Introduction

Digital images have vital advantages in health services. Image quality has been improved and patient radiation dose reduced by the introduction of digital imaging systems including computed radiography (CR) and digital radiography (DR). In addition, digital imaging modalities have revolutionised communication between radiographers, radiologists, and physicians. However, CR and DR also have some limitations such as higher initial cost, particularly for DR. In addition, consistent feedback that is required to obtain optimal acquisition may not be available for technologists. Potential increase in radiation dose, due to wide dynamic range of digital systems, is also
a potential drawback of CR and DR. Patients may be overexposed with more radiation than is required for a diagnostically sufficient image. Diagnostic information may be suppressed as a result of suboptimum image processing.\(^2\) Therefore, it is essential to regularly investigate image quality to ensure correct and accurate image interpretation.

No clinical detector can perfectly absorb all the incident x-ray photons. Some x-ray photons pass straight through the x-ray detector. Others that are absorbed may be re-emitted and exit the detector. As a result, there is loss in primary information. Additionally, noise arises from the amorphous array and readout electronics of the detector. These factors degrade image quality.\(^4\) Reliable diagnosis requires regular maintenance of the technology employed and alongside regular clinical evaluation of image quality.\(^3\) The criteria of optimum image quality should be determined and recognised.\(^7\)

The purpose of this review is to provide an overview of the parameters and their factors that influence image quality and to recognise the different evaluation methods and their corresponding approaches that are used to assess image quality and system performance.

**Image quality parameters**

There are several parameters that characterise the quality of digital images. Resolution, noise, and artefacts are the main parameters of image quality.\(^4\) Some studies include blur factors which relate so far to the spatial resolution.\(^2\) Figure 1 summarises these parameters and their influences.

Resolution describes the ability of medical imaging process to discriminate adjacent structures in organ tissues being examined. Signal from detected photon should be recorded with sufficient resolution in space, intensity and possibly time to produce a digital image that enables a medical interpretation of tissue structure and function. Therefore resolution is of three main categories, spatial resolution (space), contrast resolution (intensity) and temporal resolution (time).\(^8\) However, temporal resolution is more related to the digital radiography application of fluoroscopy; therefore it will be not discussed in this paper.

**Spatial resolution and/or blur**

Spatial resolution refers to the ability of imaging system to detect and discriminate small objects that are close together.\(^7\) The size of pixels and the spacing between them (the pitch) define the maximum spatial resolution. The smaller the pixel sizes the higher the spatial resolution. However, this is not always true because the spatial resolution is influenced by other causes such as blur factors.\(^7\)\(^,\)\(^9\)\(^-\)\(^10\)\) Image processing alters image spatial resolution however the image noise is excessively increased.\(^11\) Zooming or targeting and scanned field of view functions influence spatial resolution.\(^2\)

Measurement methods including the point-spread function (PSF), line spread function (LSF) and the modular transfer function (MTF), are used to quantify and evaluate spatial resolution.\(^11\)

Spatial resolution is affected by four blur factors, namely subject blur, geometric blur, motion blur, and receptor blur.\(^7\) Image blur refers to the element of blurring to boundaries in the object (patient). Sharp image describes the well-defined boundaries of the object (patient).\(^12\)

Subject blur is caused by object shape or/and structure composition. This factor is also called object blur.\(^12\)

Geometric blur results from the geometry of the image-construction procedures. The main influences of this factor are focal spot size of the x-ray tube, the distance between the x-ray source and patient and between the patient and image receptor. Border-blur increases with the increasing of focal spot size and with increases in the distance between patient and image receptor. Unequal magnification of different organ structures cause distortion in the radiographic images, which is called image distortion. For example, tissues close to the image receptor are magnified less than those further away.\(^4\) When the distance between the patient and image receptor increases, blur factor decreases.\(^11\)

Motion blur is the most problematic blur factor. When motion occurs, the boundaries of patient structures will move from their actual position during image processing. Consequently, the boundaries are blurred in the image. This motion originates from anatomic region being imaged and it can be either voluntary action of the patient or involuntary physiologic process. Voluntary motion can mostly be controlled by applying short examinations, instructing the patient to remain still during the examination and in certain situation using physical restraints and anaesthetics. However, such techniques are sometimes ineffective. Involuntary motion such as heart beats and bowel peristalsis cannot be stopped or minimised its influences on the images by using examinations of very short duration.\(^12\)

Receptor blur refers to the blur results from the image receptor. Image receptor gathers data produced during the imaging process and presents it as a visual image. Spatial resolution basically depends on physical detector characteristics. For example, the intrinsic spatial resolution of amorphous selenium utilised in direct conversion DR system is higher than that of structured caesium iodide utilised in indirect conversion systems. The detectors of structured caesium iodide has much higher intrinsic spatial resolution than that of unstructured scintillators.\(^11\) The thickness and material composition of the detector will determine its blur features. The factor of the blur increases with increasing thickness of receptor. The thickness also influences the sensitivity of the receptor which increases with increasing thickness.\(^6\)

Receptor blur is also caused by scattering or photoelectric interactions within the image receptor when the photon energy dissipates. A part or all energy of the photon deposited somewhere in the detector other than the original point of entry causing the blur: The scattering and movement of the laser beam, that is used to stimulate storage plate in the CR system, are sources of blur.\(^13\) Scattering of the laser light beam during storage plate readout is the primary source of special resolution loss in CR.\(^11\) The thicker the phosphor plates, the greater the scattering depth and blur. Dual reader systems reduce scattering problems. The introduction of structured phosphor allowed the use of thicker plates and provided improved detection efficiency without much loss of spatial resolution.\(^7\)

In indirect conversion DR (IDR), the source of spatial resolution loss is the spread of light photons during the x-ray-to-light conversion process which results in blur. Utilising structure phosphor increases detection efficiency and minimises the scattering light. However, direct conversion DR (DDR) does not suffer from this effect; because of the limitation of the spread of the electrons within the photoconductor material as they are directed towards the thin-film transistor (TFT) array.\(^2\)

Width of the detector, matrix size, pixel size, detectors pitch (spacing between detectors) are factors of spatial resolution loss in CR and DR systems.\(^2\) Locations of different x-ray absorptions within an element may be indistinguishable because all x-rays within an exposure contributes to a single quantity (the summed charge read from that element). So that, when the imaged structures of a patient are smaller than the size of a

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single element of the detector, they are smeared out and their contrast is reduced unless they are inherently high contrast objects. For example, when microcalcification is smaller than an element, it may be recognised as a calcification since its attenuation properties are so different from the other tissue in the element.6

Contrast resolution
Contrast resolution refers to the ability of an imaging system to discriminate objects with small density differences and/or differentiate small attenuation variety on the image.7 Contrast resolution explains how well the image discriminates subtle structures in organs being examined.8 Contrast resolution can be inherited by recording the information of interest with sufficient intensity resolution to discriminate the contrast details of interest.7 While the first step of digitisation, sampling in space, affects the spatial resolution, the second step, quantisation in signal intensity, influences the contrast resolution or the gray-scale bit depth.19

Contrast resolution is sometimes called tissue resolution.7 If there are two small objects with large difference in densities, the area between them is considered as high frequency or high contrast region. Conversely, low contrast region refers to the area between two small objects with small difference in densities.4

Contrast resolution is affected by tube collimation, number of photons, noise, scatter radiation, beam filtration, detector properties and algorithmic reconstruction used.4 Image contrast depends on subject contrast, detector contrast and displayed contrast.4

Subject contrast
The anatomical and physiological characteristics of the region being imaged are considered to be the intrinsic factors of image contrast, which are called intrinsic, subject, object, or patient contrast. Low intrinsic contrast tissues such as breast tissues have very subtle differences in composition. In radiography, the physical properties of atomic number, physical density differences among different tissues and patient thickness influence intrinsic or subject contrast.4

Imaging methods and techniques are the second major factor which control image contrast. Selecting careful exposure techniques for specific tissues and for certain purposes greatly enhances image contrast to obtain the desired information. For example, low kVp and small amounts of beam filtration are preferable in mammography to discriminate subtle differences among tissues. In chest radiography, however, high kVp and large amounts of beam filtration are used to demonstrate the wide range of varying tissues densities (lung, bone tissues). This technique helps in detecting lesions of increased physical density in the under the ribs.9

Introducing enhancement material or medium into the body improves image contrast by altering subject contrast. Contrast media changes photon attenuation properties from those of the surrounding tissues and therefore provide signal differences.4

Detector (receptor) contrast
A detector’s characteristics play an important role in producing contrast in the final image. Detector contrast is determined principally by how the detector detects and converts the energy into the output signal. The dynamic range of the detector influences the contrast resolution of image.4 The dynamic range of CR and DR, which is the ratio of the maximum to minimum input x-ray intensities incident on the detector surface, ranges from 1,000:1 to 10,000:1 compared with the dynamic range of film screen radiography which ranges from 10:1 to 100:1.2

Displayed contrast
The attributes of image displaying that is utilised to produce and demonstrate the final image influence the contrast of diagnostic images. For example, displaying images on a video screen gives one the flexibility to alter and adjust image contrast, unlike film based images. Viewing diagnostic images digitally demonstrates the data of images in a wide range of grayscale images. It also allows use of a wide range of exposures for display image. Consequently, image contrast is enhanced and radiation dose is reduced by utilising digital system.6 Therefore displaying process and devices of digital imaging systems (particularly for primary display or diagnostic interpretation) should be in compliance with the current Digital Imaging and Communications in Medicine (DICOM) standard of the American College of Radiology (ACR) and the National Electrical Manufacturers Association (particularly on grayscale displays).13

There are two categories for displaying digital images, small matrix (for CR, digital fluorography, and digital angiography) and large matrix size (CR and DR and digital mammography). A monitor of 5 megapixel (MP) typically 2048 x 2560 pixels, is sufficient for viewing digital images particularly CR and DR images. It is important to utilise zooming and roaming display functions to achieve a correspondence between the display pixel matrix and the detector element matrix in order to avoid resolution limitations of the monitor for partially displayed images. Moreover, display luminance influences image quality and therefore appropriate luminance should be uniform over the entire display and at a level of at least 200 cd/m2, especially for CR and DR. Bit depth resolution, which controls luminance quantification of soft copy display, is recommended to be large to prevent the loss of contrast details or the appearance of contour artefacts. Viewing environment and conditions also affect image display quality such room lighting and other display monitors light reflection.13

Noise
Noise is produced by the statistical fluctuation of value from pixel to pixel. Noise is recognised by a grainy appearance of the image. It is also characterised by a salt and pepper pattern on the image.5 Noise is un-useful information.14 The noise level is explained by the standard deviation, a measure of how spread out the pixel’s values are. The lower the standard deviation, the higher the accuracy of the average pixel value.

Noise images relates to the number of x-ray photons that are logged in each pixel (for DDR) or in each small area of the image (for CR and IDR).2,16

Goldman6 categorised the noise sources into three types, namely quantum noise, electronic or detector noise and computational or quantisation noise.

Quantum noise
Quantum noise appears when too few photons, after being attenuated by organs, are received. The lower the number of attenuated photons at the detector the higher the image noise.2 The main factors of quantum noise are anatomical structure size, decreasing pixel size, and scatter radiation. The disturbing anatomic background variability is often called anatomical noise.15
Detector noise

Noise originates from internal sources mainly image receptors which contain what is called electronic noise. Detector or receptor noise is produced because of non-uniform response to a uniform x-ray beam. This type of noise has fixed correlation to locations on the receptor, therefore it is called fixed pattern noise. Fixed pattern noise can be largely eliminated in digital imaging systems through post processing stages. Additionally, defects in the receptor's elements which may occur during the manufacturing process form unrelated structure in the image. Structured noise originates from different causes which creates unwanted signals or features on the image. Variations in pixel-to-pixel sensitivity and linearity, dead pixels and detector-response non-uniformities are the main causes of structure noise, particularly in DR.

Conversion noise occurs because of the fluctuations of the generated energy per detected photons. Conversion noise which is also called instrumentation noise can be reduced by utilising higher-intensity scanning laser in CR detectors and brighter phosphor screens in indirect flat-panel detectors to collect and generate more secondary energy carriers and hence improve QDE. In addition, lowering the number of conversion stages of process can also reduce conversion noise.

Quantisation noise

Quantisation noise is another source of noise which occurs during the digitisation process, translating analogue output voltage of detectors to discrete pixel values (grayscale values). The range of these values is determined by bits, binary on-off channels. Detectors of 10 to 14 bits (1024 to 16,384 digital values) are recommended to minimise quantisation noise in CR and DR systems.

Noise is also produced by scatter radiation which reduces subject contrast and decrease signal to noise ratio (SNR) and consequently degrades image quality. Using grid in CR and DR reduces scatter radiation and consequently reduces noise effect. However, the signal (incomplete transmission of the primary radiation by the grid) is also reduced.

Artefacts

Features that occur on the image and mask or mimic clinical features are called artefacts. Digital image artefacts are caused by image acquisition or object artefacts, hardware or image receptor artefacts, and software artefacts.

Image acquisition/object artefacts

Radiographers usually perform image acquisition by using image receptor. Therefore, image acquisition artefacts are due to operator errors. These artefacts include inappropriate exposure factors, un-collimated images, improper grid usage, scatter radiation, delayed scanning, twin artefacts, exposed image receptors and handling carelessness. Incorrect patient position, patient motion, improper x-ray beam collimation, and double exposure cause object artefacts. Inappropriate histogram analyses can cause object artefacts. Errors of histogram analyses are associated with improper collimation of exposure field, leading to very noisy, very dark or very white images. Metal objects also cause artefacts.

Hardware/receptor artefacts

Dead pixels in image receptors cause artefacts during the image processing stage and are called software artefacts. A few dead pixels may not interfere with diagnosis however many of these faults must be corrected. Radiation variation of x-ray beam over the image produce irregular configuration which again interfere with diagnosis. This can be corrected by equalising the response of each pixel to a uniform x-ray beam by utilising software pre-processing manipulation, namely flat fielding. Image compression is employed to facilitate transmitting and archiving of images. However, lossy compression techniques may cause redundancy of data and hence create software artefacts. Artefacts may occur through inappropriate use of software filters of grid suppression, low pass spatial frequency filter, and blur masking. Image transmission (communication) errors or failures cause artefacts. Incorrect flat field corrections and a failing amplifier are other sources of artefacts.

The above discussed parameters are judged objectively (statically measurement) or/and subjectively (human observation) to determine image quality level. In order to improve the quality of image, image quality parameters are manipulated because they are not independent. There are trade-offs in manipulating these parameters individually. Figure 2 demonstrates the dependent relationship between image quality parameters. Therefore image quality should be optimised for each specific purpose and specific region. For example, when spatial resolution is increased to get better image quality for bone tissue, the noise of image is also enhanced or hence increased visually.
However, there is a fundamental principle, radiation dose minimisation, which should be considered beside these parameters. Therefore, image quality is the balancing between image quality parameters and radiation dose.\textsuperscript{21,22} Optimised image quality relies on the balancing of the image quality and patient dose and depends on the region being studied and case being examined. To optimise image quality, image quality parameters mentioned previously should be manipulated and altered according to the purpose of examination with respect to the patient dose. Moreover, eliminating or limiting the effects of image degradation factors are also essential in optimising image quality.\textsuperscript{21,22}

**Image quality and radiation dose**

Optimal image quality is achieved at the lowest possible patient radiation dose. The high flexibility of CR and DR increases the opportunity of image quality optimising and radiation dose lowering.\textsuperscript{23} The minimum level of image quality and radiation dose should be determined based on diagnostic purpose.\textsuperscript{3} It is essential to recognise the parameters that affect radiation dose and their influences on image quality. Exposure factors including mA, time and kVp are the most important factors that control the radiation dose to the patient. The other factors that also affect radiation dose are patient size and detector properties.\textsuperscript{23}

Reducing mAs decreases radiation dose and consequently decreases SNR as the noise is associated with lower radiation dose. Lower radiation dose deteriorates contrast resolution of the image. High noise level images increase the risk of diagnostic details loss.\textsuperscript{24}

Lowering the kVp is essential to increase x-ray attenuation and consequently the contrast resolution of structures is improved. Lower voltage increases DQE of the detectors of digital system. As a result, image quality can be improved.\textsuperscript{25} In CR and DR, lower kVp techniques are more likely to improve SNR and hence the contrast resolution of image. However, low kVp techniques may increase radiation dose and image blur as a result of time increasing.\textsuperscript{7} Uffmann, et al.\textsuperscript{26} in their study found that selecting 90 kVp demonstrates the anatomic structure superior than that of 120 and 150 kVp without increasing the radiation dose to the patients. Changing tube voltage from 102 to 133 kV did not significantly improve contrast resolution of CR and DR.\textsuperscript{7} However, higher kVp should be used for thicker body organs to optimise the contrast resolution of the image.\textsuperscript{28}

In addition, selecting higher kVp may cause blooming or pixel saturation. The problem of blooming occurs when the saturation of the detectors is exceeded by illumination. The charge leaks to other pixels when the overfilled pixels lose their ability to accommodate additional charge. As a result the image quality is degraded.\textsuperscript{29}

Different detector systems have different detection efficiency and radiation dose reduction ability and hence different image quality. For example, the detector of IDR can provide better image contrast resolution than that of CR.\textsuperscript{28} Thicker detectors have better detection efficiency and hence higher ability of dose reduction.\textsuperscript{26} Spatial resolution of the image can be improved with small detector elements however higher radiation dose is required.\textsuperscript{25}

Therefore, good understanding of the influences of radiation dose factors on image quality is essential to obtain optimal image quality while maintaining lower radiation dose.

**Evaluation methods of image quality and imaging system performance**

The utility of radiologic images and the accuracy of image interpretation depend on two main factors; the quality of images and the ability of the interpreter. Good image quality is a major factor that allows physicians to interpret the image most accurately, correctly and timely.\textsuperscript{31}

Certain attributes are required for image quality evaluation tools and techniques to be used as quality control constancy examination. These tools should directly describe diagnostic performance, sensitively detect changes in the imaging system and not be expensive or too labour-intensive.\textsuperscript{15}

Several methods are used to measure the quality parameters of DR images and the performance of imaging systems. These methods are either physical, psychological or clinical (observers/diagnostic) performance (Figure 3). Physical methods include modulation transfer factor (MTF), noise, SNR, and detection quantum efficiency (DQE). Psychophysical evaluation methods include rose model (RM), contrast-detail analysis (CDA) and subjective assessment of physical parameters. Clinical performance measurement methods include receiver-operating characteristics (ROC) and visual grading characteristic (VGC).\textsuperscript{32,33} Figure 4 summarises the different evaluation methods of image quality and imaging system performance.
Detective quantum efficiency
The evaluation method of detective quantum efficiency (DQE) focuses on detector “image receptor” performance to assess image quality of certain imaging systems. Assessing detector performance method is based on purely quantitative analyses by measuring objective parameters related to detector performance. Such methods are considered indirect methods of image quality evaluation. DQE has been commonly used as a tool for image quality assessment and medical imaging system performance in general.\(^\text{15}\) DQE is based on linear-systems analysis (LSA) which is used to assess the ability of the system to transfer a signal and to characterise the noise associated with the system. The main measurement parameters of DQE methods are the modulation transfer function (MTF) of the system and the noise power spectrum (NPS). The MTF describes a system with the ability to reproduce and preserve the information of spatial frequency contained in the incident x-ray signal. The NPS describes the frequency content of the noise in the spatial frequencies of the system image.\(^\text{7,35}\)

There are several ways to calculate MTF which alter DQE approach and quantities. In fact, MTF was used separately before as a tool of image quality assessment. However, the sharpness of the final image is not described by DQE.\(^\text{15}\) DQE quantifies signal-to-noise ratio to the number of incident x-ray photons and characterises image quality.\(^\text{35}\) The main limitation of this method is that it ignores significant factors that affect image quality such as scatter radiation and image processing. Additionally, time consuming is considerable limitation of this method which makes it impractical in hospital basis environment.\(^\text{54}\)

Recently, DQE has been modified and improved to another method of image quality evaluation called effective detective quantum efficiency (eDQE).\(^\text{37}\) Some limitations of DQE are removed in eDQE. For example, factors that influence image quality such as scattering, magnification and image processing are now considered in eDQE.\(^\text{37,54}\) However, observers who are the second element in reliable radiology diagnosis are totally ignored in these methods. Moreover, they are difficult to implement as regular evaluation tools of image quality assessment due to the fact that they are time consuming and complex to some extent.\(^\text{15}\)

In general, the main limitations of the DQE method and its relative approaches have two drawbacks. Firstly, they do not provide description of all components in the imaging process. They give limited information about the characteristics of the produced image. Factors such as dose level and display characteristics which influence final appearance of the image are not considered in these methods and relative approaches. Secondly, they do not consider the anatomical background which limits the observer performance in detecting pathology. Anatomical background is considered as a factor of hindering detection of pathology.\(^\text{35}\) The ability of observers to detect details is reduced by anatomical details, even though the mechanism of this effect is not clear and is not really understood.\(^\text{15}\)

Therefore, the reliability and validity of recent approach of DQE and relative approaches are high in providing accurate measurement of the ability of information transfer. However, their validity is low in assessing the entire imaging system.\(^\text{35}\)

The Rose model
The method of RM, SNR based method, is another tool used to evaluate image quality of digital radiographic images. Rose, in 1953, used images to estimate the maximum amount of information that can be translated into visible image by numbers of photons. Quantum efficiency (absolute scale) is used in this method to evaluate the performance of imaging systems by utilising a simple model of signals detectability which is assessed by human observation. Later, Rose’s quantum signal detection model is based on SNR. It gives a description of visibility of an object in an image.\(^\text{15}\) Phantom of a number of disc-like objects of different size (0.3–8.0 mm diameters) and diverse contrast, represented by sample depth (0.3–8.0 mm), is utilised as well. SNR is calculated to measure image quality in this method based on linking the mathematically calculated SNR to the results of detection examinations. SNR describes noise and resolution characteristics of image and human visual system.\(^\text{15}\)

There are some problems with this method which influence its validity and reliability in evaluating image quality. First, the size of the objects are not considered in SNR measurements in this method. Second, the noise description used in SNR is overly simplistic for observers who are sensitive to the noise characteristics. Third, to offer the same imaging conditions, a larger number of photons for the image are used with smaller pixels. Meanwhile, the observers are mostly not interested in single pixel values and are not affected by the pixel-to-pixel variations. Fourth, observers are not often affected by pure noise from the anatomical background. Hence, the validity of using SNR methods is very low to measure image quality. Therefore, it is not recommended that using SNR methods to compare different imaging systems or various image processing procedures.\(^\text{35}\)

Information entropy
Tsai, et al.\(^\text{1}\) suggested a new evaluation method of image quality, IE, which is a quantitative measure of the information transmitted by the image. The concept of information entropy describes how much information (randomness/uncertainty) is provided by the signal or image. It is a simple and straightforward method based on single parameter, transmitted information.\(^\text{1}\) Step wedge phantoms of varying thicknesses are used in this method. Images of phantoms are detected, for example, by storage phosphoric plate for CR. Several images are taken with a variety of exposure times. Because of the variety of thickness of step wedge phantom, the images demonstrate a gradual scale of grey level with diverse values. The more information conveyed the better the image quality.\(^\text{41}\)

The authors found that IE is a useful method for the evaluation of physical image quality in medical imaging system. The results of their study demonstrated that there was a correlation between the transmitted information and both image noise and image blurring.\(^\text{41}\)

The main advantage of this method over DQE is that the final image is considered in the evaluation procedure. Other advantages of this method which include simplicity of computation and experimentation and the combined assessment of image noise and spatial resolution. However, its validity still low as human observers are not used in this method. In addition, the simplicity of the used phantom reduces the reliability of this method. Step wedge phantom is limited by several different thicknesses without considering sample sizes. Tsai, et al.\(^\text{1}\) state some limitations of IE evaluation methods. Unlike MTF and NPS, information entropy measures do not provide frequency information. In addition, this method does not separately demonstrate the effects of different noise sources such as the electronic noise and structural noise.

Information loss theory is a newly evolved method of image quality based on entropy theory. Contrast detail phantoms are also used in this method.\(^\text{12}\) In fact, utilising CDA allows to consider the whole parameters of image quality.\(^\text{12}\) This method proves more sensitive in evaluating the contrast-detail image quality than using the average values of detectability,
and it allows to measure observers' differences in unit of bits. This method is based on measuring the Total Information Loss (TIL). Niimi, et al. concluded that TIL can be used as a tool of performance assessment of different medical imaging modalities

Contrast-detail analysis

A commonly and widely used tool to evaluate image quality is contrast detail analysis (CDA). This method provides quantitative evaluations of low contrast and small detail measurement of medical images. CDA originated from the theory of signal detection which implies that low contrast-detail detectability is related to internal signal-to-noise ratio of the observer. The main assumption of this theory is that noise from different sources interferes with sensory stimuli to the observer. The ability of imaging system to visualise small objects which are of very low contrast describes low contrast-detail detectability of the system.

Low contrast-detail detectability can be assessed by measuring the ability of observers to detect the smallest objects which have varying contrast differences with the background. CDA is an approach to describing the image quality in terms of detail (drilled holes of varying diameter) and contrast (varying depth). Low contrast-detail detectability implies that the detectability of details increases with increasing the size of objects and/or contrast between the objects and the background. For example, when the objects' size increases while keeping the contrast differences the same, the detectability will increase. The detectability will also increase when the contrast differences between the objects and the background increases while maintaining object size. In other words, the large objects can have lower contrast than smaller objects for the same detectability performance.

Human observation is mostly involved in the process of evaluation to visually measure contrast-detail on the image. Therefore, this method is considered a subjective evaluation.

Observers are asked to score what they can detect on the phantom image on the first three rows and to score and locate (in which corner the object is) what they detect on the rest rows in order to limit false positive score. By plotting the smallest visible diameter (Cj) against the smallest visible depth (Dj), for all rows i, a contrast-detail curve is obtained. The next equation is used to calculate inverses image quality figure inverse (IQFinv). The greater value of the IQFinv, the better is the low contrast detectability.

\[
IQF_{inv} = \frac{100}{\sum_{j=1}^{15} C_j \times D_{0,j}}
\]

Where \(C_{i,th}\) is threshold contrast
\(D_{0,i}\) is threshold detail

CDA method provides quantitative evaluations of low contrast and small detail measurement of medical images. Therefore, it is considered straightforward and direct method of image quality assessment. Moreover, low CDA studies consider the whole processes of imaging systems such as detector design, x-ray parameters, image acquisition and processing, image post processing, and image displaying. Therefore, CDA is selected to provide insightful understanding of CR and DR systems.

A recent study by De Crop, et al. investigated the correlation of low contrast-detail performance measurement and clinical image quality assessment in chest radiography. The findings of this study suggested that there is significant correlation between physical (low contrast-detail measurements) and clinical evaluation methods. The researchers concluded that the CDA method is relevant for image quality optimisation. While this method was based on a phantom, it does not require volunteer patients. Therefore, the evaluation method of CDA can be used to compare and contrast the image quality of different systems.

In fact, the reliability and the detectability of such methods are affected by the variation of human perceptions and decisions. Furthermore, visual assessment of image quality by the human observer is time-consuming and arduous and may lead to wrong results in many situations.

Pascoal, et al. in their study suggested an objective CDA method to assess image quality by utilising automated scoring by employing a software package (CDRAD analyser). It is suggested that this avoids the subjectivity of commonly used method, CDA, because it is based on measurements of image data such as signal-to-noise ratio. Even though the method of employed CDRAD analyser proves more sensitive to detect smaller low-contrast variations; human observation is able to detect smaller details. CDAs are useful for quality control, standardisation purposes and for indicating typical or acceptable performance of medical imaging systems.

However, using CDA is still criticised because they are based on homogeneous patient simulating phantoms and they do not represent the real situation. Noise from anatomical background which effects detecting ability is simply not considered in such evaluation methods. The detectability of objects are often much more limited by anatomical background structure than by noise from an imaging system.

Receiver-operating characteristics analysis

The evaluation method of ROC is another tool used to evaluate imaging performance of imaging systems. ROC is a task-based method and it involves human observers. It measures the sensitivity and specificity to evaluate the accuracy of diagnostic imaging systems. The sensitivity and the specificity measurements describe the abilities of imaging system to assist interpreters to correctly diagnose the disease when the patient actually has the disease and to correctly exclude the disease when the patient truly does not have the disease. The sensitivity measures the probability that a patient who actually has the disease is determined as having a disease by image interpreters. On the other hand, the specificity measures the probability that the patient who truly does not have the disease is determined as not having the disease by image interpreters.

In ROC, the results of imaging system are compared to the true disease status of the patient to evaluate the accuracy of that imaging system. There are several types and approaches of ROC analysis methods such as ROC curve, multiple-reader multiple-case, and free-response ROC analysis. ROC and ROC-related methods are considered as the gold standard for image quality evaluation, particularly when evaluating the accuracy of imaging system or comparing different imaging modalities in terms of detectability of specific pathology.

However, applying the ROC method and related types requires a large number of cases and therefore such methods may be cumbersome. Observers, even experienced radiologists, may behave differently in an experimental environment compared with that in a clinical environment. Hence, the reliability of ROC and related methods are relatively low.

Visual grading characteristic

The common clinical based evaluation method of image quality is visual grading characteristic (VGC). The theory of this method, which
is based on the ability to detect and perceive pathology, correlates well with precise anatomical demonstration. There are two ways to perform VGC, namely relative grading and absolute grading approaches. In the relative grading approach, one or several reference images are used by observers to evaluate the quality of images. The observers compare the display quality of the image being assessed with the matching landmark of the reference image. The observers categorise their decisions to a scale of 3, 5, or 7 points. For example, a 5 points scale is +2 = much better, +1 = slightly better, 0 = equal, -1 = slightly worse and -2 much worse.\(^{35}\)

In the absolute grading approach, the observers state their decisions on the visibility of specific features in the assessed image without using a reference image to evaluate the image quality. Typical grading scale of this approach ranges from 4 to 7 points to categorise an observer’s decisions. This feature is also called quality criteria.\(^{52}\) For example, five point grading scale analysis of absolute approach includes excellent image quality (no limitations for clinical use), good image quality (minimal limitations for clinical use), sufficient image quality (moderate limitations for clinical use but no considerable loss of information), restricted image quality (relevant limitations for clinical use, clear loss of information), and finally poor image quality (image must be repeated because of information loss).\(^{31}\)

Quality criteria for each radiologic examination are used to evaluate image quality of specific examination. These criteria, developed by professional radiologists, technologists and physicists, describe physical and anatomical characteristics of image appearance and dose level. For example, chest examination criteria are used to evaluate chest images by letting experienced radiologists and technologist to determine the level of fulfilling these criteria in that image.\(^{30}\) This method has several advantages which make it preferable, but again it still has some limitations.\(^{21}\)

Several factors make this method useful. First, almost all process components of imaging system which control image quality are considered in the evaluation procedures of VGC. These components include image processing, recording, post processing, and reading by expert radiologist. Therefore, the practical validity of this method is considered high. Second, VGC is based on the visualisation of clinically relevant available standards to evaluate image quality. The conducting process of VGC is similar to that of daily clinical situation. Third, this evaluation method has easier procedures and makes for less work than some other methods such as ROC. Furthermore, the required time that the observers are required to read the images is reasonable and therefore there is no real barrier with this regard to have participants.\(^{12}\) This method can also be used to compare the performance of different imaging modalities in terms of image quality and dose level.\(^{35}\)

The limitations of these methods include false positive fractions of limited or no clinical relevance. Furthermore, fulfilled criteria that are judged by the observer may correspond to an unacceptable image.\(^{22}\) Additionally, there are difficulties in analysing the uncertain data from VGC. Hence, the underlying reasons of the uncertainty cannot be identified whether these reasons are related to poor image quality, observer influences or other factors.\(^{51}\) In addition, VGC suffers from the subjectivity of observers which minimises its reliability.\(^{25}\)

According to the above discussion, evaluation methods related to pure statistical measurement such as DQE has a low validity when used to measure the clinical performance of an imaging system unless complete imaging procedures, including image processing, display and the response of the observer, are considered.\(^{38}\) However, DQE is the most effective evaluation method for objectively assessing the performance of the detectors of imaging systems.\(^{7}\) Even though, the reliability and validity of DQE is high in providing accurate measurement of the ability of information transfer, its validity is low in assessing an entire imaging system.\(^{25}\)

On the other hand, methods that involve human observers such as ROC and VGC are valid for evaluating an entire imaging system. Even though, ROC and VGC methods are also considered as gold standard for evaluating the accuracy of imaging system or/and comparing different imaging modalities, their reliability is limited because they suffer from the subjectivity of observers (Figure 5).\(^{23,30,31}\)

Tapiovaara\(^{31}\) investigated the relationship between the results of different procedures of image quality evaluation including physical quantities measurements, phantom experiment evaluations and clinical performance assessment. It was found that this relationship is not clear and not fully understood.\(^{13}\) Until recently, there has been no image quality method to resolve the gap among methods of physical measurements, CDA, and human observers, despite continuous studies and heavy efforts.\(^{35}\)

**Conclusion**

The relationship between the quality parameters of digital radiographic images including resolution (spatial resolution and contrast resolution), noise, and artefacts is complicated, meaning that there is a trade-off between them, improving one parameter may deteriorate another. Hence, optimising these parameters is not a simple task. Optimising image quality parameters in regard to radiation dose make it a more complicated task. Additionally, the effect levels of these parameters on image quality of different digital radiography systems and units are not exactly the same even though they share the principles of image quality parameters. The only way to optimise image quality parameters while maintaining low
radiation dose is to deeply understand the effects of these parameters on each other, the influence factors and their impact on the radiation dose for each different digital radiographic systems. Each of the available evaluation methods has its own advantages and limitations. Therefore each evaluation method should be utilised and employed according to its aptitudes to improve image quality and imaging process.

References


