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PHYSICS

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Towards image quality, beam energy and effective dose optimisation in digital thoracic radiography

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Abstract This paper outlines how objective measurements of both image quality, in terms of signal-to-noise ratio, and effective dose may be used as tools to find the optimum kVp range for a digital chest radiography system. Measurements were made with Thoravision, an amorphous selenium-based digital chest X-ray system. The entrance surface dose and the effective dose to an anthropomorphic chest phantom were determined demonstrating how effective dose is related to beam quality. The image quality was measured using detective quantum efficiency, threshold contrast and a radiologist preference trial involving 100 patients. The results show that, despite the fact that the entrance surface dose decreases as the kVp increases, the effective dose, a better measure of the risk, reaches a minimum value between 90 and 110 kVp; however, the image quality decreases as the kVp increases. In this study the optimum kVp for chest radiography, using a selenium-based radiography system, is in the range 90–110 kVp. This is contrary to the 120- to 150-kVp range that is commonly used. Also, this study shows how objective measurements can be used to optimise radiographic technique without prolonged patient trials.

Key words Dose optimisation · Digital radiography · Image quality · Radiation dose · Radiation protection

Introduction

In recent years numerous national and international surveys have demonstrated large variations in the radiation dose being received by patients for the same examinations. The work of the UK National Radiation Protection Board (NRPB) and the Institute of Physical Sciences of Medicine (IPSM) [1, 2] led to the Council of the European Communities (CEC) guidelines in 1996 [3]. These guidelines recommend appropriate techniques and reference doses for a large variety of standard diagnostic radiology procedures. The reference doses, defined empirically from the survey results, act as a trigger for further investigation should they be exceeded. Experience has shown that the use of reference doses has been instrumental in betraying poor equipment and techniques in hospitals throughout Europe [4]; however, in setting the reference doses and recommending radiographic techniques, little effort has been made to achieve true optimisation.

The technology used in plain film radiology is no longer confined to silver halide film detectors. A large variety of digital detectors are becoming available. Digital radiography systems are not constrained to a narrow dynamic range, as film-based systems are; therefore, there is no reason why the techniques that have been found to be the optimum for screen film should be the same for digital detectors. Departments that introduce digital technology need to undertake an optimisation exercise. There has been much work undertaken to find optimised techniques for the new generation of digital image detectors. Experience has

shown that this is often a controversial process, where subjective opinion accounts for at least as much as hard evidence. The end result is rarely satisfactory. The main problem is the lack of availability of any objective measure of image quality with sufficient relevance to the clinical task.

The aim of this paper is to show how objective measures of image quality and dose may be utilised in an attempt to optimise the balance between kVp (radiographic technique) and the effective dose. The system under consideration is Thoravision (Philips Medical Systems, Hamburg, Germany). Thoravision is a dedicated chest radiography system based on an amorphous selenium detector [5].

Methods

Objective image-quality measurements

Experiments designed to discover which characteristics determine the visibility of details within an image, i.e. the image quality, have found that the single most important measure is the signal-to-noise ratio (SNR). Rose found empirically that for a detail to be visible the SNR must exceed a threshold [6, 7]. The Rose model is the principle underlying the routinely used threshold contrast detail detectability (TCDD) measures [8].

It is possible to measure image quality with threshold contrast detail test objects; however, the variability of human observers tends to obscure differences in image quality and limit the usefulness of the technique [9]. The tests may only be used to consolidate the results from other measurements. An alternative approach outlined by the International Commission of Radiological Units and Measurements (ICRU) is to measure the SNR properties objectively [9, 10, 11]. Three metrics are described which characterise an imaging system: the modulation transfer function (MTF); the noise equivalent quanta (NEQ); and the detective quantum efficiency (DQE). Of these metrics the DQE gives the most useful overall measure of image quality. A full description of these are in the references listed.

Thoravision has been evaluated at this laboratory using the physical imaging characteristics described [12, 13]. The image quality, in terms of DQE, was determined using an X-ray beam emulating a real patient, with 20 mm of additional Al filtration, in a range of kVp values between 60 and 150 kVp. The results demonstrated that Thoravision has significant advantages and differences over conventional film-screen systems.

In addition to the objective measures, the TCDD response has also been determined as part of the previous evaluation [12, 13]. The TCDD was measured using four experienced observers analysing four images at each kVp and dose level of interest. The observed threshold contrast values have been converted to the detection index. The detection index, defined as the inverse product of the threshold contrast and the square root of the detail area [8], allows the straightforward comparison of TCDD curves. The detection index assumes that in an ideal system the image quality is independent of detail size and is directly related to the SNR of the output image. Therefore, an ideal imaging system and observer would, in theory, produce a constant detection index for all detail sizes. The fact that the detection index is not constant is probably due to deficiencies in both the imaging system and human visual system. The higher the detection index the better the system.

To assess the clinical significance of these measures further studies have been undertaken both with real patients and an anthropomorphic chest phantom.

Clinical trials

The effects of varying the radiographic factors were evaluated in a clinical trial. Receiver operating characteristics (ROC) studies [14] are regarded as the most definitive method available in assessing the sensitivity and specificity of a radiographic system; however, ROC studies are highly demanding of resources, so as a first step radiologists were asked to assess normal clinical images produced using the radiographic techniques investigated. Two prospective comparisons were made:

1. The effect of reducing the kVp: images were obtained at 150 and 90 kVp with the same automatic exposure control (AEC) setting.
2. Varying the dose: with the kVp fixed at 90 kVp, one of two AEC settings, either the standard- or low-dose setting, were randomly allocated. The AEC was calibrated so that the low dose setting was half that of the standard dose.

The assessment took the form of a rating trial. A group of four radiologists, three consultants and one senior registrar were asked to complete a simple form during routine film reporting sessions. No radiographic technique data was displayed on the hardcopy images (Dryview 8700, Imaton) enabling a true blind comparison to be made. For each image the radiologist simply rated six regions of the image: trachea and main bronchi; lung parenchyma; bony thoracic cage; costophrenic angles; subphrenic angles; and the overall impression. Each region was rated as inadequate, adequate or good. For each technique being investigated 100 images were collected and rated. The radiographs were obtained using patients undergoing routine clinical examinations. No patients were exposed twice unless the radiographer judged an image as being inadequate. This follows normal clinical procedure.

The results obtained, although they cannot be seen as a measure of image quality, do show the radiologist's preferences. This approach is similar to that taken in the European guidelines [3]. The method assumes that a radiologist is able to assess whether an image contains all the information required to make a confident diagnosis.

Dose measurement

To measure the dose in terms of radiation beam quality an anthropomorphic chest phantom was used (Gothic Crellon, Woking, UK). All clinical exposures made with the Thoravision are controlled by an AEC. The AEC monitors the charge decrease on the selenium layer during the exposure, and when the charge drops to a preset level the exposure is terminated. This method has the inherent advantage that the response of the actual detector is measured, and not that of an independent measuring chamber, as in standard AEC designs. The phantom has the advantage that any number of exposures is possible. There may be small differences evident between the absolute dose values measured with the phantom and in a patient dose survey; however, the relative difference between doses measured with different beam conditions should be similar.

Exposures of the anthropomorphic phantom were used to measure how the dose required by the AEC varied with the kVp. The phantom was positioned for a standard posteroanterior chest exposure. The entrance surface dose (ESD) was then measured,

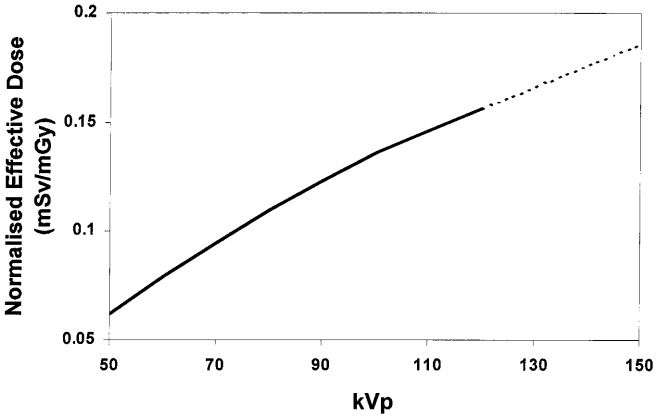

Fig. 1 The relationship between the normalised effective dose and kVp (NRPB data)

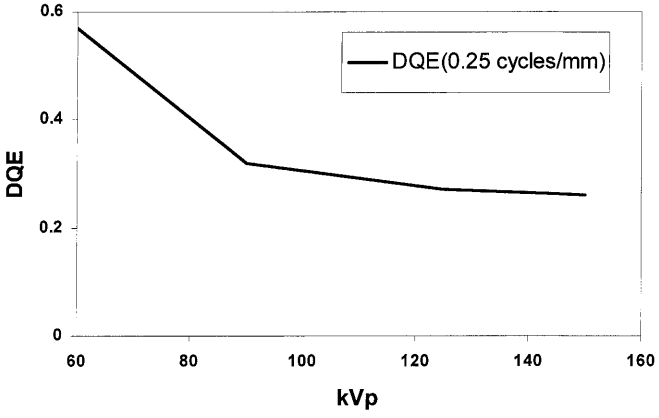

Fig. 2 The relationship between DQE and kVp (0.25 cycles mm^{-1})

including backscatter, using an ionisation chamber (DALi type 77217 dosimeter, with type 77334 ionisation chamber, PTW, Freiburg, Germany), calibrated by the manufacturer. The ESD was measured for a series of kVp settings between 70 and 150 kVp using the automatic exposure control and standard clinical beam quality (2.5 mm Al).

The ESD does not, however, give a good measure of relative risk. Also, ESD values cannot be directly compared when measured with different X-ray beam qualities and projections. An alternative method has been developed known as the effective dose (sievert, Sv) [15]. The effective dose is a weighted sum of the dose equivalents received by each organ of interest. The dose equivalents are calculated using Monte Carlo techniques while the weighting factors are defined by the International Commission of Radiation Protection [15]; therefore, the effective dose cannot be measured directly. In practice, effective dose can be calculated using published tables or computerised algorithms.

In this study the effective dose was calculated using data published by the NRPB [16, 17] together with suitable software (XDOSE, by J.C. Le Heron, National Radiation Laboratory, Christchurch, New Zealand). The calculated normalised effective dose data, i.e. mSv/mGy, are displayed in Fig. 1; however, the NRPB Monte Carlo data only covers the range 50–120 kVp, and the results have been extrapolated to 150 kVp (Fig. 1). The extrapolation is a simple logarithmic curve fit. The logarithmic curve fit matches the calculated data very closely ($r^2 = 0.99$), so it assumed that the extrapolation is a reasonable estimate of the true behaviour of the effective dose with kVp. From the measured ESD data and the calculated normalised effective dose data it is straightforward to calculate the effective dose from the ESD. All calculations made with the extrapolated data are clearly labelled as such. This conversion is necessary since the effective dose is regarded as a more appropriate indicator of the risks associated with different kVp settings compared with a simple measure of ESD.

Results

SNR properties

The data in Fig. 2 and Table 1 show how the DQE varies with kVp over the full clinically relevant range. For clarity, a single point is selected from each DQE spectrum (0.25 cycles mm^{-1}). This represents the most clinically useful part of the spatial frequency spectrum [18]. The results show a marked decrease in DQE as the kVp is increased. This response can be attributed to the reduced X-ray absorption by the selenium layer. Comparing the results with the calculated absorption using published X-ray spectra [19] show that, throughout the clinical range, the DQE is approximately 85 % of the X-ray absorption. These results mean that to maintain a constant SNR as the kVp is increased would require an increase in the number of X-ray quanta and therefore dose.

Table 1 Detective quantum efficiency (DQE) values (0.25 cycles/mm)

kVp	DQE(0.25)
60	0.57
90	0.32
125	0.27
150	0.26

The differences in DQE, and therefore SNR, are further confirmed with the TCDD responses. The TCDD responses calculated at both 90 and 125 kVp are plotted in Fig. 3 as detection index curves. Although no statistical difference can be shown, due to unavoidable errors in measurement [20], the increased detection index is in agreement with the DQE results.

Radiologist preference

The results of the clinical preference study are summarised in Table 2. The results were analysed with a chi-squared test to ascertain whether there was any statistical difference between radiographic techniques, setting the confidence level at 95 %. These show that when

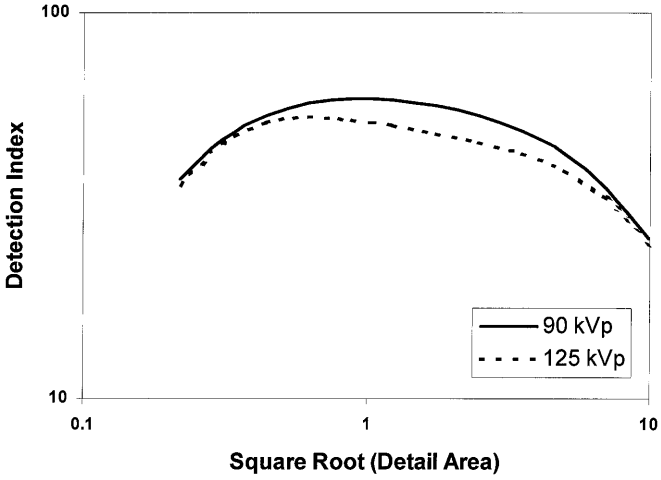

Fig. 3 The threshold contrast detail detectability response of Thoravision at 125 and 90 kVp

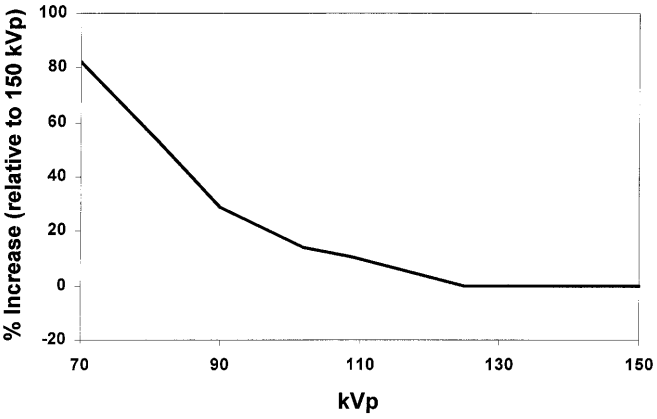

Fig. 4 The relationship between entrance surface dose and kVp

Table 2 Radiologist's preferences

Region evaluated	150 or 90 kVp	Standard or low dose
Trachea		Standard
Lung parenchyma	90	Standard
Bony thoracic cage	90	Standard
Costophrenic angles		Standard
Subphrenic angles	90	Standard
Overall		Standard

comparing kVp at the same dose, the 90 kVp images were preferred for three of six regions. When comparing dose, at the same kVp, the higher dose was unambiguously the preferred option at both 150 and 90 kVp. This does not mean that the lower dose is inadequate, only that radiologists were able to tell the difference in a blind study. The result gives some credibility to the technique being used to measure "image quality". It was also noted that when comparing 150-kVp images to images obtained at 90 kVp at half the ESD, the 150-kVp images were only preferred in two of six regions (trachea and costophrenic angles).

These results suggest that the lower kVp (90 kVp) has advantages in terms of perceived image quality over the generally accepted higher kVp (150 kVp) which is becoming the standard technique in Europe and the U.S. These results are in agreement with the findings of the DQE analysis of the system.

Effective dose and optimisation

The results from the ESD measurements are plotted in Fig. 4. As expected, the ESD decreases with kVp. This is the argument used to persuade radiology departments to use the higher kVp settings. However, the calculated normalised effective dose data, plotted in Fig. 1, clearly shows that the effective dose increases with kVp in contrast to the measured ESD. This is in agreement with data published by Huda and Bissessur [21]. Simply stated, this suggests that an ESD of 1 mGy at 70 kVp carries less risk than 1 mGy at 120 kVp; however, as Fig. 4 shows, the ESD needed at 70 kVp is greater than the ESD at 120 kVp.

From the measured ESD and the normalised effective dose data, it is possible to calculate the effective dose as in Table 3 plotted in Fig. 5. An inspection of the range of effective dose values shows a small range of values. This curve shows that the large range of ESD (70 kVp requiring 80% greater dose compared with 120 kVp) is much reduced, so that the effective dose at 70 kVp is now only 10% greater than the effective dose at 120 kVp. It is interesting note that the effective dose reaches a minimum value at around 110 kVp. The effective dose at 110 kVp is, if the interpolation is correct, significantly less than the effective dose at 150 kVp.

Taken together, these results show that there are few advantages to using kVp values greater than approximately 110 kVp. The optimum radiographic technique, in terms of kVp, is a compromise between the increased SNR available at low kVp and the effective dose.

Discussion

These results indicate that the image quality when using Thoravision, measured both objectively and in a preference study, is improved at a lower kVp (90–110 kVp) range when compared with the normally recommended high kVp (150 kVp) technique.

This is in conflict with results published by Chotas et al. [22] based on a photostimulable phosphor digital radiography system (CR). Their results showed that for

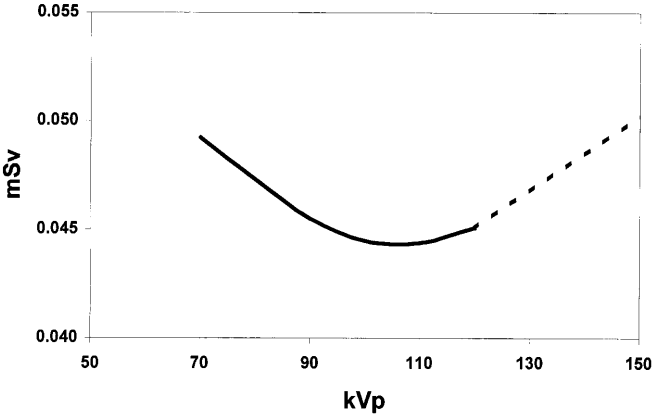

Fig. 5 The relationship between effective dose and kVp

Table 3 Calculated effective dose

kVp	mSv
70	0.0492
80	0.0473
90	0.0455
100	0.0445
120	0.0451
150	0.0503

matched effective dose the SNR in the lung areas decreased as the kVp increased, whereas the SNR in the mediastinum the SNR was independent of kVp. Other results made with CR have shown no dependence on the SNR with regard to kVp [20], whereas the results reported here indicate that there is an optimum region between 100 and 110 kVp for Thoravision. The biggest difference between the studies is the imaging system. The SNR results for Thoravision indicate that it has significant advantages in terms of SNR over CR. In particular, Thoravision is able to maintain reasonable DQE over a much greater exposure range compared with CR [12, 23].

The main reasons for the improved image quality as kVp is decreased is the increase X-ray absorption by the selenium detector and the higher subject contrast available. This characteristic of X-ray imaging systems has led practitioners to use a low kVp when high contrast is needed. Conventional film systems are constrained by their dynamic range, so the increased contrast can be a hindrance. The relatively low SNR of CR conceals the advantages of lower kVp.

It is important to note that the DQE clearly reflects the improvement in image quality as the kVp is decreased; however, when using DQE to assess image quality great care must be used to verify that the X-ray conditions used (kVp, filtration and dose) emulate clinical conditions. A system may show near 100% DQE at low kVp and high dose.

The major weakness of DQE as a measure is that it does not aid in determining the minimal acceptable dose. The only robust method available for this purpose would be to use an ROC study. Chest radiography does not carry a large risk when compared with other radiographic examinations. The primary reason that chest radiography is the subject of so many investigations is the frequency with which it is performed. With the large number of patients available it would be feasible to select a subgroup of patients for whom alternative data confirming the X-ray diagnosis is available. This gold standard would then allow a full ROC study to be performed to properly validate the objective and subjective findings of this study.

This study has clearly demonstrated that the use of entrance surface dose as a measure of risk can be misleading. When calculating the effective dose with an anthropomorphic chest phantom, it was found that the decrease in measured ESD as the kVp was increased did not translate into such large differences in effective dose. In fact, a minimum effective dose was found at approximately 110 kVp.

Conclusion

These results, although only pertinent to the Thoravision system, show how objective measures can be applied to other technologies. Digital detectors have an inherent advantage over traditional film-based systems in this respect. Whereas measuring the DQE of a film involved using time-consuming specialised microdensitometer scanning equipment to sample the data, digital data is already sampled and ready to process. In the past DQE was used only to compare systems. Now it is possible to obtain far more results and, therefore, extend their use to optimisation. For example, how does filtration affect the image quality and dose?

In an ideal world it would be possible to state the minimum acceptable SNR for each radiographic examination. With modern computing power and digital detectors it is conceivable that SNR measurements could be made on a more routine basis. Appropriate use of such measures in combination with an appropriate measure of the associated risks would allow a more pragmatic approach to the problem of optimisation.

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References

1. Shrimpton PC, Wall BF, Jones DG, Fisher ES, Hillier MC, Kendal GM (1986) A national survey of doses to patients undergoing a selection of routine X-ray examinations in English hospitals. NRPB-R200, HMSO, London
2. IPSM working party (1992) Dosimetry working party of the institute of physical sciences in medicine. National protocol for patient dose measurements in diagnostic radiology. NRPB, HMSO, London
3. Council of the European Communities (1996) European guidelines on quality criteria for diagnostic radiographic images, Rep. EUR 16260, EN 1, ISBN 92–827–7284–5
4. Hart D, Hillier MC, Wall BF, Shrimpton PC, Bungay D (1996) Doses to patients from medical X-ray examinations in the UK – 1995 review. NRPB-R289, HMSO, London
5. Neitzel U, Maack I, Gunter-Kohfahl S (1994) Image quality of a digital chest radiography system based on a selenium detector. Med Phys 21:509–516
6. Rose A (1948) The sensitivity performance of the human eye on an absolute scale. J Optical Soc Am 39:196–208
7. Burgess AE (1999) The Rose Model, revisted. J Optical Soc Am 16:633–646
8. Cowen AR (1991) Digital X-ray imaging. Meas Sci Technol 2:691–707
9. ICRU (1996) Report 54. Medical imaging: the assessment of image quality. International Commission of Radiological Units and Measurements
10. Wagner RF, Brown DG (1985) Unified SNR analysis of medical imaging systems. Phys Med Biol 30:489–518
11. Giger ML, Doi K, Metz CE (1984) Investigation of basic imaging properties in digital radiography, vol 2. Noise wiener spectrum. Med Phys 11:797–805
12. Launders JH, Kengyelics SM, Cowen AR (1998) A comprehensive physical image quality evaluation of a selenium-based digital X-ray imaging system for thorax radiography. Med Phys 25:986–997
13. Launders JH, Cowen AR, Bury RF, Hawkridge P, Belford L, Jevon G (1996) An evaluation of the clinical and technical performance of Philips Medical Systems Thoravision chest X-ray system. Department of Health Evaluation Report. MDA/96/31, HMSO, London
14. Metz CE (1986) ROC methodology in radiological imaging. Invest Radiol 21:720–733
15. ICRP (1991) 1990 recommendations of the International Commission of Radiation Protection. ICRP Publ 60, Ann ICRP, 21, nos. 1–3
16. Hart D, Jones DG, Wall BF (1995) Software, Normalised organ doses for medical X-ray examinations calculated using Monte Carlo techniques. NRPB-SR262 Chilton, UK
17. Hart D, Jones DG, Wall BF (1994) Estimation of effective dose in diagnostic radiology from entrance surface dose and dose area product measurements.NRPB-R262, HMSO, London
18. Ziskin MC, Revesz HL, Kundel HL, Shea FJ (1971) Spatial frequency spectra of radiographic images. Radiology 98:507–517
19. Birch RM, Marshall GM, Adran GM (1979) Catalogue of spectral data for diagnostic X-rays. Hospital Physicists Association, UK
20. Launders JH, Cowen AR (1995) A comparison of the threshold detail detectability of a screen-film combination and computed radiology under conditions relevant to high kVp chest radiography. Phys Med Biol 40:1393–1398
21. Huda W, Bissessur K (1990) Effective dose equivalents, H_E, in diagnostic radiology. Med Phys 17:998–1003
22. Chotas HG, Floyd CE, Dobbins JT, Ravin CE (1993) Digital chest radiography with photostimulable storage phosphors: signal-to-noise ratio as a function of kilovoltage with matched exposure risk. Radiology 186:395–398
23. Kengyelics SM, Davies AG, Cowen AR (1998) A comparison of the physical imaging properties of Fuji ST-V, ST-VA, and ST-VN computed radiography image plates. Med Phys 25:2163–2169