DIGITAL HALFTONE TECHNIQUES IN MEDICAL IMAGING

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ABSTRACT

Conventional approaches to the evaluation of halftone techniques usually assess well known features such as the MTF or Moiré patterns. Applications to photographic images are commonly treated. However, in the specific case of halftone rendering of medical images, the physics of image production and noise sources lead to unique image characteristics. Furthermore, the use of the image by a radiologist is often case-specific. For example, the search for spinal cord damage in a magnetic resonance image requires different skills from those employed to search for metastases in the liver using an ultrasound B-scan instrument. In this report, we quantitatively assess the degrading effect of halftoning on one important task in medical imaging, the lesion detection problem. A simulated magnetic resonance image with lesions of varying size and contrast is halftoned using different techniques. Psychovisual experiments show the changes in detectability that result from halftone rendering.

INTRODUCTION

In three modern medical imaging modalities, computed x-ray tomography (CT), magnetic resonance imaging (MRI) and ultrasound (US), images are derived from calculations on extensive data. This is distinctly different from the case of photographic imaging or conventional x-ray imaging where film is directly exposed by a photon source. Furthermore, the physics and noise sources in CT, MRI, and US are unique and should be considered separately.

In medical imaging there are also unique and specific goals that may depend on the clinical situation. A common example is the lesion detection problem, where a slight difference in image characteristics within an organ such as the liver is scrutinized by a radiologist for possible diagnosis of a tumor or benign lesion. Because of the unique nature of medical images, and the different criteria for the evaluation of image quality, we argue that the effects of halftoning need to be evaluated, at least initially, in the context of specific modalities and specific tasks. In this paper we briefly review the physics and noise sources of CT, MRI and US imaging modalities. Then, a lesion detection model is developed with properties appropriate for MRI. This lesion detection "phantom" is then halftoned using clustered dot, dispersed dot, and error diffusion methods. The results of psychovisual experiments are presented to quantify the degradation of lesion detection caused by halftoning. We show that the more computationally intensive error diffusion method does result in an improved low-contrast detectability as compared to simpler clustered dot and ordered dither methods.

CT IMAGING

In computed x-ray tomography, profiles of x-ray energy transmitted through a cross section of the body are obtained at different angles. Using the projection-slice theorem, these profiles are used to calculate the 2-dimensional Fourier transform of the object's x-ray absorption image. Some blurring is caused by source and detector geometrics. Quantum noise, due to quantization of the energy of photons and limitations on cumulative exposure, is modelled as a Poisson process [1]. Thus, the power spectrum of a CT image has the following characteristics:

- limited coverage of transform space (centered around zero frequency) due to lowpass filtering effects,
- noise derived from Poisson processes.

MRI PHYSICS

This recent modality is built on a fundamental electromagnetic resonance that is exhibited by certain nuclei with magnetic moments. Hydrogen, which is abundantly present in tissues, is one such element with a magnetic moment, and resonance frequencies in the low Megahertz range are obtained using strong magnetic fields around 1 Tesla. MRI typically uses selective RF excitation, with frequency and phase encoding using magnetic gradients, to obtain data related to the density of hydrogen protons over a cross section. The RF data acquired from excited protons are encoded so that lines or curves in transform space are obtained. Since the time required to obtain data (and keep the patient motionless) increases with the area of transform space covered, the resulting coverage of transform space is usually limited, particularly along one axis which is referred to as the "phase encoding axis"[2] . Thus, the non-zero region of the image transform is a rectangle, with abrupt truncation of the data along the short, phase encoding axis. This can result in "ringing" artifacts in the image. Noise is primarily derived from the electronic RF detection circuits, and can in some cases be considered a white Gaussian noise process. Thus, the power spectrum of an MRI image has the following characteristics:

- data are available over a limited, rectangular region of transform space (centered around zero frequency),
- noise is derived from white Gaussian statistics.

US IMAGING

Conventional ultrasound images, termed "B-scans", are similar to sonar or radar images. A single pulse is sent out from a transducer along a line of sight into the body. Returning echoes from tissue structures are digitized and stored. After many lines of sight are collected, an image of the location and strength of reflectors can be made. Since the ultrasound pulses are coherent, band limited signals in the low Megahertz range, the resulting images have a "speckle" or "salt and pepper" texture that is also seen in certain radio and laser applications[3]. Thus, an US image has the following characteristics:

- strong influence of the "speckle" process, which is governed by the bandwidth and aperture of the ultrasound transducer,
- "speckle" noise effects plus electronic noise are present.

LESION DETECTION

Because each of these modalities has unique physics, noise processes, image and power spectra characteristics, test images or "phantoms" will be different depending on the modality studied. As an example, we have designed a lesion detection phantom for MRI.

In this image, the background is 35% gray, and the rows of circles are, from top to bottom, 36.5%, 38%, 39.5% and 41%. The diameter of circles are, out of 512 pixels across, 68, 48, 28, 16 and 8 pixels from left to right. The image is stored as an 8 bit, 512 x 512 array. To emulate the effects of an MRI scanner, the two dimensional DFT of this image was limited to a rectangular low-pass area (centered around the zero frequency) that was non-zero over 512 x 256 points, with a cosine taper and zero padding on the remaining (high frequency, phase encoding axis) transform data. Also, 15% white multiplicative noise was incorporated. The result is shown in Fig. 1, and in this case the noise appears to be a more significant artifact than any effects from the truncation of the transform. The detection of lesions is a well-studied problem, and it is known that within certain limits the detectability is proportional to the image contrast (difference in mean gray levels divided by the standard deviation of the noise level) and to the size of the lesion[1]. This can be verified by an inspection of Fig. 1 where smaller, low contrast lesions are more difficult to detect. Less well studied is the influence of halftone techniques which are known to degrade the image and distort the power spectrum. We quantify the loss of detectability that results from different halftoning techniques in the next section.

HALFTONE TECHNIQUES

We employ three different techniques to halftone the MRI lesion phantom of Fig. 1. These are:

- clustered dot dither, using a 45° screen and an 8 x 8 point threshold array.
- dispersed ordered dither using Bayer's method and an 8 x 8 point threshold array (5).
- error diffusion, using a 4 point filter with perturbed weights as described by Ulichney (4).

The results are given in Figs. 2, 3 and 4, respectively where the loss of detail compared to the original gray scale MRI phantom is evident.

QUANTIFICATION OF LESION DETECTABILITY

In a psychovisual experiment, photographic copies of Figs. 1, 2, 3, and 4 were shown in random order but under identical conditions to 11 (eleven) volunteers. The observers were told that not all lesions were present on each picture, and were asked to identify only those lesions that were detectable *with highest confidence*. The results are summarized graphically in plots of the limits of detectable contrast vs. lesion area (pixels), shown in Fig. 5. As expected, the detectability of low contrast areas in the gray MRI phantom image is degraded by halftoning. Note the absence of any value for clustered dot dithering for the case of lowest contrast (C = 0.25) lesions. This means that observers could not, with high confidence, identify any of the top row of lesions in the clustered dot image (Fig. 2). We also note that the error diffusion method is superior to the other halftone techniques in that smaller, lower contrast lesions can be detected.

Since the error diffusion method is known to have properties of "edge enhancement", some may argue that the superior performance in this lesion detection study is due to "sharpening effects" which could have been applied in separate steps before using other halftone techniques. However, this property of the error diffusion method is not likely to be significant in our study of low contrast detail. The white noise in the phantom image would be significantly enhanced by traditional "sharpening" filters, *decreasing* the detectability of small, low contrast lesions. Furthermore, the presence of noise in the low contrast lesions creates an irregular, poorly defined "edge" between the background and the lesion. For these reasons, we do not believe that the superior performance of the error diffusion method results from edge enhancement effects. More significant are the anistropic patterns produced at the different gray levels, with a lack of noticeable low spatial frequency components.

SUMMARY

Each of the major medical imaging modalities has unique noise and power spectrum characteristics, and require separate models and evaluations. In a lesion detection experiment using a simulated MR image, we found that halftoning caused reduced detectability, however the more computationally intensive error diffusion method resulted in better lesion detectability as compared to simpler halftoning methods. Additional work is required to evaluate the effects of halftoning on lesion detection in CT and US imaging.

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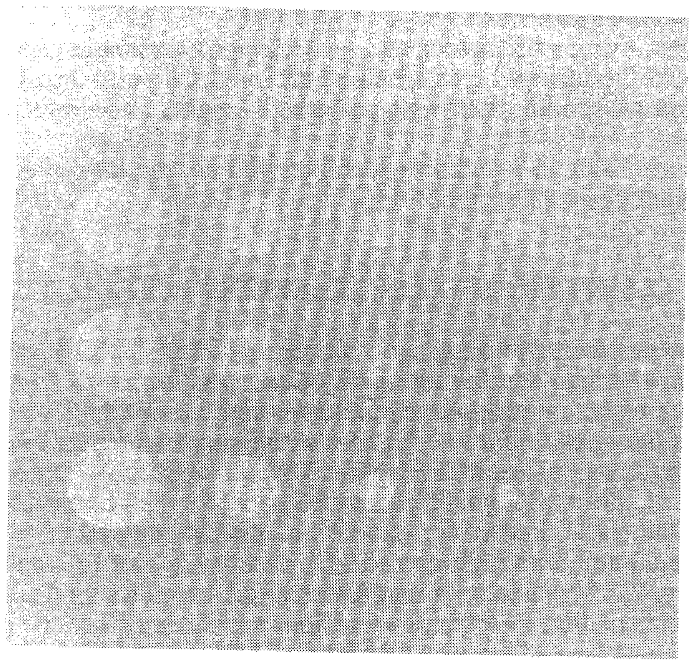

Figure 1 - Lesion Detection Phantom for MRI (originally shown as gray scale image). A series of low contrast "lesions" are placed in a 35% gray background. The image has a power spectrum of limited coverage in transform space, with white noise, in order to mimic the data collected by conventional MRI scanners.

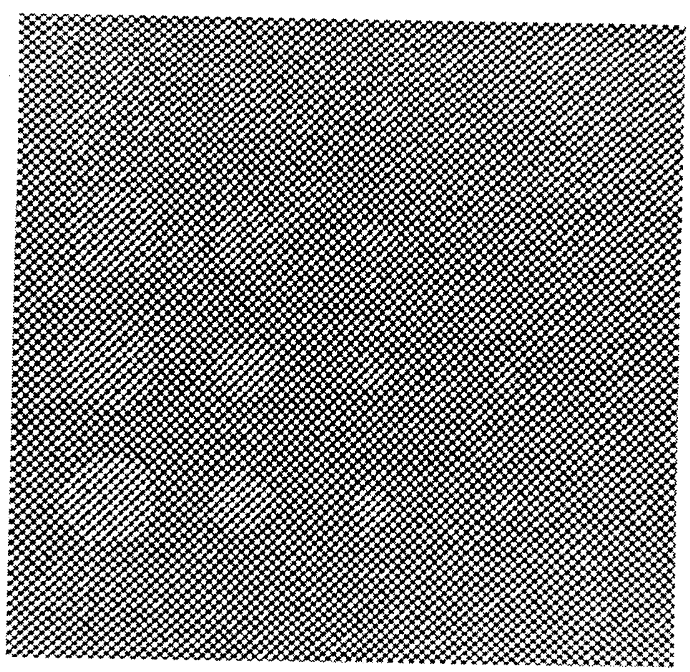

Figure 2 - Clustered dot dither applied to the phantom image.

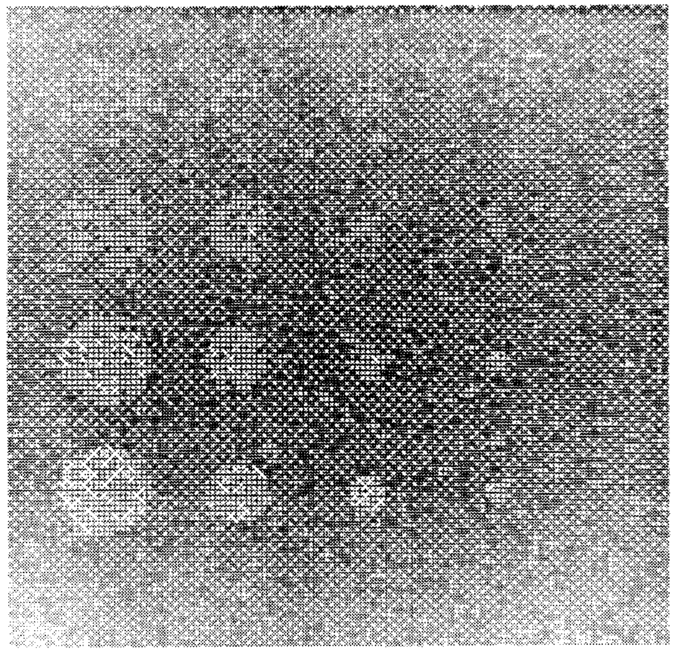

Figure 3 - Dispersed dot dither applied to the phantom image.

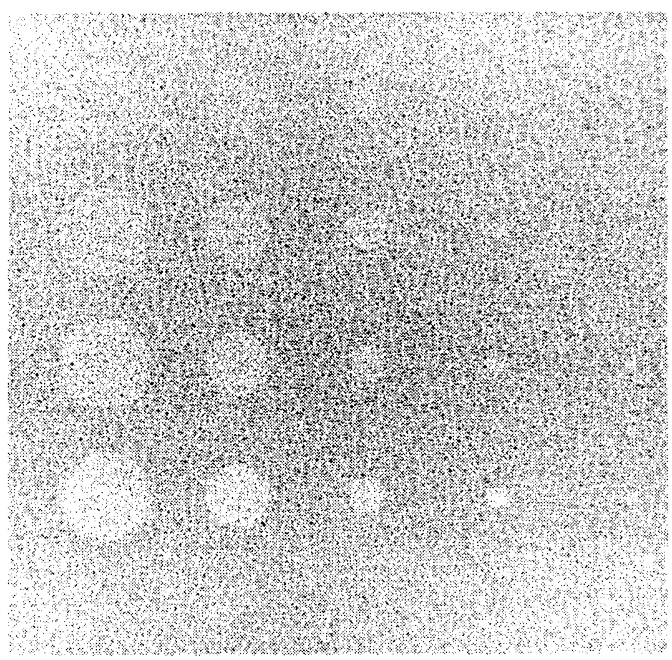

Figure 4 - Error diffusion dither applied to the phantom image.

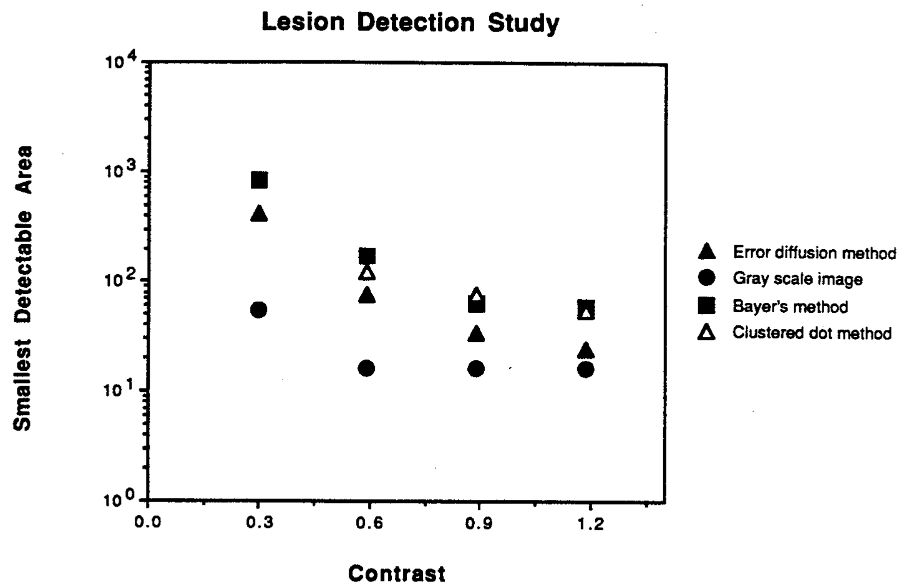

Figure 5 - Results of a psychovisual experiment. Eleven volunteers were asked to identify those lesions which were detectable with highest confidence. The vertical axis gives the smallest detectable area in pixels. The horizontal axis gives contrast in dimensionless units (see text for definition). Typical standard deviations (not shown) were ±30% of the mean value for any data point, over eleven volunteers.