EUROPEAN ORGANIZATION FOR NUCLEAR RESEARCH

CERN-PPE/93-219
2 December 1993

EVALUATION OF A LOW-DOSE DIGITAL X-RAY SYSTEM WITH IMPROVED SPATIAL RESOLUTION.

A.Martinez-Davalos and R.D.Speller
(Dep.of Med.Phys. and Bioengineering
University College London, London, UK)

D.J.Miller
(Dep.of Phys. and Astronomy
University College London, London, UK)

L.I. Shekhtman
(CERN, Geneva, Switzerland)

S.E.Baru, A.G.Khabakhpashev and O.A.Ponomarev
(Budker INP, Novosibirsk, Russia)

Abstract

We present results from an evaluation of a significantly improved version of the Siberian Digital Radiographic Device (SDRD). The SDRD is already in clinical use in Novosibirsk and Moscow where it produces images of high diagnostic value at considerable lower doses than conventional film-screen combinations. It uses a fast Multiwire Proportional Chamber with highly parallel readout as a detector. Improvements in the readout system have given a reduction by a factor of two in the pixel size. We have used well established methods to compare imaging performance of the new SDRD with the old version as well as with a film-screen combination used in routine clinical practice in the UK. At the same level of exposure the threshold contrast for objects of the order of the pixel size is two times better for the new SDRD than for the old one, and the minimum size of objects visible in the image is two times smaller (0.25 mm) for the new system than for the old SDRD.

Presented at the 3rd London Conference on Position Sensitive Detectors , Brunel University, Uxbridge, 6-10 September 1993

1. Introduction

Digital radiography offers three major advantages over the photographic film-screen techniques which are most commonly used:

i.) The dose can be reduced to the quantum limit
ii.) The images cover a large dynamic range of contrast
iii.) Computerised storage and display techniques can be used

We describe an improved version of a digital radiography system based on a MWPC which is already in clinical use in Russia [1]. Improvements in the readout electronics have reduced the effective pixel dimension by a factor of two. Measurements with a standard phantom of the contrast-detail performance (threshold contrast vs. detail diameter) of the new system show that this gives significantly better position resolution, while maintaining the low dose characteristics of the device, which is between one and two orders of magnitude better than film-screen for some clinical examinations [2].

2. The Digital Radiographic Device

Previous versions of the SDRD have been described elsewhere in some detail [3]. Figure 1 shows a layout of the scanning gantry which carries the main components of the system: the x-ray tube, a pair of slit collimators -one before and one after the patient- and the linear multiwire proportional chamber. The chamber is filled with Xe+ 20% CO2 at 3 atm (absolute). The sensitive area is 384 mm wide by 50 mm deep, giving a conversion efficiency of ~30% for 60 keV x-ray photons. The anode wires are directed to the focal spot of the x-ray tube to avoid parallax errors, and the field electrodes are inclined with respect to the anode plane to give uniform gain over the length of the wire. In order to give high count rate capability, close to the space charge limit, the readout is almost entirely parallel. Each line of the image is exposed for 30 ms while scalers, one per wire in the original version, count pulses from converted x-ray quanta. When the exposure of one line has been transmitted to the computer RAM (PC AT) the data are collected for the next line, giving an overall scan time of about 8 seconds for a complete clinical image. After simple processing the image can be presented within a few seconds from exposure.The wire spacing of approximately 1.2 mm (tapered) in the chamber converts to a horizontal pixel width of approximately 1 mm at the patient in the standard resolution (ST) version. The vertical pixel size (also ~1 mm) is governed by the collimator gap sizes and the linescan time.

3. High resolution (HR) system

If an x-ray quantum is absorbed near the centre between two neighbouring anode wires, the primary ionisation in the majority of cases is divided into two parts, and two channels of the detector count simultaneously. In the original version such events were rejected by a special selection circuit. The new system [4] introduces an extra set of scalers between every pair of wires. Coincidences (within 75 ns) between neighbouring wires are recorded only in these extra scalers, at a rate which is comparable to the single hit rate. The information for each line is stored in 640 16-bit scalers, corresponding to a horizontal pixel size at the patient of approximately 0.5 mm. The vertical pixel size is matched to this by reducing the time for each linescan to 15 ms and adjusting the aperture of the collimators. The non uniformity of response due to the two interleaved types of readout channel can be reduced to less than 1% by a simple normalising algorithm [4] for count rates up to 600 kHz/channel.

4. Experimental Techniques

4.1 The test object

We have used the Leeds test object TO.10 [5] as the basis of this study. The TO.10 is a commercially available phantom originally designed for the assessment of fluoroscopic equipment. It consists of a 1 cm thick perspex plate with embedded objects (details) covering a range of diameters and thicknesses. These objects produce calibrated contrast values when imaged under specific exposure conditions. Figure 2 shows a layout diagram of the details contained in TO.10. Diameters range from 11.1 mm (row A) down to 0.25 mm (row M), and for each diameter there are nine objects of different contrasts. The established method of presenting the results of this test is in the form of Contrast-Detail diagrams, where threshold contrast is plotted as a function of detail diameter.

4.2 Exposure conditions and dosimetry

Exposure conditions for the images used in this study are given in Table 1. Normal exposure conditions in the SDRD include the use of an added filtration of 0.1-0.3 mm Cu (depending on the type of examination) to optimise the x-ray spectrum for the energy response of the detector and reduce the entrance dose to the patient. All the images in this test were taken at 70 kVp with an added filtration of 0.1 mm Cu and 6 cm perspex in order to approximately reproduce the exposure conditions for a thorax examination.

Absorbed dose values were measured at the surface of the perspex block, at ~112 cm from the focal spot of the x-ray tube, using either an air ionisation chamber or an air equivalent plastic scintillator probe. The choice of a particular dosimeter was governed by the sensitivity and availability of each system.

4.3 Evaluation of the images

All the SDRD images were re-displayed for their evaluation on a Sun SPARCstation IPX. The software used allows the operator to window the displayed pixel counts at any point in the full range and to select the grey levels assigned to a given subrange of pixel values. Film-screen images were viewed on standard light boxes.

For the threshold contrast measurements a group of observers viewed the whole set of images under subdued background lighting conditions, and the number of objects of a given size that could be seen was recorded. These values were averaged for all the observers and used along with the dose measurements to produce the Contrast-Detail-Dose diagrams.

The detail signal-to-noise ratio (*dSNR*) was calculated from mean pixel values taken directly from the digital images. The contrast computed from the measured pixel values correlates very well with the nominal contrast values specified by the phantom manufacturers (less than 3% difference at 30 μGy for the HR system).

Figures 3a and 3b show images of a section the TO.10 phantom taken with the ST and the HR versions of the SDRD, respectively. Both images were taken with the same tube voltage and filtration, as specified in section in section 4.3, only changing the tube current to obtain the same entrance dose, which was 20 μGy in this case. No image processing was done in any of these images, apart from overall contrast enhancement. Objects down to 0.5 mm are clearly visible in the high resolution image.

5. Results

5.1 Signal to Noise Ratio measurements

The basic quantity for image formation is the number of x-ray photons per area of the spatial resolution element that are detected at the image plane by the imaging system.

The signal produced by a contrasting detail in the radiological image will correspond to the difference of x-ray photons per pixel detected at different points in the image. If N_o is the average number of counts per pixel over the area of detail and N_b the mean level of counts for the same area of background close to it, then the signal would be $|N_b - N_o|$. The statistical noise associated with the detection of this signal would be $\sqrt{Nb+No}$ assuming Poisson statistics, and therefore the *dSNR* would be given by:

$$dSNR = \frac{|Nb - No|}{\sqrt{Nb+No}}$$

The ultimate limit to the image information content is set by such statistical noise. The imaging system has optimal design if any other source of noise is negligible compared to the statistical noise, and if it detects every photon that interacts with the system [6].

Figure 4a shows the measured *dSNR*s of large (11.1 mm diameter) objects as a function of the entrance dose for the ST and the HR systems. The dotted lines represent the square root behaviour predicted by the model of Motz and Danos [7].

Figure 4b shows the dSNR of both systems at the same level of exposure as a function of detail diameter, for objects with the same subject contrast (27.7%). The ST system gives better SNR for the largest objects, but the SNR decreases more rapidly than for the HR system, which performs better below ~2 mm diameter.

5.2 Contrast-Detail measurements

A direct comparison of the Contrast-Detail performance of the SDRD (ST and HR) and a UK film-screen combination is shown in figure 5. All three images were taken at the same level of exposure (20 μGy entrance absorbed dose). It can be seen that both versions of the SDRD have better contrast sensitivity than the film-screen system for medium and large low-contrast objects. On the other hand, film-screen offers better detectability of small high-contrast objects due to its higher intrinsic resolution.

The HR system performs better than the ST system for all but the largest details. Threshold contrast sensitivity is approximately two times better for objects below 2 mm diameter, and at this level of exposure was able to detect details down to 0.35 mm diameter.

6. Conclusions

The results of the present work show that the method of separate counting of coincident hits in the MWPC permits the number of pixels in the digital image to be increased by a factor of 4 and correspondingly allows the effective pixel dimension to be reduced by a factor of 2 without significant complication of the system.

At the same level of entrance dose, image quality was better for the high resolution system as compared to the standard resolution system. That is, the threshold contrast was as much as two times lower in the new system for objects of the order of the pixel size, despite of the lower pixel statistics.

References

[1] E.A. Babichev et al, *Physica Medica* **3** (1991) 113-117

[2] A. Martinez-Davalos et al, to be published in *Phys. Med. and Biol.*

[3] E.A. Babichev et al, *Nucl. Instr. and Meth.* **A310** (1991) 449-454

[4] E.A. Babichev et al, *Nucl. Instr. and Meth.* **A323** (1992) 49-53

[5] G.A. Hay et al, *Br. J. Radiol.* **58** (1985) 335-344

[6] M.J. Tapiovaara and R.F. Wagner, *Phys. Med. Biol.* **30** (1985) 519-529

[7] J.W. Motz and M. Danos, *Medical Physics* **5** (1978) 8-22

Tables

Table 1. Exposure conditions for the images used in this comparison. All images were taken at 70 kVp with 0.1 mm Cu + 6 cm Perspex added filtration.

Image	System	Exposure settings	Entrance Dose (μGy)	N_b
1	ST	10 mA	21.0 ± 2.1	7135
2	HR	25 mA	20.0 ± 2.0	2343
3	f-s	1.6 mAs	20.0 ± 2.0	N/A

Figure captions

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

Figure 2. Layout diagram of the Leeds TO.10 phantom.

Figure 3. Section of the TO.10 phantom imaged with the ST system (a) and the HR system (b) under the same exposure conditions (20 μGy entrance dose). Objects down to 0.5 mm diameter are clearly visible in the HR image.

Figure 4. **a.)** Comparison of the *dSNR* produced by object A1 (11.1 mm diameter, 27.7% contrast) as a function of the entrance dose for the ST and the HR systems. The dotted lines represent the square root behaviour predicted by the model of Motz and Danos [7].

b.) *dSNR* as a function of detail diameter for ST and HR. The HR system performs better for objects of less than 2 mm diameter.

Figure 5. Contrast-Detail curves for the SDRD (ST and HR) and a UK film-screen combination (f-s). All three images were taken under the same exposure conditions (20 μGy entrance dose).

Figure 6. Digital image of a thorax taken with the HR system. Exposure conditions were 70 kVp, 50 mA and 0.3 mm Cu added filtration, which gave an entrance skin dose of ~30 μGy.

Figure 1.

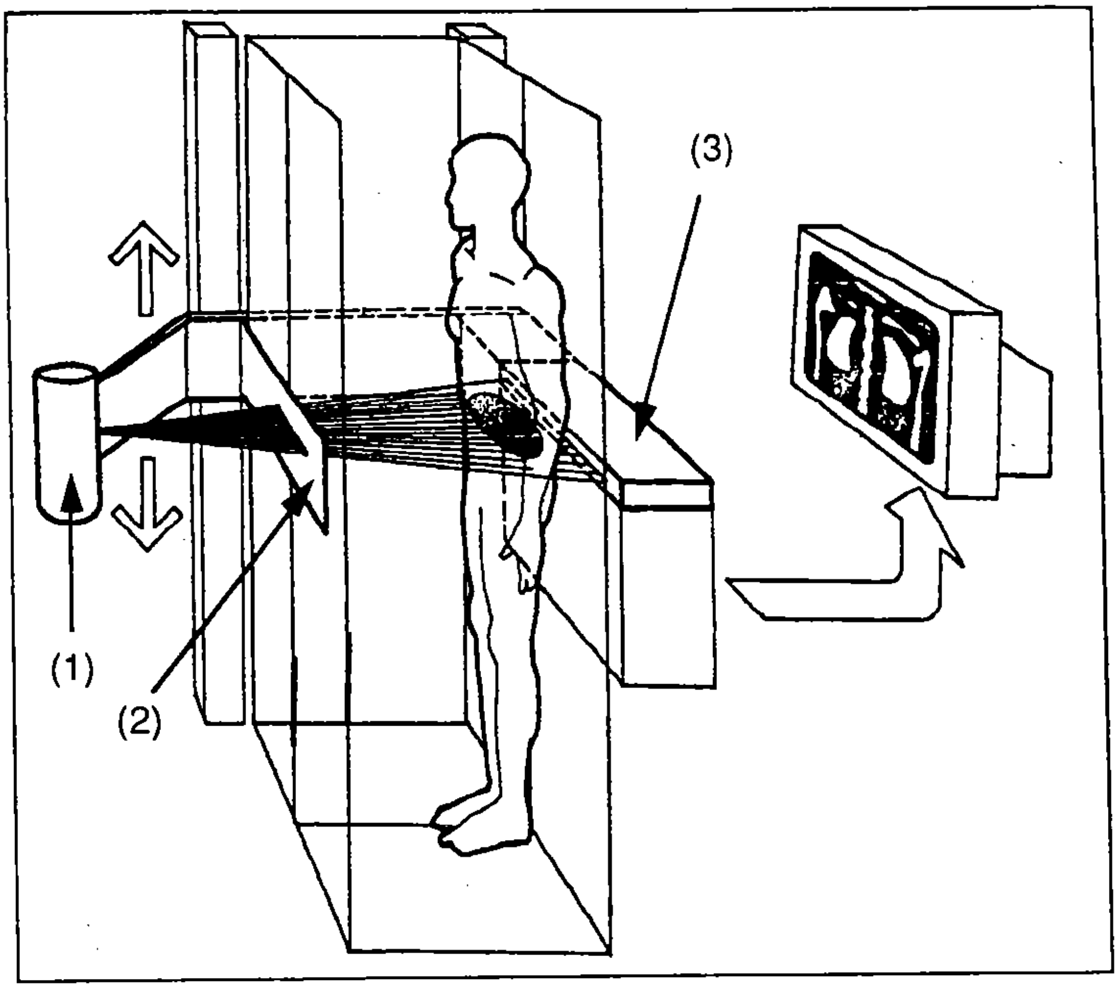


Figure 2.

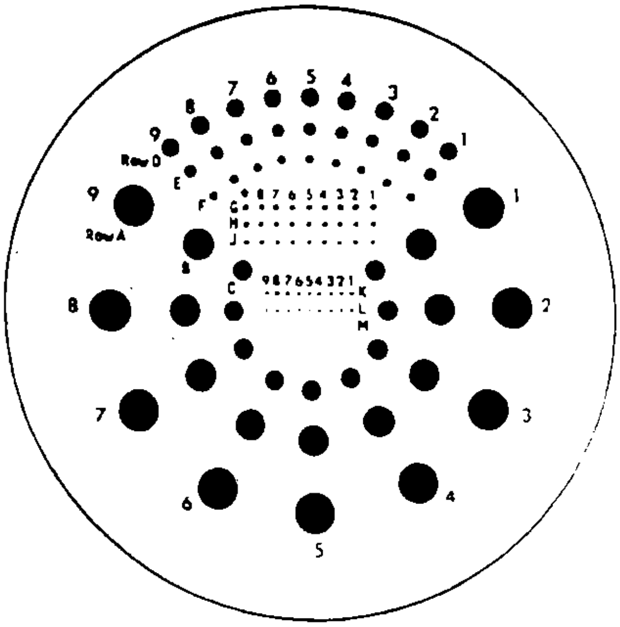
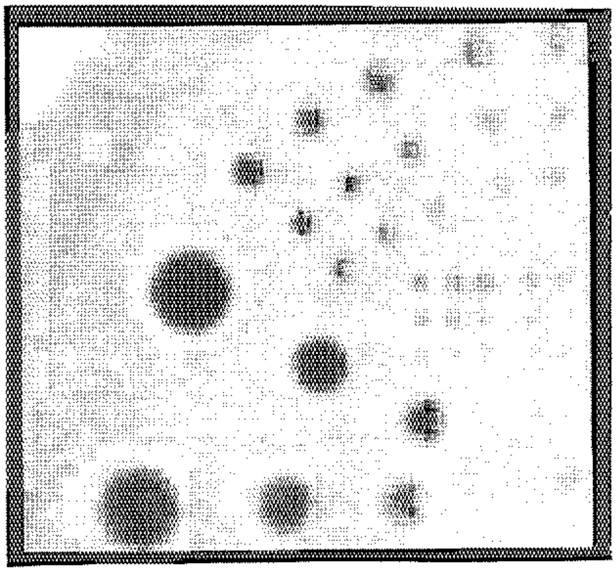


Figure 3.

a.)



b.)

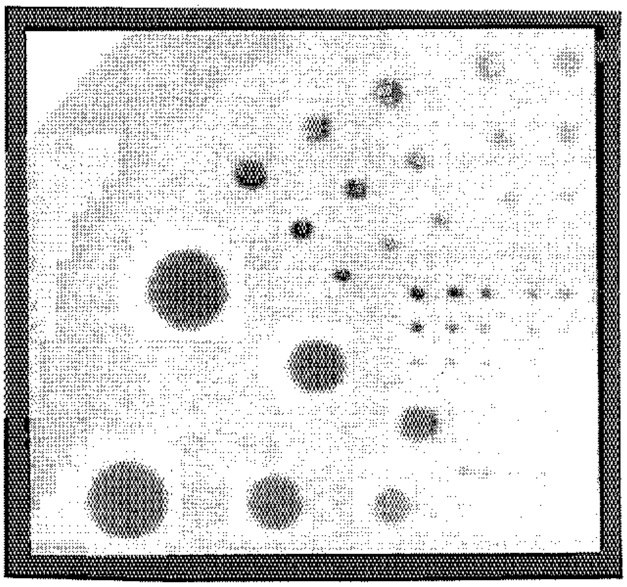


Figure 4.

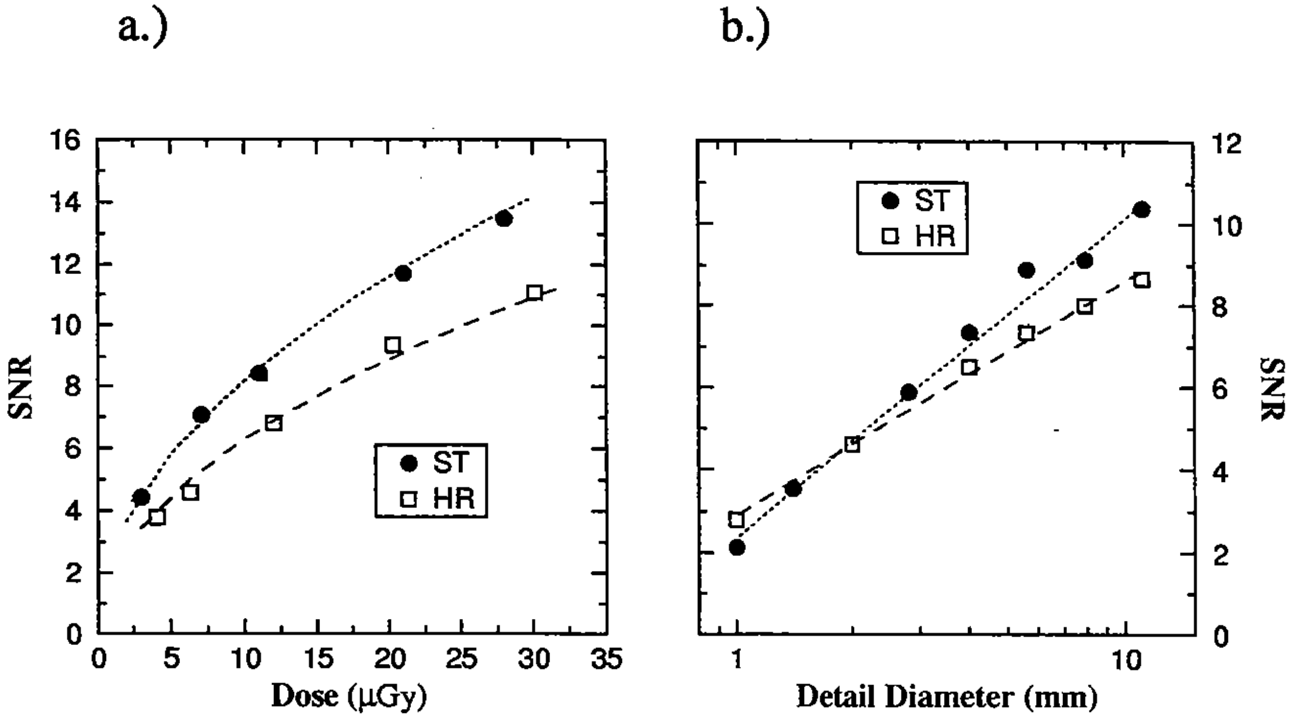


Figure 5.

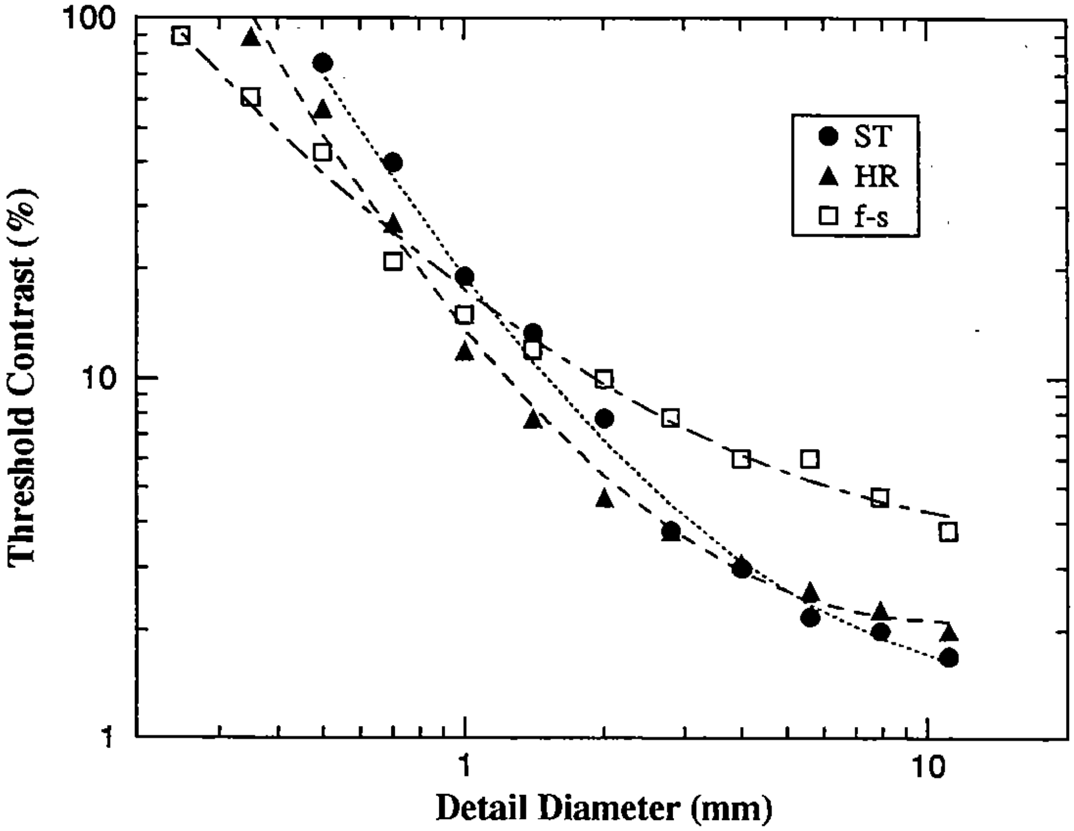


Figure 6.