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A MEDICAL FEATURE ENHANCING SPECKLE SUPPRESSION TECHNIQUE BASED ON A NOISE MODEL FOR THE SPECKLE NOISE AS REPRESENTED IN THE B-SCAN IMAGE

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ABSTRACT

In this paper we introduce a new speckle suppression technique for medical ultrasound images, based on a noise model for the speckle noise as represented in the B-scan image. Using this model and prior statistics ensuing from it we are able to distinguish speckle noise from important resolvable details. All individual speckles are located, and exploiting our knowledge on the speckle classification, the filter removes the noise speckles, while it preserves valuable details. We conclude with comparing the results of the proposed technique with two classical speckle suppression methods.

keywords: ultrasound, speckle, morphology, filter, neonatal brain

1. INTRODUCTION

Ultrasonic imaging is a widely used, non-invasive, quick and cost effective way to image soft tissues. Unfortunately the image quality is severely degraded by the presence of speckle noise. The suppression of this noise is a particularly delicate and difficult task. The problem is that for an experienced radiologist speckle noise may present useful diagnostic information [1, 2]. This leaves us with the task to suppress the noise without losing valuable medical information.

Several techniques for suppressing speckle noise have been developed [3–8]. Most of them involve the use of general speckle statistics, and not all are equally well-suited for medical ultrasound images, because they do not take into account the special features typical for medical images. Examples of such features are: bright large scale interfaces between organs, structures with dimensions comparable to the speckle size, and boundaries between two regions with slightly different grey-levels.

The new speckle suppression method we present in this article, is designed to remove a lot of speckle noise, while preserving valuable medical image data. It is based on the following noise model, proposed in [7], which is adopted for speckle noise in B-scan images:

$$x = s + n\sqrt{s}.$$

Herein represent: x the observed signal, s the noise-free signal, and n the noise (independent of x, with mean 0) [7].

This model is particularly suitable for our purpose, since the presented filter works on the images as displayed by the ultrasound machine, rather than the envelope detected echo signal. Wagner et al. have shown that the histogram of amplitudes within the “resolution cells” of the envelope detected RF signal backscattered from a uniform area with a sufficiently high scatterer density has a Rayleigh distribution with mean μ proportional to the standard deviation σ ($\sigma/\mu = 1.91$) [9, 10]. This implies that speckle could be modelled as multiplicative noise. However, the signal processing stages inside the scanner (mainly logarithmic compression in order to adjust the large echo dynamic range (60-100 dB) to the 8-bits of the digitization in the scan converter) modify the statistics of the original signal. In [7] the effects of these nonlinear operations have been studied and the above mentioned model has been demonstrated to hold for the final images. In particular, it should be noticed that speckle is no longer multiplicative in the sense that on homogeneous regions, where s can be assumed constant, the mean is proportional to the variance rather than the standard deviation.

2. MEASUREMENTS

Although this new speckle suppression technique can basically be applied to all kinds of ultrasound images, we restrict ourselves in the rest of our discussion to ultrasound images of the neonatal brain. All of them were taken at a frequency of 7.5 MHz.

As said, the model described in section 1 implies that on a uniform area r the ratio $\alpha_r = \sigma_r^2/\mu_r$ of the variance and the mean of the grey-values of the pixels within r is constant.

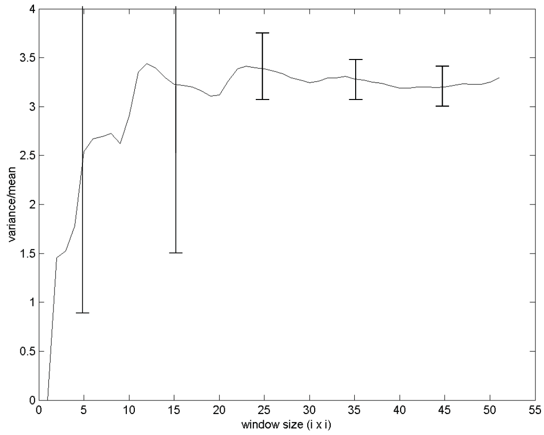

Figure 1: Dependence variance/mean on window size.

We will determine the exact value of α_r in our images experimentally, and decide the minimum dimensions the area r needs to be in order to calculate α_r reliably.
In 10 brain images showing a uniform area near the focal zone of the sound beam we performed the following experiment: in each image we selected 51 windows V_i $(i = 1, \ldots, 51)$ located in this uniform area, varying in size from 1×1 to 51×51 pixels and we calculated the values $\alpha_i = \sigma^2_{V_i}/\mu_{V_i}$. In figure 1 the averages $\overline{\alpha_i}$ of the 20 α_i's for each i are plotted. The vertical bars indicate the variance. As can be seen the value of $\overline{\alpha_i}$ converges to 3.24 when i (i.e., the size of the window V_i) increases. Furthermore the variance drops below 0.5 when V_i grows bigger than 22×22 pixels. This means that we must consider an area of at least a size of 22×22 pixels to be able to determine reliably whether a uniform speckle pattern is displayed at that location. In [11] similar results were found for images of the liver.

3. FILTER

The method proposed in this paper follows the new approach of first attempting to segment each individual speckle, after which each speckle is classified as a "noise speckle" or as an important resolvable detail. The segmentation of the individual speckles is achieved by a region growing procedure based on the grey-values of the pixels. In order to regulate the shapes of the regions grown, we preprocess the image morphologically by constructing the "top-hat transform", using the morphological structure element of figure 3, which thereby serves as a kind of "ideal speckle". This top-hat transform consists of the collection of foreground parts from the original image that fit the structure element. This structure element was constructed by manually selecting and averaging six speckles from near the focal zone. The top-hat

Figure 2: "Top-hat transform" of image in figure 4, with structure element as displayed in figure 3 .

0	0	0	0	0	0	0	0	0	0	0	0
0	0	0	0	1	1	1	1	0	0	0	0
0	1	1	1	1	1	1	1	1	1	1	0
0	0	1	1	1	1	1	1	1	1	0	0
0	0	0	0	0	0	0	0	0	0	0	0

Figure 3: Morphological structure element.

transform of figure 4 is shown in figure 2.
The filter itself works as follows:
1) First we determine all local maxima in the preprocessed image above a prefixed lower threshold Λ, and put those in an array M.
2) Let Γ be the highest of the grey-values of all elements in M. Choose one of those maxima (i, j) with grey-value Γ. This pixel serves as a seed pixel for a region growing procedure, controlled by the grey-values of the pixels. In short: a pixel (m, n) belongs to the region of a seed pixel (i, j), when the following criteria are satisfied:
- Pixel (m, n) is "connected" to pixel (i, j),
- $\alpha_{(m,n)} > \alpha_{(i,j)} - T$, where T is a prefixed threshold,
- Pixel (m, n) does not belong or is adjacent to a speckle which has already been grown. In this way we segment one single speckle Σ.
3) Next we determine the centre of gravity g_Σ of Σ. Take a square V_Σ of 22×22 pixels around g_Σ in the original image, calculate $\alpha_\Sigma = \sigma^2_{V_\Sigma}/\mu_{V_\Sigma}$.
4)If $\alpha_\Sigma \leq 3.24$ then, as can be inferred from the measurements discussed in section 2, the speckle is located in a uniform area, and hence is a noise speckle that should be removed. We calculate the mean ν_Σ of the grey-values of all the pixels adjacent to Σ, and give all pixel of Σ the value ν_Σ. In this way we actually "cut off" the speckle.

Figure 4: Original image.

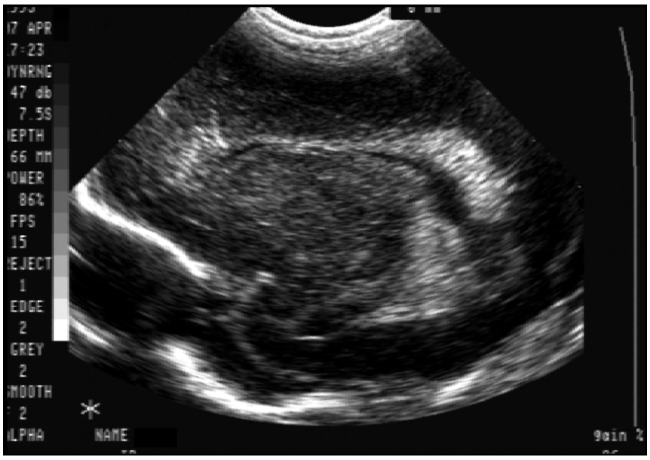

Figure 5: Lee filter.

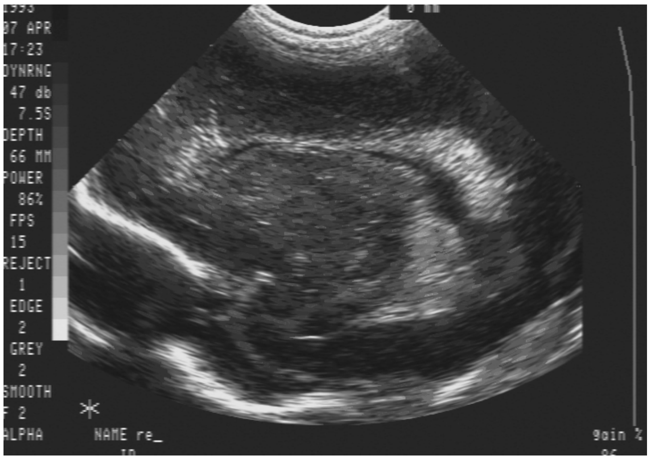

Figure 6: Proposed method.

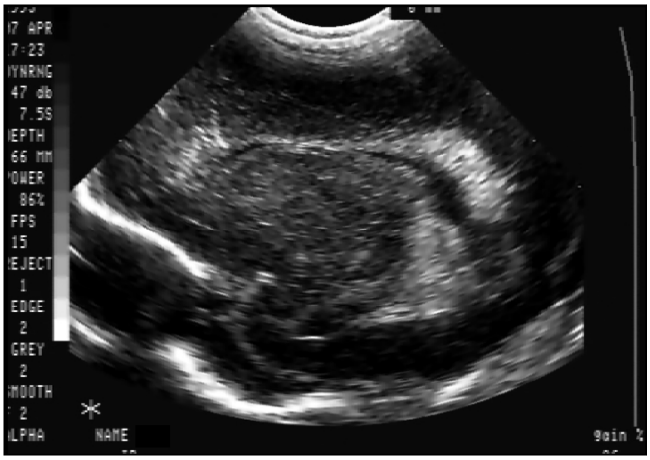

Figure 7: Frost filter.

5) If $\alpha_\Sigma > 3.24$, we conclude that the speckle is a resolvable detail, and hence the speckle Σ needs not be removed.
6) All local maxima (from M) that belong to Σ, or are adjacent to Σ, are removed from M.
7) We repeat the whole process from step 2 until M is empty.
Both T and Λ are adjustable parameters of the filter, and are dependent on the exact qualities of the ultrasound machine. For our brain images, we obtained the best results with $\Lambda = 63$ and $T = 17$.

4. EXPERIMENTAL RESULTS

We applied the technique on the images of figures 4 and 10. The result of the technique are depicted in figures 6 and 11 respectively. As can be seen a considerable speckle suppression is achieved, while medically relevant details (like the tissue interfaces indicated in figure 10) are well-preserved. In figures 8 and 9 the speckles, as they are found by step 1), are indicated. The ones classified as noise (i.e., located in a uniform area) are coloured red, the other ones are coloured blue. In the figures 5, 7, the fist image has been processed with two classical speckle suppression techniques, namely the Lee and the Frost filter [3, 4]. The Lee filter generally suppresses speckle, but has the disadvantage that, on some places, it tends to enhance a speckle. The Frost filter tends to blur the image slightly. The proposed method does have neither of those disadvantages.

5. CONCLUSIONS

In this paper we introduced a denoising technique for medical ultrasound images. As opposed to most techniques developed so far, it is based on a model of the speckle statistics as displayed in the image, after the various signal processing steps in the scan converter, rather than on the RF signal. The method successfully reduces speckle noise in ul-

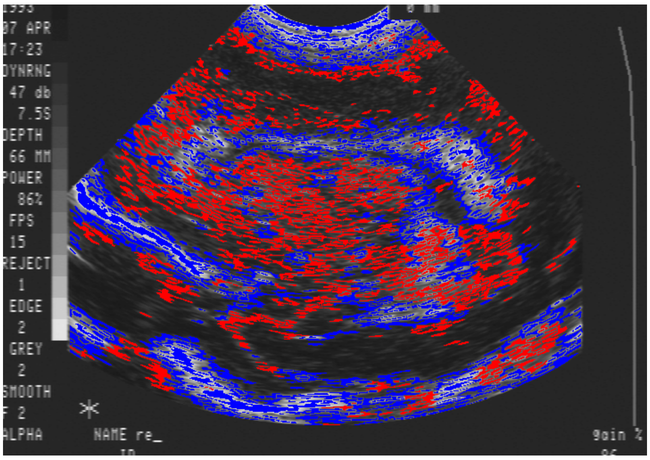

Figure 8: Classified speckles of image 4.

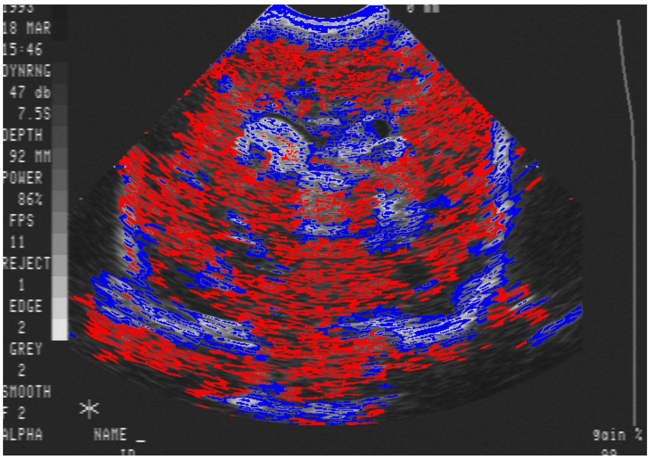

Figure 9: Classified speckles of image 10.

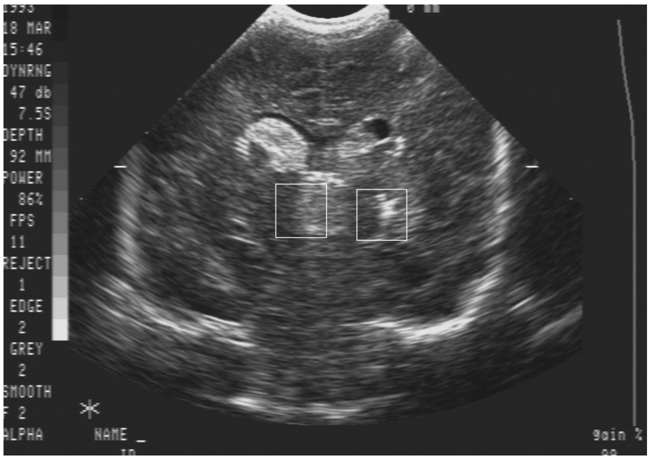

Figure 10: Original image.

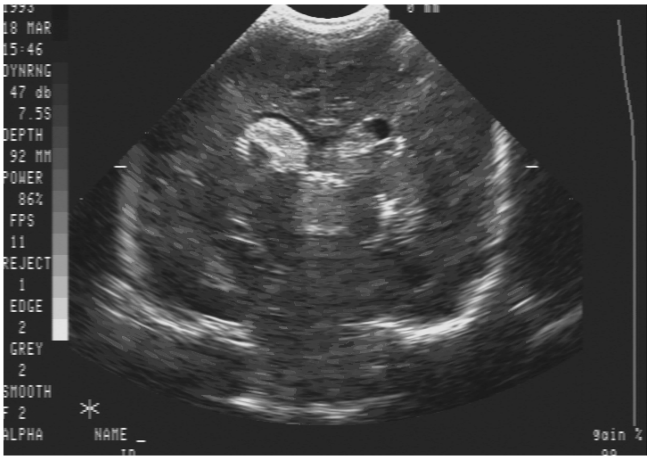

Figure 11: Proposed method.

trasound images, and performs comparative if not better than several classical techniques. Finally the method is computationally not intensive: processing a image of 722×506 pixels (the size of the example images) on a Pentium III (600 MHz) takes less than 1 second.

6. REFERENCES

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