Radiation dose reductions for X-ray images contaminated with Poisson noise

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Authors: T. Aspelmeier\textsuperscript{1}, G. Ebel\textsuperscript{1}, U. Engeland\textsuperscript{2}; \textsuperscript{1}Göttingen/DE, \textsuperscript{2}Götingen/DE
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Purpose

The X-ray dose for patient images is usually chosen such that sufficiently detailed images are obtained since low doses result in high quantum (Poisson) noise.

The purpose of this study is to introduce and evaluate a method of improving low-dose images numerically which allows significantly lower doses while maintaining image quality.
Methods and Materials

Mathematical and physical background

The aim of this study was to develop and test a method to reduce Poissonian image noise stemming from low radiation doses. The first step is to make an accurate physical model for each of the components of the X-ray system consisting of

- X-ray tube
- object imaged (patient)
- scintillator crystal
- light sensitive TFT matrix
- readout electronics.

Our model does not consider the image degrading effects caused by the imaged object itself (scatter). Most other factors are, however, at least approximately taken into account. In particular,

- the X-ray tube has a focus of finite dimensions, causing image blur
- the scintillator crystal causes additional blur
- the finite number of X-ray photons emitted by the tube and registered by the crystal induces Poisson noise
- the readout electronics add additional (approximately Gaussian) noise.

These ingredients were composed into a negative log-likelihood functional

\[ L[f] = -\log P(y|f) \]

(the negative logarithm of the probability of measuring the image \( y \) given the "true" image \( f \)). This term covers the blur generated by the focal spot and the crystal, the Poisson noise of the photon statistics and the Gaussian noise generated by the electronics.

For a given measured image, \( y \) is available but \( f \) is not. Finding that \( f \) which minimizes \( L[f] \) gives the image which has the highest probability of having produced the measurement \( y \). However, simply maximizing \( L[f] \) with respect to \( f \) does not work since this is an ill-posed inverse problem. Additional regularization is required.

Regularization uses additional information about the possible image contents. Here, we use the fact that all real-world images are compressible in a suitable wavelet basis.
Choice of the wavelet basis, however, is not trivial. Simple (separable) standard wavelets produce visible artifacts along the vertical and horizontal directions, even though their "objectively" measured performance may be good. We therefore used nonseparable 2 dimensional wavelets [1].

With these wavelets, the likelihood functional is augmented by a sparsity enforcing term $\| T f \|_p$ ($T$: wavelet transform matrix, $\| ... \|_p$: $p$-norm with $0 < p < 1$) to

$$F[f] = L[f] + a \| T f \|_p.$$ 

Compressive sensing theory provides a suitable value for the regularization strength $a$ in the case of $p=0$ and an error bound for the minimizer $f$ of $F[f]$ (for sparse sampling, but also for full sampling as in the present case) [2].

The resulting minimizer $f$ is, under the assumptions of the model and wavelet sparsity, the best estimate for the "true" noisefree image which caused the noisy and blurred measurement $y$.

**Implementation**

The method of improving the image quality consists of finding the minimizer of $F[f]$. This can only be solved by iterative algorithms. We used a variant of the separable paraboloidal surrogates algorithm [3].

Since the dimensionality of the image space is very high (equal to the number of pixels, i.e. ca. 3000×3000), iterations are very time consuming. We therefore implemented the algorithms on a GPU which uses massive parallelization to speed up the calculations. Processing a 3000×3000 image takes roughly 30s.

The sparsity enforcing term involving the $p$-norm was chosen with $p=1$ such that $F[f]$ is a convex functional, hence convergence to a global minimum is guaranteed. The prefactor $a$ was chosen empirically.

**Phantom measurements**

Normal and low dose (25%, 32%, 50% and 100% of the normal dose) X-ray images of an anthropomorphic skull phantom were acquired on a Swissray ddR formula system which is equipped with a Trixell Pixium 4600 flat panel detector. The acquisition parameters
used for the different dose settings were 80kV in all cases and 3.125mAs (25%), 4mAs (32%), 6.25mAs (50%) and 12.5mAs (100%). We did not use a scatter grid. Neither phantom nor detector nor X-ray source were moved during the acquisition in order to allow for direct pixelwise comparison of the resulting images.

Usually, the raw output from the X-ray detector is immediately contrast processed in order to aid the visual inspection by human observers. Since our method requires unprocessed input data, we obtained the raw detector data files, processed them with our noise reduction software and then applied our own contrast enhancement software. This allowed for identical and reproducible parameter settings for all images which is important for an unbiased evaluation of the results. We also used the contrast enhancement software on the unprocessed raw data files in order to compare between processed and unprocessed versions. All X-ray images shown here have been contrast enhanced except the line phantoms (Fig. 3 on page 8 and Fig. 6 on page ) which do not require it.

**Fig. 1 on page 7 and Fig. 2 on page 7** show 4 original (unprocessed but contrast enhanced) images with the 4 different doses. The decrease of noise with increasing dose can be clearly seen, in particular in the zoomed images. The processed images are shown in the results section.

**Evaluation criteria**

In order to assess the improvement in image quality objectively, we also obtained a series of 10 images with identical settings. These 10 images were averaged, resulting in an estimate of the ideal noisefree image. The quality of the processed images can be evaluated by calculating the squared deviations of the processed and unprocessed images from the the estimate of the ideal image for each pixel. These squared deviations can be compared and the amount of noise reduction quantified.

In addition to evaluating the quality of the anthropomorphic phantom images, we also used a line phantom to show the improvement of the contrast to noise ratio (CNR). **Fig. 3 on page 8** shows the phantom image. The contrast to noise ratio was determined by dividing the contrast of a line pair (difference of maximum and minimum intensities) by the noise (standard deviation of the pixel values in a region of uniform intensity on the rim of the phantom).

Such objective figures of merit are however not enough. While they are essential to prove the effectiveness of the method, it is even more important that no visually disturbing or diagnostically misleading artifacts are present in the processed images. This can not be judged by objective figures alone which by their nature can not identify such artifacts and gauge their effect on the human visual system. We therefore deemed it very important...
to evaluate the results by an additional visual inspection rather than to count on figures of merit alone.
Images for this section:

**Fig. 1**: Skull images with 4 different doses. Top left: 25% dose (3.125mAs), top right: 33% dose (4mAs), bottom left: 50% dose (6.25mAs), bottom right: 100% dose (12.5mAs).

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Fig. 2: Detailed zoom of the 4 images in Fig. 1. Top left: 25% dose (3.125mAs), top right: 33% dose (4mAs), bottom left: 50% dose (6.25mAs), bottom right: 100% dose (12.5mAs).

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Fig. 3: Image of a line pair phantom at 40kV and 0.5mAs.

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Results

Squared deviations

The squared deviations of an unprocessed and a processed image from the approximate ideal image obtained by averaging 10 identical exposures indicate the amount of noise reduction achieved by our software. Fig. 4 on page 12 shows a plot of the squared deviations along a cut across a phantom image as shown in Fig. 5 on page 12. Fig. 4 on page 12 shows that the processing has reduced the amount of noise by a factor of 2 with respect to the unprocessed images. According to Poisson statistics, this corresponds to an effective increase of the dose by a factor of 2.

Contrast to noise ratio

Fig. 6 on page 13 shows an image of a line phantom acquired at 40kV and 0.5mAs. The comparison between unprocessed (right) and processed (left) image shows that the noise has been significantly reduced while the resolution is essentially unchanged. This is exemplified further in Fig. 7 on page 14 which displays the contrast to noise ratio. The CNR has been improved significantly by the processing to match or exceed that of a measurement at 1mAs, i.e. twice the dose, except for very high spatial frequencies already beyond the Nyquist limit.

Visual assessment

Visual comparisons of the processed images at 50%, 32% and 25% with the unprocessed full dose image are shown in Fig. 8 on page 15, Fig. 9 on page 16 and Fig. 10 on page 17, respectively. The figures show the 100% dose image in the middle, the unprocessed low dose image on the right and the processed low dose image on the left. The noise level of the processed images is on par with the 100% dose image (which is also demonstrated in Fig. 4 on page 12).

Note that the processed images have reduced noise but are not blurred, i.e. even very fine structures remain visible. This is corroborated by the CNR results in Fig. 7 on page 14. For the very low dose images (25%), however, some image details are beginning to be lost.
Importantly, the images show no visual artifacts such as streaks or jpeg-like structures which would lower the diagnostic quality of the images.
Fig. 4: Squared deviations of a processed and an unprocessed image along a cut through Fig. 5.

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Fig. 5: One of 10 identical phantom images used to generate an estimate of the ideal noise free image. This exposure was taken at 3.2mAs. The arrow indicates the cut along which the squared deviations have been plotted in Fig. 4.

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Fig. 6: Image of a line phantom at 40kV and 0.5mAs (right). On the left, the same image after processing is shown. Note the reduction of noise without degradation of line resolution.

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Fig. 7: Contrast to noise ratio for the unprocessed line pair image at 0.5mAs (red), the processed image at 0.5mAs (green) and an unprocessed line pair image at 1mAs (blue). The processing improves the CNR of the lower dose image even beyond that of the image with the higher dose.

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Fig. 8: Comparison of a 100% dose image (middle), a 50% dose image (right) and the same 50% dose image after processing (left).

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Fig. 9: Comparison of a 100% dose image (middle), a 32% dose image (right) and the same 32% dose image after processing (left).

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Fig. 10: Comparison of a 100% dose image (middle), a 25% dose image (right) and the same 25% dose image after processing (left).

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Conclusion

1. A method of reducing image noise without loss of resolution in low dose X-ray images yields high quality images comparable to normal doses. The radiation dose can be reduced by up to 75% although some loss of image detail starts to occur at these low dose levels.
2. Noise reduction is particularly effective since the influence of all elements in the acquisition chain and the image sparsity are accurately accounted for.
3. Processing is very time consuming but can be implemented on a GPU for fast computation.
References


Personal Information

T. Aspelmeier, G. Ebel and U. Engeland, Globe Medical Technology International GmbH, Hansmatt 30, CH-6370 Stans, Switzerland