Bundoora, Victoria 3083
Telephone: +61 3 9925 7075
Fax: +61 3 9925 7063
medicalsciences@rmit.edu.au

Invitation to participate in research project

Investigators
The main investigator is Salem Alghamdi (PhD student)
The supervisor is Associate Professor Moshi Geso.
Co supervisors is:
Madeleine Shanahan (Senior Lecturer in Medical Sciences at RMIT University)

You are invited to participate in a research project being conducted at RMIT University. Please read this sheet carefully and be confident that you understand its contents before deciding whether to participate or not. If you have any questions about the project, please contact any of the investigators.

Who is involved in this research project? Why is it being conducted?
This research is a student project (Salem Alghamdi) for PhD degree requirements; the supervisor is Associate Professor Moshi Geso. Co supervisor is Madeleine Shanahan (Senior Lecturer in Medical Sciences at RMIT University). This project has been approved by RMIT Human Research Ethics Committee.

What is the project about? What are the questions being addressed?
This study will focus on advocating a novel method to determine the amount of information loss in images obtained by the latest CT scanners and digital projection type images such as computed radiography, direct radiography, and digital mammography. The outcome of such a study will be of great value to the radiology community and to all the CT users where it will be used to estimate the level of information loss during imaging procedures followed by such scanners. Therefore, the main rationale for this study is based on assessment of medical images. This project involves image assessment based on technology and also inclusion of observers.

This study aims to
- Implement a new method for Image Quality “IQ” assessment based on the level of information loss in digital images in general “projection type imaging” and in CT type images in particular. This will be achieved by combination of objective and subjective procedures.
- Assess the performance of the available image quality assessment software package in determining the diagnostic images quality based on the level of information loss in images as a new role.
- Investigate the feasibility of including this procedure of image quality assessment based on information loss determination (with the aid of the software) into the medical images quality assurance procedures as practiced in radiology departments.
Appendix VIII: Ethics Approval letter

18th December 2013

Moshi Geso
Building 261 Level 8, Room 22
School of Medical Sciences
RMIT University

Dear Moshi

ASEHAPP 62 – 13 GESO-ALGHAMDI Innovative method for CT and digital radiology image quality assessment based on information loss

Thank you for submitting your amended application for review.

I am pleased to inform you that the CHEAN has approved your application for a period of 2 Years from the date of this letter to 18th December 2015 and your research may now proceed.

The CHEAN would like to remind you that:

All data should be stored on University Network Systems. These systems provide high levels of manageable security and data integrity, can provide secure remote access, are backed up on a regular basis and can provide Disaster Recovery processes should a large scale incident occur. The use of portable devices such as CDs and memory sticks is valid for archiving; data transport where necessary and for some works in progress.

The authoritative copy of all current data should reside on appropriate network systems; the Principal Investigator is responsible for the retention and storage of the original data pertaining to the project for a minimum period of five years.

Annual reports are due during December for all research projects that have been approved by the College Human Ethics Advisory Network (CHEAN).

The necessary form can be found at: www.rmit.edu.au/staff/research/human-research-ethics

Yours faithfully,

Lincoln Jones
Chair, Science Engineering & Health
College Human Ethics Advisory Network

RMIT University
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Appendix IX: Gafchromic film calibration curve