GEOMETRIC AVERAGING OF X-RAY SIGNALS IN AUTOMATIC EXPOSURE CONTROL

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ABSTRACT

Improper dose control in X-ray cardio-vascular systems leads to a reduced Signal-to-Noise Ratio (SNR) in regions of interest of the X-ray image. We aim at reducing the influence of direct radiation, entering a measuring field for X-ray dose control in a Flat Detector which gives too bright areas (highlights) in the image. It is our desire to use a norm-like signal size that represents a minimal dose value while maximizing information transfer and thus image quality. In a dose control system, it is common practice to employ a special averaging technique for computing a representative signal level controlling the X-ray. We have found that the geometric averaging outperforms the existing techniques and significantly improves the image quality. Our approach reduces the highlight influence and guarantees an adequate Contrast-to-Noise ratio for de-centered objects. We provide convincing experimental results showing a strongly improved image quality with respect to contrast and detail.

Index Terms—X-ray, cone beam CT, dose control, signal size, image quality.

1. INTRODUCTION

In X-ray imaging, a proper dose control is indispensable to maximize information transfer, while keeping the absorbed dose of the patient as low as reasonably achievable. When dose is not properly controlled, the Signal-to-Noise Ratio is severely decreased, which leads to a reduced contrast resolution, thereby negatively influencing the diagnostic value of the image. In this paper, an improved method is described pertaining to the optimization of dose control and Image Quality (IQ) in moving C-arm Flat Detector X-ray 3D imaging. With a moving C-arm, direct radiation sometimes passes un-attenuated by the object, which initially leads to very bright areas in the image. These conditions are found in case of de-centered objects or in case of highly transparent parts of the body. As a consequence, the Contrast-to-Noise Ratio will be lowered in these projections. An example of the resulting effect would lead to a poor separation of bone and vessels in 3D vascular imaging, or a reduced contrast-resolution for soft tissue imaging. The problem is obviously not limited to 3D cases, as in the presence of direct radiation, vessels crossing bone in subtraction angiography, may also be poorly delineated due to noise.

Typical dose control concepts use a signal size for control purposes which integrates the signal content within a measuring field. Unfortunately, this results in an average grey level that is susceptible for outlier contamination in the Probability Distribution Function (PDF). Our aim is to find a norm-like signal size that maximizes the information transfer of the image, i.e. optimally maps its latitude onto the detector dynamic range. The computed signal level for dose control is typically derived from the image with a real-time weighting technique. Existing techniques lead to dark, low-level images. Although the arithmetic mean is widely used for weighting, we have adopted the geometric average for the signal norm. We investigated this control method which optimizes patient dose and detector dose and found that it yields a well-defined IQ in terms of system contrast resolution.

We structure this paper as follows. Section 2 gives a brief description of an X-ray dose control system. Section 3 describes requirements and performance in terms of Signal-to-Noise Ratio and dose. Section 4 gives the improvement of the proposed method and experiments. Section 5 discusses the results and presents conclusions.

2. DOSE CONTROL SYSTEM

In X-ray systems equipped with a flat detector, as schematically shown in Fig. 1, an X-ray beam at the left irradiates the object (ellipse) and a flat detector (solid bar) receives the radiation and converts it to an electrical signal. At the right side of Fig. 1, we show a qualitative impression of a grey-scale profile contained in a measuring field for a de-centered X-ray elliptical phantom. The profile at the bottom clearly shows a highlight area from direct radiation, giving potentially too bright areas in the image. In the above concept, a dose control system is embedded which operates as follows: a control signal is derived from the detected signal and controls the X-ray source such that the detected signal is kept constant.
In typical systems, when highlights occur in the measuring field, the control system will reduce the input radiation to a lower dose, resulting in a reduced image signal in the field of interest. This lowers the Signal-to-Noise Ratio (SNR) in the image. De-centering of the object further amplifies this effect. It is our objective to operate the system at the highest possible stability and reduce the influence of the direct radiation parts (highlights in the X-ray image) on the control loop. Let us now model the control system into a signal processing system. A diagram of this model is visualized in Figure 2.

\[ T_i = a_i / I_0 \]  

These per pixel transmission values serve as input for an image reconstruction algorithm.

3. REQUIREMENTS AND PERFORMANCE

As the flat detector has an inherent low contrast loss, several orders of magnitude may exist between attenuated and un-attenuated detected radiation. The control system will respond to direct radiation detected within a measuring field in an image by setting the input radiation to a lower value. In regions of interest, the output values will be too low and consequently the Signal-to-Noise Ratio (SNR) will be lowered in the projected image, since the output signal is kept constant. This determines directly the SNR in the reconstructed image. The voxel noise upon reconstruction is described by:\[^{11}\]:

\[ \sigma_{voxel}^2 = \frac{4\pi^2}{m_{proj} \eta M_{geo}^2 NEB_{proj} NEB_z} \sum_{j=0}^{n_{proj}} \frac{1}{q_j \mu_j^2} \]  

Here, the standard deviation is expressed as a function of the number of views \( m_{proj} \), the geometric magnification \( M_{geo} \), the detector Detective Quantum Efficiency \( \eta \), the noise equivalent bandwidth in the reconstruction plane \( NEB_{proj} \), and the noise equivalent bandwidth in an axial plane \( NEB_z \), the input quantum density \( q_0 \), the linear object absorption coefficient \( \mu \) and the absorption length \( d \). The input quantum density \( q_0 \) directly determines the voxel standard deviation. The image quality deteriorates as \( q_0 \) is lowered. In case of direct radiation, the control system regulates towards a lower value of \( q_0 \) as the input is too high. This regulation is performed by measuring the signal level, using an averaging technique. A typical function is arithmetic averaging having the disadvantage that the value is dominated by highlights. Consequently, the control system will lower the input dose (lower \( q_0 \)), so that the standard deviation of the voxel signal is increased. Using arithmetic averaging in the signal processor, the pixel grey values \( a_i \) are evaluated with:

\[ aV_{\text{arithmetic}} = \frac{1}{n} \sum_{i=1}^{n} a_i \]  

Equal weights are applied to every pixel value \( a_i \) such that the influence of direct radiation is directly visible. The Contrast-to-Noise Ratio will be deteriorated and consequently the contrast-detail visibility reduced.

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4. IMPROVEMENT AND EXPERIMENTS

Numerous compensation techniques have been studied to deal with highlights or backlights in automatic exposure control systems. All use more or less knowledge about the relevant parts of the histogram to exclude the highlight portion of the histogram. The X-ray control system is not different in that respect. Some general examples of histogram-based compensation are given in: [2][3]. Instead of the above arithmetic averaging and highlight-excluding methods, we propose an alternative control signal norm such as geometric averaging:

\[
\text{av}_{\text{geometric}} = \sqrt[n]{\prod_{i=1}^{n} a_i} \quad (4)
\]

for \(a_i > 0\). The advantage of this signal norm is that it does not favor the direct radiation terms, but instead it balances the weighted distance between the average of the subset of interest region and the average of the highlight portion of the PDF. Equation (4) may be rewritten for calculation purposes to:

\[
\text{av}_{\text{geometric}} = \exp \left( \sum_{i=1}^{n} \ln a_i^{1/n} \right), \quad \text{or}
\]

\[
\text{av}_{\text{geometric}} = \exp \left( \frac{1}{n} \sum_{i=1}^{n} \ln a_i \right) \quad (5)
\]

The method achieves an expansion of low pixel values and compression of highlights (see Figure 3). Since zero values are forbidden, the lower part of the logarithmic function is adapted slightly in order to avoid numerical overflow. This signal norm complies with the necessary positivity and uniformity (scalability) properties.

Let us now describe how we will compare the new dose control with the typical one. First, we will determine the object transmission factor for both methods. Second, the ratio between the two transmission factors will give us the improvement factor. Third, the improvement factor is inserted in the expression of the voxel noise (2), so that an improvement of the standard deviation can be calculated.

The method has been applied in a commercial vascular imaging system\(^1\), equipped with 3D reconstruction software. For comparison, the arithmetic averaging method could be switched on as well, allowing for the following experiments. A proprietary elliptical phantom of 32 cm by 24 cm with a density of 12HU (Hounsfield Units) was placed off the system rotation centre by 5 cm in a lateral (vertical) direction and a regular run of 620 images was acquired. The phantom contains inserts with circular cross sections of varying diameter and density (Fig. 4).

\(^1\) The Xper vascular system of Philips Healthcare.

Visibility can be found by viewing the set of diameter-contrast combinations that can just be discerned \(^1\). For both signal norms, the AGL was measured in a 60% circular measuring field (referred to the detector larger side), as well as the input \(I_0\).

Figure 4: An elliptical phantom.

![Normalized AAGL vs GAGL](image1)

**Figure 5:** Stabilized average signal AAGL vs GAGL.

![Transfer](image2)

**Figure 6:** The object signal transfer function.

Fig. 5 shows the measured results for the Arithmetic Average Grey Level (AAGL) and Geometric Average Grey Level (GAGL) signals, as a function of the acquired image number. The values are normalized with respect to the grey levels corresponding with the required detector dose. Apart
from a transitional response at the onset of the run, the average level is virtually kept constant for both cases. This transitional response can be completely removed by setting the system initial conditions. Fig. 6 displays the object transfer function for both cases with a considerable overshoot for the AAGL case, which is essentially due to the lowered input signal upon sensing direct radiation in the measuring field. In order to evaluate the effect of the sensing method, the ratio between the transmission factors is shown in Fig. 7. The decrease will account for an increase in noise as can be evaluated by Eq. (2). The resulting Image Quality can be checked by the test phantom and yields a measured value of 20% increase in the Contrast-to-Noise Ratio for the geometrical averaging case. The calculated value was 23%. The previously mentioned percentages lead to a substantial increase of the dose in the dark areas giving a much better image quality. This result is confirmed by contrast-detail visibility results. Fig. 8 shows a qualitative effect of the method on the grey values for a lateral de-centered skull projection. The improvement is virtually independent of a limited off-axis position of the body. Clearly the levels are higher for the right part referring to the geometric average. Fig. 9 displays the effect on contrast resolution for a 3D reconstruction. The various densities are clearly better resolved as can be shown in the histogram for the geometric case on the right part of the image.

5. DISCUSSION AND CONCLUSIONS

Our study demonstrates that it is possible to ameliorate the effects of highlights entering a measuring field by using a geometric average for the control signal size, rather than rigorous highlight exclusion measures. The stabilized value of the detector dose in the region of interest guarantees a sufficient SNR. Summarizing, the new averaging technique balances the weighted distance between the average of the probability density subset of the interest region and the average of the highlight portion in the measuring field, so that the highlight areas are not favored. Consequently, the contrast-to-noise ratio of details with respect to their anatomical background is considerably improved.

The use of a signal norm based on the geometric average for dose control in a C-arc X-ray system equipped with a flat detector largely eliminates the detrimental effect of direct radiation entering a measuring field. The improvement can be readily explained by the analysis of the voxel noise and is confirmed by contrast-detail measurements.

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5. REFERENCES