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Frequency equalized compounding for effective speckle reduction in medical ultrasound imaging

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ABSTRACT

Frequency compounding (FC) is commonly used to reduce the speckle variance in order to enhance contrast resolution by averaging two or more uncorrelated sub-band images. However, due to the frequency dependent attenuation, the contrast resolution cannot be enhanced to the theoretical limit when imaging deep-lying tissue. In this paper, we propose the frequency equalized compounding (FEC) method to achieve contrast enhancement in the area of imaging as a whole. In this proposed method, a sub-band signal is divided into several zones along the imaging depth (or time), and the center frequencies and weighting factors for each zone are estimated; the estimated values are used in dynamic quadrature demodulation (DQDM) and image compounding respectively. The performance of the proposed method was evaluated through simulations and experiments. During the evaluation, the contrast resolution was quantified by speckle's signal-to-noise ratio (SSNR) in speckle regions and contrast-to-noise ratio (CNR) in hyper- and hypoechoic regions. Theoretical values of the SSNR and the CNR by the FC were computed by multiplying the SSNR and CNR values measured from the original image by $\sqrt{N}$, where N is the number of sub-bands used in the compounding. From in vitro phantom experiments, it was learned that the SSNR and CNR values from the proposed method were similar to the theoretical values; the maximum and minimum errors from the theoretical value were 9% and 1% while those of the conventional FC (CFC) method were 25% and 7%. Similar results were obtained from the in vivo experiments with RF data acquired from the liver and the kidney. In addition, signal-to-noise ratio (SNR) improvement was measured. The SNR also improved due to the DQDM; maximum improvements for the in vitro and the in vivo experiments were 2.3 dB and 4.8 dB higher the results from the CFC method. These results demonstrate that the proposed FEC method can improve the contrast resolution up to a theoretically achievable value and may be useful in imaging technically difficult patients.

1. Introduction

Medical ultrasound imaging is widely used as a non-invasive diagnostic imaging modality due to its several advantages such as safety, real-time imaging capability, portability, and relatively low cost. However, it generally suffers from low contrast resolution due to small difference of characteristic acoustic impedance (~1%) between different soft tissues [1]. Furthermore, speckle reduces the ability to detect low-contrast targets because the speckle often appear surrounding the targets as granular patterns with different sizes and brightnesses. The speckle pattern results from the inherent nature of coherent imaging systems [2]. The speckle is typically regarded as noise due to the degradation of structure detection in medical ultrasound imaging although these are frequently used in clinical diagnoses. The speckle patterns affect not only human interpretation of diagnoses but also the accuracy of computer-aided detection/diagnosis (CAD). One tricky problem is that the noise reduction methods, such as simple averaging and sophisticated system designs for low acoustic and electronic noises, do not actually work well in mitigating the pattern. This is because the same speckle patterns consistently appear when under the same operating conditions.

Speckle noise can be reduced by using frequency compounding (FC) and/or spatial compounding (SC) at the expense of spatial and/or temporal resolution [3–13]. These compounding methods are generally based on the average of a series of uncorrelated or partially correlated sub-images that have different speckle patterns. As a result, the compounding methods can improve the speckle's

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signal-to-noise ratio (SSNR) and contrast-to-noise ratio (CNR) by a factor of $\sqrt{N}$, where N is the number of uncorrelated sub-images used in the compounding [12,13].

In SC, the sub-images are acquired at different beam orientations at the expense of spatial (i.e., lateral) and temporal resolutions [3–5]. Under the same loss of spatial resolution, it was ascertained this method provided a higher speckle reduction ratio compared to the FC method [7]. However, since multiple transmissions are required for acquiring sub-images in SC, this method is only suitable to image stationary or slow-moving objects such as in breast imaging.

In FC, on the other hand, uncorrelated sub-images are obtained by either varying the center frequency on transmissions or by dividing the spectrum of radio-frequency (RF) signal on reception [6–15]. The latter method is also called frequency diversity [9]. Since the conventional FC (CFC) image is produced by dividing a signal spectrum into sub-bands and averaging the sub-band images, the loss of axial resolution is inevitable. Furthermore, the efficacy of the CFC method is lowered due to the frequency dependent attenuation of ultrasound; since higher sub-band signals experience more attenuation, lower sub-band signals become dominant when those sub-band signals are averaged [14]. As a result, spatial resolution and SNR of the CFC images are severely degraded, particularly when imaging deep-lying tissue. In fact, the loss of signal energy stems from both the frequency dependent attenuation and the diffraction of ultrasound as it propagate through the tissue. A time gain compensator (TGC) can be used to secure uniform brightness of ultrasound images by compensating for overall decreases in ultrasound energy along the imaging depth. However, it is not suitable to compensate for asymmetric decrease in signal spectrum due to the frequency-dependent energy loss.

To alleviate the problems of the CFC, the wavelet packet decomposition was utilized to divide the received RF data into several sub-bands (e.g., 16 sub-bands) and to suppress signal noise by applying a soft threshold algorithm to wavelet coefficients [10]. Due to its efficient noise suppression, this method is capable of enhancing SNR and CNR, especially, in deep area. However, speckle noise reduction is inefficient because of a relatively high correlation between the sub-band signals, which originates from inherent characteristics of the wavelet transform. Furthermore, this method used Hilbert transform (or Hilbert filtering) to obtain envelope signals. The Hilbert filtering may be viable for B-mode imaging, but there is a large concern about the imbalance problem in Doppler and color flow imaging [16].

Also, Erez et al. proposed the depth-dependent FC method to address this issue [11]. In this method, based on assumed attenuation coefficients, lower weighting factors are applied to higher frequency sub-bands while higher weighting factors are used at lower frequency sub-bands as depth increased. However, the efficacy of compounding in such a method would be reduced since the attenuation coefficient varies in soft tissues as a function of imaging depth and only lower sub-band signals contribute to the majority of compounding in the deep tissue due to their dominant energy. This brings about a considerable loss of contrast resolution in deep tissue imaging.

Previously, we proposed the adaptive frequency compounding method in which the speckle energy distribution (SED) of the received RF signal along the imaging depth is analyzed using the short time Fourier transform (STFT) [15]. Based on the SED, the compounding weighting factors are computed to produce identical energy at each sub-band. The method could improve the contrast resolution compared to the CFC method. However, STFT is computationally expensive and thus hampers the real-time implementation. In addition, this method assumed a fixed center frequency, so that its performance would be limited in the deep imaging area. This is because the frequency dependent attenuation leads to not only energy loss at higher sub-band but also center frequency downshift of backscattered ultrasound waves [17]. As a remedy, therefore, this paper propose the frequency equalized compounding (FEC) method to be capable of effective speckle reduction even in deep imaging area.

Table 1
Acronyms.

CAD	Computer-aided detection/diagnosis
FC	Frequency compounding
SC	Spatial compounding
SSNR	Speckle's signal-to-noise ratio
CNR	Contrast-to-noise ratio
RF	Radio-frequency
CFC	Conventional frequency compounding
FEC	Frequency equalized compounding
BPF	Band-pass filter
DQDM	Dynamic quadrature demodulation
PSD	Power spectrum density
SNR	Signal-to-noise ratio
Short time Fourier transform	STFT

In the proposed method, the center frequency downshift of echoes due to the frequency dependent attenuation is estimated. The estimated values are used not only in the design of depth-dependent band-pass filters (BPFs) for the spectral division of the echoes but also in dynamic quadrature demodulation (DQDM) for down-mixing. With this method, we can avoid the high contribution of noisy signals in a higher sub-band. In addition, the weighting factors are determined to be inversely proportional to the estimated power spectrum density (PSD) of each sub-band along the imaging depth. The performances of the proposed method were evaluated through simulation, in vitro phantom and in vivo experiments with RF data acquired by a commercial ultrasound machine. The acronyms used in this paper are summarized in Table 1.

2. Methods

In Fig. 1, a block diagram of the proposed FEC method is illustrated. The gray boxes represent the unique functional blocks of the FEC method, i.e., the estimation of center frequency and sub-band weighting factors. The beamformed RF signals, $r(n)$, are sent to the dynamic center frequency estimator in which the signals are divided into a certain number of subsets along the imaging depth and the center frequencies of each subset are estimated. The estimated center frequencies are used to determine the center frequency of the depth dependent BPFs (i.e., dynamic BPFs) which divide the spectrum of $r(n)$ into sub-bands. In addition, these estimated center frequencies are used in DQDM for the down-mixing of sub-band signals. In the sub-band weighting factor estimator, PSDs of the subset baseband signals are computed in each sub-band processing block and used to obtain weighting factors. The envelopes of each sub-band signal are multiplied by the weighting factors and added together in order to generate a compounded image. The theoretical ground of the unique functional blocks will be explained below in greater detail.

2.1. Dynamic center frequency estimation

The received RF signal after beamforming can be represented by [17]

$$r(n) = A(n)\cos(2\pi f_c(n)nT + \phi(n)), \quad (1)$$

where $A(n)$ is the amplitude of the RF signal, $f_c(n)$ is the time-varying center frequency, T is the sampling period, and $\phi(n)$ is the phase. The analytic representation of $r(n)$ can be expressed by

$$r_a(n) = I_{bp}(n) + jQ_{bp}(n), \quad (2)$$