Noise Reduction in Nuclear Medicine Images

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Common methods of reducing random noise in nuclear medicine use lowpass filtering, which has the disadvantage that it affects high-frequency components of the image. We developed a noise-reduction approach that estimates signal and noise levels in each of several frequency bands and removes the appropriate amount of noise with little effect on the signal in each band.

J Nucl Med 24: 532–534, 1983

Nuclear medicine images produced in rapid sequence often suffer from high noise levels caused by poor counting statistics. The noise has a wide bandwidth, often extending to frequencies above the signal. Since the high-frequency components of noise are most distracting to the viewer (*1*,*2*), low-pass filtering (smoothing) has been used in an effort to improve noisy images (*3*,*4*). In this approach, the filter's cutoff frequency is chosen as a compromise between removal of noise from the image and deletion of high-frequency components from the signal.

A more satisfactory approach to image processing would be to use an algorithm tailored to the power spectra of signal and noise. In frequency bands with no signal, the algorithm would eliminate the noise. In frequency bands with high noise and low signal, both the signal and the noise would be removed. In bands with similar signal and noise amplitudes, both would be replaced by sine functions of appropriate amplitude, resulting in zero noise. This approach to noise suppression would be most successful when the frequency bands are narrow.

METHOD

Figure 1 shows the approach used in this experiment. At 1 the image to be processed is acquired as two half-count images; for example, an image of 100,000 counts would be acquired either as two sequential 50,000-count images or as alternate counts entered into two image memories until each contained 50,000 counts. These two half-count images are identical except for noise: if the images were noiseless, corresponding pixels would have identical counts, whereas noisy images would have relatively high count differences in corresponding pixels.

At 2 each half-count image is transmitted without prior processing through a series of octave rectangular band-pass filters that cover the spatial frequency range of the signal. The filtered image pairs are both added and subtracted at 3. The maximum count difference found between corresponding pixels at 3 is an unbiased indicator of the peak image noise in the octave being analyzed, and is used at 4 as described below.

The filtered and summed images are passed through an operator at 4, which removes low-amplitude signals but transmits unchanged all signals with an amplitude above a selected threshold. Figure 2 shows the transfer function of this operator, and Fig. 3 shows the effect of the operator on one dimension of the image. The dead zone Z_D of the operator is set by the maximum count difference between corresponding pixels determined at 3. Pixel count differences in the filtered image that have lower amplitude than $\pm Z_D$ are likely to represent noise and are removed by the operator. Pixel count differences outside $\pm Z_D$ are preserved, but the count-against-distance function now has steep edges, with resulting high-frequency components.

The undesirable high-frequency components generated by the nonlinear operator at 4 are removed by bandpass filters at 5. These filters have cutoff frequencies equal to those of the filters at 2. A frequency range of 1 to 16 cycles per image is covered by four bands, each one octave wide. That is, the filter bandwidths are 1–2, 2–4, 4–8, and 8–16 cycles per image. The outputs of the filters at 5 are added at 6 to form the final processed image.

In Figure 4, a noiseless image is shown with a sinusoidal count profile along any radius from the center of the bull's-eye. A noisy version of this image was developed by adding two half-count noisy images, each generated as follows.

$$[I_N] = [N] \times [\sqrt{[I]}] + [I],$$

where $[I]$ = the noiseless image,
$[I_N]$ = the noisy image,
$[N]$ = A noise pattern consisting of a Gaussian distribution with mean = 0 and standard deviation = 1.

A similar approach was used to generate the noisy image shown in Fig. 5 from the noiseless image. All images are displayed in 64 × 64 format. In the noiseless image, the average count per pixel

Received Oct. 29, 1982; revision accepted Feb. 17, 1983.

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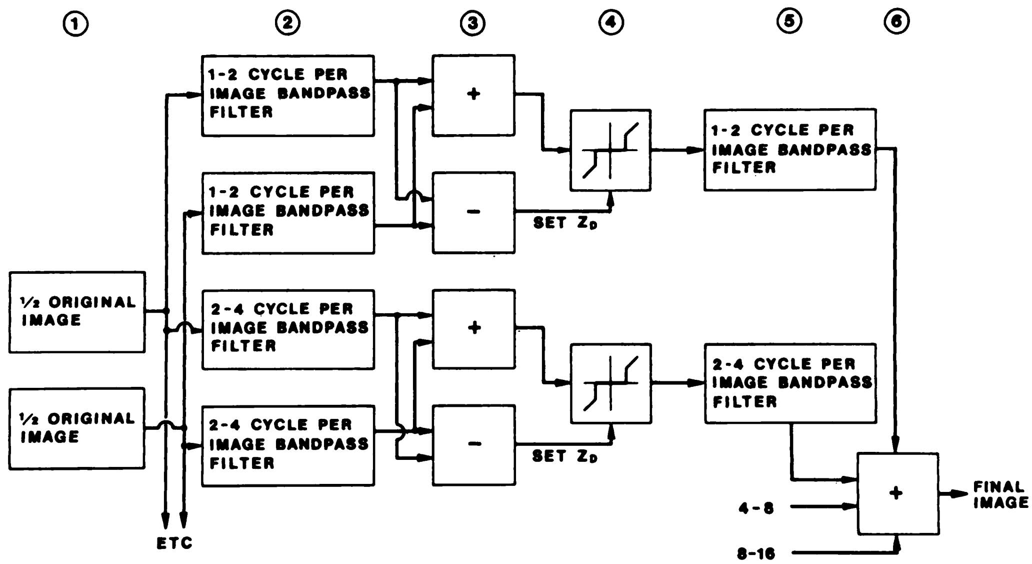

FIG. 1. Algorithm for reducing noise in nuclear medical images.

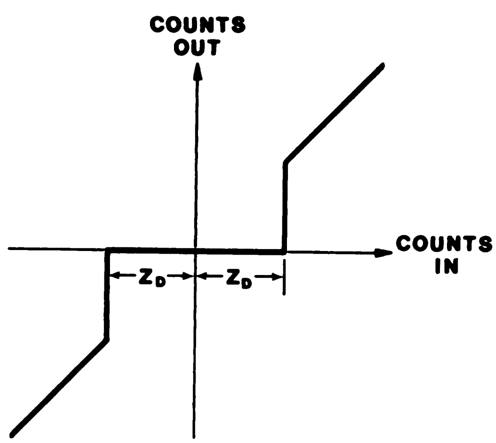

FIG. 2. Transfer function of operator at step (4) in Fig. 1.

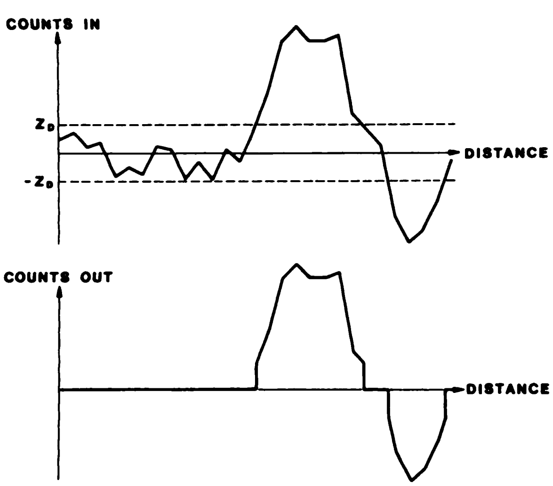

FIG. 3. Effect of operator at step (4) in Fig. 1 on the image.

is 5. Therefore, each half-count image has a noise level comparable to a nuclear medicine image of 20,480 counts (5 × 64 × 64), and the summed image [Fig. 4(center) or 5(center)] is comparable to a 40,960-count nuclear medicine image.

RESULTS

Figures 4(right) and 5(right) show the processed images corresponding to Figs. 4(center) and 5(center). The sinusoidal signals are preserved, and the distracting elements of noise are reduced. In regions of Fig. 5 with no sinusoidal signal, noise of all frequencies is greatly reduced. When noise-free images (Figure 4(left) and 5(left) were processed, the only artifact introduced was a minimal low-frequency irregularity in intensity, most noticeable near the edges.

DISCUSSION

The method described here appears to offer promise as an effective approach to removing noise in nuclear medicine images with significantly different noise and signal amplitudes. It is similar to matched filtering (*5*), but the nonlinear operator removes information from regions and frequency bands with a low signal-to-noise ratio. Image noise power is reduced to the extent that noise exists without significant signal amplitude in each octave. Because the attenuation outside the filters' bandpasses is finite, the method may be most appropriate for simple images having relatively well-defined spikes in their frequency spectra; the improvement in more complex images may be less impressive. In these early efforts we applied the algorithm to single images in the spatial domain, but it could also be applied to sequential images in the time domain. For example, individual images in a gated blood-pool study could be processed spatially, then each pixel could be processed temporally.

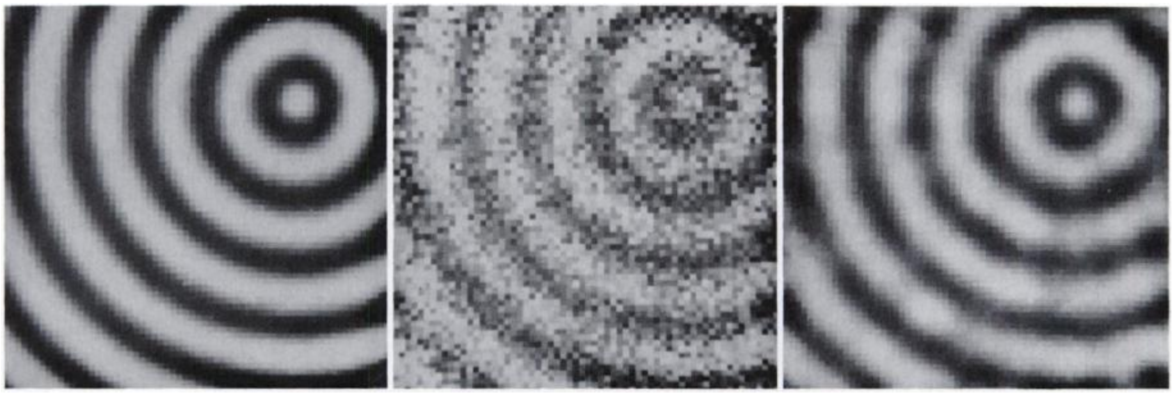

FIG. 4. Noiseless image (left). Count profile from center of bulls-eye radially is sinusoidal. Same as left but containing random noise (center). Final processed image (right).

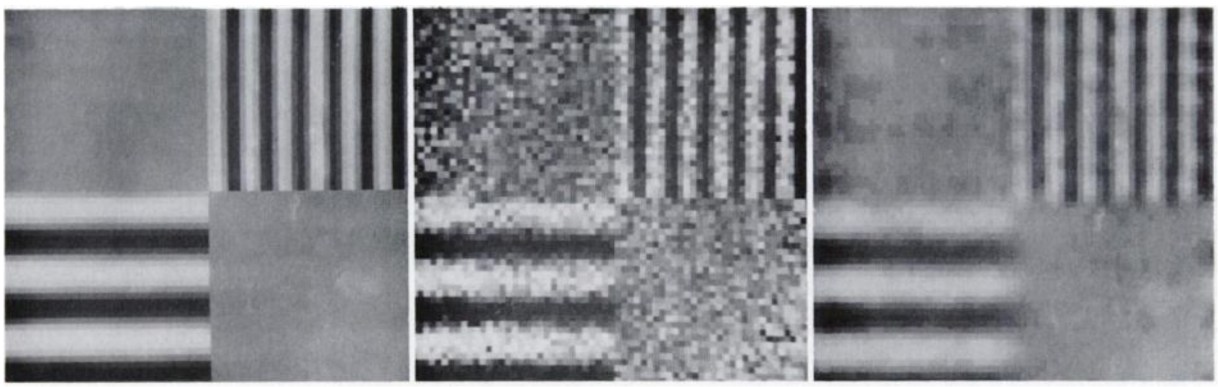

FIG. 5. Noiseless image (left). Count profiles across banded regions are sinusoidal. Same as left but containing noise (center). Final processed image (right). Regions of constant activity in Fig. 5(left) have minimal noise.

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