Calculation of contrast and signal-to-noise degradation factors for digital detectors in chest and breast imaging

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Introduction
The Monte Carlo model of an x-ray imaging system, used in the EU 5th framework project by the Linköping and London partner to study chest and breast imaging, was developed jointly by the London and Linköping partners (McVey et al 2003, Sandborg et al 2001). It incorporates a model of the x-ray imaging system (x-ray tube, filtration, anti-scatter device and image receptor etc.) and the patient by using a voxel phantom of an adult male (Zubal et al 1993). Validation and calibration experiments have been performed for both the chest (Ullman et al 2003b) and the breast model (Hunt and Dance 2002, Hunt et al 2002).

The model allows inclusion of anatomical or pathological details at particular positions in the anatomy (Ullman et al 2003a) and is able to calculate measures of image quality such as contrast and signal-to-noise ratio and measures of radiation risk for example entrance air kerma and effective dose. It allows alteration of imaging system settings such as tube voltage, filtration, beam size and position, choice of anti-scatter device and choice of image detector etc. The model is a useful tool for optimisations since it has been shown that in chest and lumbar spine radiography is able to predict clinical image quality as assessed by a group of radiologists (Sandborg et al 1999, 2001).

In the Monte Carlo model (MC-model) the image quality measures are calculated assuming a perfectly sharp imaging system and correction factors need to be applied to the computed data in order to make the image quality measures agree on an absolute scale. The calculation of correction factors for contrast and signal-to-noises are described in this report. A similar report focusing on analogue screen-film chest and lumbar spine radiography was completed some years ago (Sandborg et al 1999) and some of the concepts and methods are similar.

Method

Contrast degradation
The MC-model itself assumes that imaging system unsharpness is negligible. This is because the MC-model does not consider the transport of secondary charged particles released by photons in the receptor or the transport of light photons in fluorescent screens. Moreover, the MC-model assumes that all x-ray photons are generated from a single point and not from a limited-sized focal spot and that the patient movement is negligible. Any spatial filtering such as post-processing of the raw image is not considered here due to lack of data. We assume in the following that its effect on the signal is to some extent balanced by a corresponding effect on the noise.

In reality the contrast is reduced by imaging system unsharpness which is here assumed to be composed of three sources of unsharpness characterised by the corresponding modulation transfer function, MTF: detector unsharpness, MTF\textsubscript{detr}, geometric unsharpness, MTF\textsubscript{geo} and motion unsharpness, MTF\textsubscript{mot}. It is also assumed that the total system MTF is the product of the three component’s MTF.
Detector unsharpness
Data on MTF_{det} is obtained from measurement of the MTF either by imaging a slit or an edge as proposed by the IEC 62B. The pre-sampling MTF is used as an approximation of MTF_{det} and spatial frequencies up to the Nyquist frequency, v_N, are considered only.

Geometric unsharpness
Geometric unsharpness MTF_{geo} is computed by assuming a certain size and shape of the focal spot and a distance between the detail of interest and the detector plane for a given focal-detector distance. Using a Gaussian-shaped focal spot emission profile, the size can be characterised by the standard deviation of the distribution, \( \sigma \), and the MTF_{geo} is then given by (Prasad et al. 1976)

\[
MTF_{geo} = \exp(-2\pi^2 \cdot \sigma^2 \cdot v^2 \cdot (m-1)^2)
\]  

It is assumed here that the full width at half maximum (FWHM) of the Gaussian distribution characterises the width, \( f_0 \), of the distribution. The FWHM and the standard deviation, \( \sigma \), are related by FWHM=2.35 \( \sigma \). In addition a boxed-shaped emission profile can be assumed and the MTF_{geo} is then given by a sinc-function (see Sandborg et al. 1999).

Motion unsharpness
Morgan (1962) showed that the MTF due to patient motion, MTF_{mot} is given by

\[
MTF_{mot} = \left| \frac{\sin(\pi \cdot h \cdot \nu \cdot t \cdot m)}{\pi \cdot h \cdot \nu \cdot t \cdot m} \right|
\]  

where \( h \) is the speed at which the imaged structure moves in a direction parallel to the image plane and \( t \) is the exposure time. If the movement is not parallel to the image plane, the actual speed should be reduced by a factor \( \cos \theta \), where \( \theta \) is the angle between the image plane and the direction of motion. If all directions of motion are equally probable, the average resultant in the image plane is given by

\[
\int_{\theta=0}^{\theta=\pi/2} \cos(\theta) \cdot \frac{2\pi \sin(\theta)}{2\pi} \, d\theta = 0.50
\]  

since the probability of an angle \( \theta \) between \( \theta, \theta+\delta\theta \) is \( \frac{2\pi \sin(\theta)}{2\pi} \, d\theta \). Movement of the detector and x-ray tube during the exposure is assumed negligible compared to patient movement. Values of motion speed is 40 mm/s (Vano et al 1999) and exposure times is 10 ms.

Spatial-frequency content of the detail of interest
In deriving the degradation in contrast due to system unsharpness we must address the spatial-frequency content of the detail, \( \Delta S(\nu) \). It depends on its size and shape. For a detail with a Gaussian transmission profile, the size can defined by the full
width at half maximum (FWHM). The frequency content is given (Jennings et al. 1993) by

$$\Delta S(\nu) = 2\pi \sigma^2 m^2 \cdot \exp(-2\pi^2 \sigma^2 m^2 \nu^2)$$

(4)

where $\sigma$ is the standard deviation of the Gaussian distribution, $\sigma = \text{FWHM}/2.35$.

For a disk-shaped ('top-hat') detail with diameter $d$, the frequency content is given by

$$\Delta S(\nu) = m \cdot d \cdot J_1(2\pi \nu d m / \nu)$$

(5)

where $J_1$ is a Bessel function of the first order.

**Contrast degradation factor, cdf**

In order to study the effect of total $\text{MTF}_{\text{tot}}$ on the contrast of the imaged detail, $\Delta s$, the frequency content of the detail, $\Delta S$, in the image plane is multiplied by the total system $\text{MTF}_{\text{tot}}$. The $\text{MTF}_{\text{tot}}$-modified frequency content $\Delta S_{\text{mod}}$ is subsequently Fourier transformed (FT), and the modified detail, $\Delta s_{\text{mod}}$, displayed (see figure 1).

$$\Delta S_{\text{mod}}(\nu) = \Delta S(\nu) \cdot \text{MTF}_{\text{tot}}(\nu)$$

(6)

$$\Delta s_{\text{mod}} = \text{FT}^{-1}[\Delta S_{\text{mod}}(\nu)]$$

The attenuation profile of the detail, $\Delta s$, and that of image, $\Delta s_{\text{mod}}$, modified by system MTF are shown in figures 1 for a Gaussian shaped detail of 0.5 mm.

One way to quantify the effect on contrast of the unsharpness is to use line profiles through the centre of the images of the original (unmodified) detail, $\Delta s$, and of the modified detail, $\Delta s_{\text{mod}}$. The reduction in contrast can then be defined by the ratio of the peak amplitude ($P$) in the original to that in the modified ($P_{\text{mod}}$) image (see middle figure in figs. 1). The contrast degradation factor due to the imaging system MTF, $\text{cdf}_{\text{MTF}}$ is thus here defined as

$$\text{cdf}_{\text{MTF}} = \frac{P_{\text{mod}}}{P}$$

(7)

The contrast of the detail, $C$, can be computed as

$$C = C_\varepsilon \cdot \text{cdf}_{\text{MTF}}$$

(8)

where $C_\varepsilon$ is the contrast in terms of energy imparted ($\varepsilon$) to the detector from primary and scattered (s) photons beside (p2) and behind (p1) the detail. $C_\varepsilon$ is obtained from the Monte Carlo simulation. The contrast in terms of energy imparted to the image detector from primary and scattered photons from the Monte Carlo calculation is given by (Dance et al. 1992)
\[ C_\varepsilon = \frac{E(\varepsilon_{p1}) - E(\varepsilon_{p2})}{E(\varepsilon_{p1})} \cdot \frac{1}{1 + E(\varepsilon_s)/E(\varepsilon_{p1})} \] (9)

**Figure 1.** The image to the left shows the sharp 'image' (attenuation profile) of a Gaussian-shaped detail with FWHM of 0.5 mm (m=1.11) and the image to the right the MTF-modified image. The figure in the middle shows the profiles through the center of each of the images. In this case, the reduction in peak contrast is \( \text{cdf}_{MTF} = 0.695 \). A Kodak Lanex 160 screen, a 'box'-shaped focal spot (\( f_0 = 0.9 \text{ mm} \)), 20 mm/s motion speed and 3 ms exposure time were used.

**Visual response function, VTF**
Carlson 1982 has measured the contrast sensitivity of the human eye over a range of viewing angles at a luminance of 108 cd/m² as a function of cycles/degree (Barten 1992) of vertically oriented square grating patterns. The contrast sensitivity as function of cycles/degree can be converted into contrast sensitivity as function of line-pairs per mm (lp/mm) in the image plane at a viewing distance \( L \) (mm) by the expression

\[ u[\text{cyc/degree}] = 180 \cdot u[\text{lp/mm}] \cdot L \cdot \pi \] (10)
When normalised to unity at its peak value the contrast sensitivity is here called visual transfer function, VTF, and its dependence on spatial frequency and viewing distance is shown in figure 2.

![Figure 2](image.png)

**Figure 2.** The visual response of the eye, VTF, as function of spatial frequency and viewing distance. At shorter viewing distance the maximum VTF shifts to higher spatial frequencies. Legends: red: 100 cm, green: 50 cm and blue: 25 cm.

**Signal-to-noise**

In calculating the signal-to-noise ratio, SNR, the Monte Carlo model takes account of the thickness and location of the detail in the patient, the detector’s response, in terms of energy imparted to the image receptor per unit area by incoming primary and secondary photons, patient size, tube voltage, filtration, anti-scatter device etc. The modelling of the most important aspects of the detector response function is done explicitly in the MC-model by specifying the atomic composition and area-thickness (mg/cm²) of the active layer in the detector. By the active layer we mean the layer that absorbs the x-ray photons and generate secondary particles. This active detector layer is typically made from a scintillator material such as CsI or Gd₂O₂S or from an amorphous semiconductor material such as a-Si or a-Se. This accounts for the quantum absorption efficiency (fractional x-ray absorption) and the Swank factor, $I_x$. The latter ($I_x$) due to fluctuations in the energy imparted to the receptor per incident photon due to the poly-energetic nature of clinical x-ray spectra and due to escape of scattered photons and/or of characteristic x-rays from the active detector layer. The absorption of x-rays in the detector protective cover is accounted for in the MC-model by specifying the atomic composition, thickness and density of this layer.

Since the DQE of the detector typically decreases with increasing spatial frequency, the SNR of a small detail with a wider frequency content would be additionally reduced due to reduced DQE at high spatial frequencies. In addition, measurement of DQE indicate that the DQE is low at very low spatial frequency indicating also a
degradation of SNR of large details with a narrow frequency-band at low spatial frequencies. In assessing the total effect we must consider the visual transfer function of the eye-brain system, VTF, which shows a maximum response at $v_{VTF-max}$ depending on light conditions and viewing distance.

The reduction of the DQE at higher spatial frequencies is due to the unbalance of the $MTF^2(v)$ and the $NPS(v)$ which in turn stems from the transfer properties of the optical photons within the active layer and at its interfaces to the optical detector (film or photo-diode matrix). At present we do not model the secondary particles (light photons or electrons) and the roll-off of DQE at high spatial frequencies has to be accounted for by using measured data. Methods exist to account for these aspects (Jennings and Badano 2003, Nishikawa and Yaffe 1990 and Nishikawa 2003) but requires detailed knowledge of the design of the coupling of the x-ray detector and of the secondary light detector. This detailed information is typically propriety information and hence not available for commercial detectors.

The correction factors to be applied to the Monte Carlo calculated values of the SNR are divided in to two groups. The first group comprise the factors that do not depend on the spatial frequency or the air kerma at the detector. The second group of factors depend on these conditions.

**Dose- and frequency-independent factors**

**Detail size**
The projected area in the image plane of the detail is fixed in the MC-model and thus a detail with a given thickness but with different area will get the same SNR from the MC-model. For a system with only white quantum noise, the SNR scales with the projected area in the image plane and a simple correction factor can be applied. The correction factor is

$$\text{SNR}^2 = \text{SNR}_{MC}^2 \cdot \frac{A_{\text{detail}}}{A_{MC}}$$

(11)

where $A_{MC}$ is the area used in the MC-model (0.25 mm$^2$) and $A_{\text{detail}}$ is the area of the detail in the patient in the image plane.

**Fill factor**
The MC-model does not account for the fill factor, $ff$, i.e. the ratio of the effective area of the detector element, $A_{\text{del}}$, and the pixel area approximated by the pixel-pitch squared. The pixel-pitch is the distance between the geometric centres of two neighbouring detector elements. The correction factor is given by

$$ff = \frac{A_{\text{del}}}{\text{pixelpitch}^2}$$

(12)

and accounts for the loss of photon fluence in inactive parts of the detector elements.

**Swank factor $I_L$**
Light-emitting or indirect conversion detectors

For a given energy imparted to the active layer of the receptor, there is a fluctuation in the amount of light collected at the output for example the photodiode matrix. The degradation of the SNR due to these fluctuations are given by the Swank factor $I_L$. They are not considered in the MC-model and $I_L$ has to be applied as a correction factor to the $\text{SNR}_{\text{MC}}$. Here, $I_L=0.80$ (Mickish and Beutel 1993, Ginzburg and Dick 1993) is used for unstructured fluorescent materials such as Gd$_2$O$_2$S. A significantly higher factor $I_L=0.95$ (less SNR-reduction) (Rowlands and Taylor 1983) was used for structured (needle-shaped) fluorescent material such as CsI (Albagli 2003).

Charge-emitting or direct-conversion detectors

Since the fluctuations in the number of collected charges is lower than for the number of collected light photons and their spatial distribution is less broad than is the case for the corresponding light photons in an indirect conversion detector material, the factor $I_L$ for a direct-conversion detector is expected to be close to unity (Henry et al. 1995). An $I_L=1.0$ was therefore used.

Dose- and frequency-dependent factors

An imaging system is quantum-noise limited if the SNR is dominated by the statistics of the energy imparted to the receptor by x-ray photons. The dose- and frequency range for quantum-noise limited operation typically coincides with the range where DQE is the highest. The DQE of real digital detectors is, however, not independent of dose (or air kerma at the detector). System gain noise (CR) and additive noise (DR) reduce the DQE at high and low doses, respectively. At low doses, the pixel design of DR systems is important (Albagli et al 2003) and it is the amount of electronic noise in relation to the conversion gain (light photons or electrons) that should be optimised.

For CR systems (image plates), three sources of noise in addition to quantum noise are identified (Hillen et al 1987). These are luminescence noise, granularity noise and readout noise, each which will be described briefly below following the description in Hillen et al 1987.

Luminescence noise arises from fluctuations in stimulated luminescence quanta detected by the photo-multiplier tube. Its frequency spectrum is white (i.e. constant with spatial frequency) and the magnitude is proportional to the square root of the x-ray photon fluence. Granularity noise is due to the grainy (non-uniform) structure of the phosphor material which corresponds to local efficiency variations in conversion from x-ray to light. The causes are fluctuations in x-ray absorption efficiency and transport of laser (stimulation) light and luminescence light in the phosphor and binder emulsion. The spectrum is coloured, with lower noise magnitude at high frequencies, and the magnitude is proportional to the photon fluence i.e. the signal. Readout noise is assigned to instability in the laser light or in the fluctuations in collection and amplification of the luminescence light. As with granularity noise the spectrum is coloured and the magnitude proportional to the photon fluence.
As granularity and readout noise are proportional to the signal whereas quantum and luminescence noise are proportional to the square root of the signal, the first two noise sources cause the DQE to decrease at high doses and the latter two noise sources to reduce the DQE at low doses. The white luminescence noise also reduces the DQE at high spatial frequency.

**Experimental determination of DQE(v,K)**

The MC-model corresponds to an ideal, quantum-noise limited system, since only fluctuations in energy imparted to the receptor are considered. The DQEMC is therefore independent of the fluence (dose) level of incident photons. Knowledge about the dependence of the DQE on air kerma at the detector was obtained from measurement of DQE(v,K) at a range of air kerma values. For the CR system, data on DQE was available for 32 different air kerma levels between 0.06 μGy and 270 μGy (Sund and Båth 2001). It was therefore possible to interpolate between these air kerma-levels. For the DR system on the other hand, the DQE was only available at three separate air kerma levels, namely 1.2, 3.9 and 12.1 μGy (Illers and Buhr 2003). The absolute noise power spectrum, NPS, the detector’s modulation transfer function, MTF and the photon fluence Φ were however known for each air kerma level. It was therefore possible to fit a second-degree polynomial to the measured NPS and extrapolate the NPS to air kerma in the range 1-300 μGy. The DQE was then computed by

\[
DQE(v,K) = \frac{MTF(v)}{Φ(K) \cdot NPS(v,K)}
\]

(13)

and is visualised graphically below.
Figure 3. Extrapolation of DQE from measured DQE at three air kerma levels using a second degree polynomial. The DQE measured by Illers and Buhr are indicated by the symbols and the lines are intermediate DQE-values using the equation above.

These extrapolations were done for each spatial frequency and the data is therefore noisy as can be noted in figure 4c. In figure 4b the DQE data was smoothed for spatial frequencies above 0.5 lp/mm.
Figure 4. 2D plots of DQE as function of spatial frequency and air kerma at the receptor for the CR (a) and DR (b: smoothed and c: unsmoothed) systems.

The spatial frequency and dose, \( K \), dependence of the DQE is obtained from the measured \( DQE(v,K) \). The DQE is then normalised to its maximum value by

\[
DQE_N(v,K) = \frac{DQE(v,K)}{\text{max}(DQE(v,K))}
\]  

(14)

The maximum DQE-value, \( \text{max}(DQE(v,K)) \), is assumed to be the value computed by the MC-model. By normalising to this value, we take the relative inefficiencies of the detector at any dose and spatial-frequency into consideration empirically.

Additional degradation factors

The MTF of the image detector is included in the spatial-frequency dependence of the DQE whereas the motion and geometric unsharpness is not. We therefore add an additional source of SNR-degradation by including them in the final SNR-degradation factor, \( \text{SNR}_{\text{DF}}^2 \). \( \text{SNR}_{\text{DF}}^2 \) is computed as an average over the spatial-frequency content of the detail and visual system VTF by
\[
\int_{0}^{v_{N}} \Delta S^2(v) DQE_{N}(v, K) MTF^{2}_{\text{geo}}(v) MTF^{2}_{\text{mot}}(v) VTF^{2}(v) \, dv
\]

\[
\text{SNR}^{2}_{\text{DF}}(K) = I_{L} \, \text{ff} \left[ \int_{0}^{v_{N}} \Delta S^2(v) VTF^{2}(v) \, dv \right]^{-1}
\]

(15)

The \( \text{SNR}^{2}_{\text{DF}} \) is computed for each detector type (CR- or DR-system) and for a given size of the pathological detail in the range 0.2-10 mm as a function of dose (air kerma) at the detector. Typical imaging system configurations are listed in table 1 below.

**Table 1.** Some important imaging system characteristics related to CDF and SNR^{2}_{DF}.

<table>
<thead>
<tr>
<th><strong>a. Chest system parameters</strong></th>
<th><strong>CR system</strong></th>
<th><strong>DR system</strong></th>
</tr>
</thead>
<tbody>
<tr>
<td>Location and x-ray room</td>
<td>Motala, room 3</td>
<td>Linköping, room 15</td>
</tr>
<tr>
<td>Trade name</td>
<td>Philips PCR5000</td>
<td>Siemens Vertix FD</td>
</tr>
<tr>
<td>Focal spot size (mm)</td>
<td>1.0</td>
<td>1.0</td>
</tr>
<tr>
<td>Focal spot shape</td>
<td>Gaussian</td>
<td>Gaussian</td>
</tr>
<tr>
<td>Focus-detector distance (cm)</td>
<td>150</td>
<td>180</td>
</tr>
<tr>
<td>Focus-detail distance (cm)</td>
<td>135</td>
<td>165 cm</td>
</tr>
<tr>
<td>Magnification</td>
<td>1.11</td>
<td>1.09</td>
</tr>
<tr>
<td>Detail size (mm)</td>
<td>0.5, 1.0, 4.0, 10.0</td>
<td>0.5, 1.0, 4.0, 10.0</td>
</tr>
<tr>
<td>Detail shape</td>
<td>Gaussian</td>
<td>Gaussian</td>
</tr>
<tr>
<td>Motion speed (mm/s)</td>
<td>40</td>
<td>40</td>
</tr>
<tr>
<td>Exposure time (ms)</td>
<td>10</td>
<td>10</td>
</tr>
<tr>
<td>Pixel pitch</td>
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<td>0.143</td>
</tr>
<tr>
<td>Detector element size (mm)</td>
<td>0.20</td>
<td>0.125</td>
</tr>
<tr>
<td>Fill factor, ff</td>
<td>1.00</td>
<td>0.76</td>
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<tr>
<td>Swank light factor, ( I_{L} )</td>
<td>0.80</td>
<td>0.95</td>
</tr>
<tr>
<td>Nyquist frequency (lp/mm)</td>
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<tr>
<td>Detector material</td>
<td>BaFCl</td>
<td>CsI</td>
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<tr>
<td>Detector thickness (mg/cm^2)</td>
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<td>190</td>
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<tr>
<td>Viewing distance (cm)</td>
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<td>25</td>
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</table>

<table>
<thead>
<tr>
<th><strong>b. Breast system parameters</strong></th>
<th><strong>CR system</strong></th>
<th><strong>DR system</strong></th>
</tr>
</thead>
<tbody>
<tr>
<td>Location and x-ray room</td>
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<td>London</td>
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<tr>
<td>Trade name</td>
<td>Fuji FCR5000M HR-BD</td>
<td>GE Senographe 2000D</td>
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<tr>
<td>Focal spot size (mm)</td>
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<td>Focal spot shape</td>
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<td>Gaussian</td>
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<tr>
<td>Focus-detector distance (cm)</td>
<td>65</td>
<td>65</td>
</tr>
<tr>
<td>Focus-detail distance (cm)</td>
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<td>1.066</td>
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<td>Detail size (mm)</td>
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<tr>
<td>Detail shape</td>
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<td>Gaussian</td>
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<tr>
<td>Motion speed (mm/s)</td>
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<td>4</td>
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<tr>
<td>Exposure time (ms)</td>
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<td>100</td>
</tr>
<tr>
<td>Pixel pitch</td>
<td>0.05</td>
<td>0.10</td>
</tr>
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</table>
The size and composition of the pathological details included in the model are listed in table 2 for chest and breast examinations. A detailed description can be found in Ullman et al 2003a.

Table 2. The pathological details included in model are listed below.

a. Chest

<table>
<thead>
<tr>
<th>Detail</th>
<th>Diameter (mm)</th>
<th>Density (g/cm³)</th>
<th>Material</th>
</tr>
</thead>
<tbody>
<tr>
<td>Metastasis</td>
<td>4 and 10</td>
<td>1.03</td>
<td>soft tissue</td>
</tr>
<tr>
<td>Calcifications</td>
<td>0.5 spheres</td>
<td>3.3</td>
<td>hydroxiapatite</td>
</tr>
<tr>
<td>Kerley B-lines</td>
<td>1mm x 10 mm</td>
<td>1.0</td>
<td>water</td>
</tr>
<tr>
<td>Fibrosis</td>
<td>1mm x 10 mm</td>
<td>1.03</td>
<td>soft tissue</td>
</tr>
</tbody>
</table>

b. Breast

<table>
<thead>
<tr>
<th>Detail</th>
<th>Diameter</th>
<th>Density (g/cm³)</th>
<th>Material</th>
</tr>
</thead>
<tbody>
<tr>
<td>Masses</td>
<td>3, 5 and 8 mm</td>
<td>1.09</td>
<td>glandular tissue</td>
</tr>
<tr>
<td>Calcifications</td>
<td>0.2 spheres</td>
<td>3.3</td>
<td>hydroxiapatite</td>
</tr>
</tbody>
</table>

Results

Contrast degradation

Figure 5 shows the contrast degradation factor $c_{d_{MTF}}$ as function of pathological detail diameter or size. The degradation factor is close to one for large details but reduces quickly as the size of the detail is reduced below 0.5 mm. The lower CDF for the DR-system is due to the low-frequency drop seen in figure 6.
**Figure 5.** Comparison of contrast degradation factor, CDF, for CR-system (upper curve) and DR-system (lower curve). The differences in CDF is primarily due to the low-frequency drop for the DR-system (see figure 6) and the fact that the total system MTF_{tot} is limited by the MTF_{det} (see figure 7).

**Figure 6.** The MTF of the CR-system (upper curve) and DR-system (lower curve). Notice the low-frequency drop for the DR-system which results in slightly lower contrast-degradation factors (see figure 5).
Figure 7. Modulation transfer function, MTF and its components, $MTF_{tot}$: ----, $MTF_{scr}$: - - -, $MTF_{geo}$: ......., $MTF_{mot}$: -.-.- for the DR-system. Imaging conditions: magnification 1.09, 1.0 mm Gaussian-shaped focal spot, 10 ms exposure time and 40 mm/s motion speed.

**Signal-to-noise degradation**

Figure 8 show the $SNR^2_{DF}$ as function of spatial frequency for a CR system operating at 2 $\mu$Gy. Included in the graph is the visual transfer function VTF of the eye-brain system, indicating the low efficiency by which low and high spatial-frequencies are received by the human eye.

Figure 8. The visual transfer function, VTF, for a 25 cm viewing distance and the $SNR^2_{DF}$ as function of spatial frequency for a DR-system (Trixel) operating at 2 $\mu$Gy. Legends: VTF: - - -, $SNR^2_{DF}$: ----.
In figure 9, we compare the SNR²_{DF} for a screen-film system (SF-system) and CR-system. The SNR-degradation for a SF-system at high and low doses, due to the film H&D-curve, is apparent but the SNR-degradation is also substantial for the CR system at high doses due to read-out noise. In intermediate dose-levels the SF-system show less SNR-degradation than the CR-system.

![Figure 9](image-url)  
Figure 9. The SNR²_{DF} as function of air kerma at the detector for a screen-film, SF-system, (Lanex 320/TML: - - - ) and a CR-system (——). Notice the low SNR²_{DF} values for the SF-system at low and high doses at the detector.

Figure 10 shows the SNR²_{DF} as function of air kerma at the detector for different detail sizes and figure 11 the SNR²_{DF} as function of detail size for different air kerma levels.

The SNR²-degradation factor decreases (i.e. larger degradation of SNR) with increasing air kerma but depends also on the size of the detail. The degradation factor has a local maximum for detail sizes around 1-2 mm which means that the degradation of the SNR by the imaging system is smallest for this size of details. For details larger than approximately 10mm the degradation factor is smaller because the low DQE for spatial frequencies below 0.1 lp/mm.
Figure 10. SNR$^2_{DF}$ for a CR system (-----) and a DR (_______) system as function of air kerma at the detector for details of 1, 4 and 10 mm in diameter. There is a significantly larger degradation of SNR with increasing dose at the detector.

Figure 11. SNR$^2_{DF}$ for a CR system (-----) and a DR (_______) system as function of detail size for three different doses at the detector 1 µGy (thin line), 10 µGy and 100 µGy (thick line). There is a maximum of the SNR$^2_{DF}$ for details of approximately 1-2 mm in diameter corresponding to the maximum sensitivity of the eyes (VTF) and the image detector (SNR$^2_{DF}$), see figure 8.
Figure 12. \( \text{SNR}^2_{\text{DF}} \) for a mammography DR system (GE 2000D) for a 200 \( \mu \text{m} \) calcification and a 8 mm mass.

Figure 13. \( \text{SNR}^2_{\text{DF}} \) for a mammography screen-film, SF system (KodakMinR2190/Kodak2000) for a 200 \( \mu \text{m} \) calcification and a 8 mm mass (tumor). The maximum \( \text{SNR}^2_{\text{DF}} \) is approximately the same as with the digital DR-system (see fig.12) but the SF-system show inferior performance (lower \( \text{SNR}^2_{\text{DF}} \)) at low and high doses at the detector corresponding to low and high optical densities.

Conclusions
A strategy for including imaging system sharpness and additional noise-sources to the Monte Carlo model has been developed and example of their effect on the
contrast and signal-to-noise ratio of pathological details are given for both cheat and breast imaging.

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