Comparison of Low-Contrast Detail Perception on Storage Phosphor Radiographs and Digital Flat Panel Detector Images

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Abstract-A contrast detail analysis was performed to compare perception of low-contrast details on X-ray images derived from digital storage phosphor radiography and from a flat panel detector system based on a cesium iodide/amorphous silicon matrix.

The CDRAD 2.0 phantom was used to perform a comparative contrast detail analysis of a clinical storage phosphor radiography system and an indirect type digital flat panel detector unit. Images were acquired at exposure levels comparable to film speeds of 50/100/200/400 and 800. Four observers evaluated a total of 50 films with respect to the threshold contrast for each detail size. The numbers of correctly identified objects were determined for all image subsets.

The overall results show that low-contrast detail perception with digital flat panel detector images is better than with state of the art storage phosphor screens. This is especially true for the low-exposure setting, where a nearly 10 % higher correct observation ratio is reached. Given its high detective quantum efficiency the digital flat panel technology based on the cesium iodide scintillator/amorphous silicon matrix is best suited for detection of low-contrast detail structures, which shows its high Potential for clinical imaging.

Index Terms-Digital radiography, low-contrast perception, observer performance.

1. INTRODUCTION

In the last years, there was a rapid evolution of digital radiography: Digital storage phosphor plate computed radiography (CR) systems and dedicated chest systems based on solid-state selenium detectors are gradually replacing film screen radiography in clinical routine as both systems proved to produce excellent image quality [1], [2]. As a next step in the development of digital radiography systems large area electronic flat panel detectors based on either scintillators or photoconductors as X-ray conversion material [3]-[6] have been introduced into clinical routine recently. These detectors have several promising features compared to conventional film-screen systems and computed radiography, like the elimination of a separate device for image read out, faster availability of images with enhanced patient throughput, electronic image transmission and display. Besides their technical and logistical advantages these new technologies have to be judged in regard to image quality compared to established imaging techniques.

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Image quality of different acquisition systems may be defined using objective measures like the signal-to-noise ratio (SNR), modulation transfer function (MTF), etc. but medical diagnosis is not made by the image alone, also the perception by the observer is crucial for the result. A number of studies have shown the important influence of psychophysical factors on the way, a given image quality is perceived by the observer [7], [8]. With the use of a contrast detail phantom a test of the observer's perception is possible on a semi objective basis, as the difference between interpretation of the radiologist and the true distribution of test objects can be calculated [9]. First, contrast-detail studies comparing flat-panel detectors to screen-film imaging have been reported recently [10], [11].

Today, fully digital radiology departments like ours usually rely on storage phosphor radiography which in this setting may be considered as the accepted "gold standard". Therefore, the aim of the study reported here was to compare state-of-the-art storage phosphor radiography and direct digital acquisition with respect to detail recognizability under stable image viewing conditions.

II. MATERIAL AND METHODS

A. Imaging Systems

The digital bucky system Philips Digital Diagnost (Philips Medical Systems, Hamburg, Germany) is based on a 43 cm x 43 cm flat panel detector provided by Trixell (Moirans, France) that consists of a 500- m-thick cesium iodide scintillation layer mounted on a 2941 x 3021 pixel read out matrix made of amorphous silicon (Fig. 1). Each of these elements sized 143 x 143 microns is connected to a tiny switch for readout. With this detector specification a pixel size of 0.143 mm is achieved, with a maximum spatial resolution of about 3.5 lp/mm. After amplification and multiplexing the detected signals are transferred to a dedicated postprocessing workstation. The detailed technical description of the detector and its physical properties is reported elsewhere [12], [13].

A standard digital storage phosphor radiography system AGFA ADC 70 (AGFA medical systems, Belgium) with an attached AGFA-M1- MOSA postprocessing workstation was used for comparison. In this system, the maximum achievable spatial resolution is dependent on the size of the phosphor plate used. We chose a 24 x 30 cm cassette with a pixel size of 0.111 mm. Together with the read out geometry this results in a comparable maximum spatial resolution of 4.2 lp/mm.

Fig. 1. Schematic view of the internal construction of the flat panel detector- a cesium iodide/amorphous silicon matrix mounted on a glass plate.
B. Contrast Detail Phantom

The CDRAD 2.0 phantom shown in Fig. 2 (Nuclear Associates, Carle Place, NY), originally designed at the medical physics department of the Nijmegen University Hospital was used in our study [14]. This phantom consists of a plexiglass tablet (265 x 265 x 10 mm) with a 15 x 15 array of cells with cylindrical drill holes of exact diameter and depth. Both diameter and depth are changed logarithmically from 0.3 to 8.0 mm. Within a row the hole-diameter is constant with exponentially decreasing depth, and within a column the hole-depth is constant with exponentially decreasing diameter. Except for the largest three diameters, the other rows have two identical targets in each square, one in the middle and one in a randomly chosen corner of the cell. This allows a four alternative forced choice protocol to be applied for verification of object perception [8].

C. Image Preparation

A total of 50 images (25 storage phosphor and direct digital radiographs each, five images at each exposure level) were acquired at 70 kVp with the Philips Optimus 80 X-ray generator and Philips Rotalix X-ray tube at a focus/object (phantom) distance of 184 cm, a small size focus of 0.6 and an additional filtration of 21 mm aluminum (radiation quality RQA5 according to IEC 1267). X-ray entrance dose (air kerma) was measured in the central beam with the DIADOS 2 dosimeter (PTW, Freiburg, Germany) at the phantom surface. Images were acquired at five exposure levels from 1.3 to 19.3 Gy (detailed parameters are given in Table I).

All images were postprocessor at the dedicated AGFA or Philips workstation respectively and a comparable look-up table and standard skeletal postprocessing algorithm including subtle edge enhancement was applied to reach a comparable image impression and grayscale representation [15]. All images were printed on laserfilm for evaluation on a film changer under equal light and film reading conditions.

![Fig. 2. Schematic drawing of the CDRAD 2.0 phantom (left) and X-ray image (right).](image-url)
D. Image Evaluation

Images were presented in random order with blinding of modality and exposure data and under consistent light conditions at the same film changer. Four radiologists (two senior staff radiologists and two residents in their fourth year of education) viewed the images independently, assisted by the conductor of the study, who directly entered their answers into a PC-based evaluation sheet. This MS-Excel based evaluation automatically corrects observer ratings according to the correction scheme described in the CDRAD phantom manual [14], [16]. According to this scheme each correctly identified target needs two or more correctly identified adjacent targets to remain true - each incorrectly identified target is considered true if it has three of four correctly detected nearest neighbors. Incidental errors of image viewing are, thus, eliminated.

E. Statistics

Before a comparison of the two systems was attempted the detection thresholds for the observers were compared. As there were only subtle differences between the ratings of different radiologists the results were averaged over all observers. To calculate the true difference in image quality between the two imaging systems at each exposure level, a one-way analysis of variance was performed. The question, if the use of a flat panel detector system may lead to a Potential decrease of patient radiation exposure while preserving the same image quality was addressed by comparing the flat panel ratings with the ratings of the storage phosphor system at the next higher exposure level.

III. RESULTS

The standard deviation in the number of correctly identified targets for the readings of the five repeated images with the same exposure ranged from 1% to 6% with the highest values at the low exposure levels. In Fig. 3, an example for the detection thresholds of the different observers for a certain detail size and exposure level is given. There was no statistically significant difference between the ratings of senior staff radiologists and residents.

Fig. 4 gives mean number of correctly identified targets for each image as function of phantom entrance dose. It shows, that for both imaging systems the number of correctly identified objects increases with rising exposure dose. The detection rate for the flat panel detector system is higher than for the storage phosphor system and this difference is slightly more pronounced at low exposure levels.

Table I gives the mean numbers of correctly detected targets averaged over all four observers and the five images obtained at each exposure level, as well as the $p$-values for the true statistical difference between the two imaging systems. It shows a 7.4% higher correct detection rate for the flat panel detector for the 19.3-Gy exposure; differences of 7.9%, 9.0%, 10.7%, and 9.5% were shown for exposures of 9.5 Gy, 4.8 Gy, 2.5 Gy, and 1.3 Gy, respectively. According to Table 1 the numbers of correctly identified objects for the flat panel detector system are equal or slightly higher than the numbers achieved with the storage phosphor system in the next higher speed class (doubled exposure).
Statistical difference for ratings of different speed classes was highly significant within the modality, but not when the flat panel detector ratings were compared with the next lower speed class of the storage phosphor system.

IV. DISCUSSION

In general, radiological practice, the detection of subtle details is one of the most important tasks for the evaluation of many clinical questions (detection of pulmonary nodules in cancer patients, detection of reticulo-nodular changes in patients with interstitial lung disease, etc.) There is a common agreement, that depiction of small, low-contrast detail is, therefore, an important quality aspect of an imaging system [17], [18]. Despite its lower spatial resolution limit compared to conventional film-screen imaging, digital radiography may allow better depiction of low-contrast detail because of its wider dynamic range, better SNR, and image postprocessing capabilities. A contrast detail analysis, as performed in this study, is an accepted semi-objective method to assess the whole chain of low-contrast detail detection, including physical properties of the imaging system and psychophysical aspects of the observer.

Our results show, that detectability of low-contrast objects is significantly better with images derived from a flat panel detector system based on the cesium iodide/amorphous silicon technology compared to storage phosphor radiographs at equal dose levels. This is in good accordance with previous studies reporting similar data for the comparison of indirect type flat panel detectors with film-screen radiography [10], [11].
In clinical routine, this may be beneficial in two ways: The better low-contrast resolution of the flat panel detector may be used to improve detection of disease, which pathophysiologically leads to the appearance of tiny low-contrast structures, like it is the case with interstitial lung disease or rheumatoid arthritis for example. On the other hand radiation exposure to the patient may be lowered substantially - i.e., one speed class according to our data - without loss of diagnostic information.

The good performance of the digital flat panel detector may be explained mainly by two distinct properties: The high detective quantum efficiency (DQE) of about 65% at 70 kV (the AGFA storage phosphor system used in our study has a DQE of only 25%) and a MTF value of more than 30% at 2.5 lp/mm, which is considerably higher than for any conventional film screen or storage phosphor system [3], [13]. Thus, a high-contrast resolution and a favorable SNP, are achieved, which are more important for the detection of subtle low-contrast structures than absolute spatial resolution or matrix size.

One possible drawback of our study is that digital images were com- pared printed on laser film. In a fully digital hospital with exclusive viewing of images on image workstations, the postprocessing algorithm could probably be changed to an algorithm in favor of even better depiction of low-contrast structures detail. Furthermore, image viewing tools on a workstation (window and level setting, zoom-functions or extra edge enhancement) may influence representation of low-contrast objects. Certainly the type of workstation monitor used (1 K versus 2K monitors) will also be important in this regard. These variables will probably affect storage phosphor and direct digital radiographs in the same way, why we believe it would not alter the results of our evaluation; however, studies addressing this question are not known to the authors.

According to our data we conclude that detection of low-contrast details is substantially increased with the use of an indirect type digital flat panel detector system, which may improve radiological diagnosis of certain types of this cases and lead to reduced radiation exposure of patients.

REFERENCES