



Automatic dynamic range adjustment for ultrasound B-mode imaging



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ABSTRACT

In medical ultrasound imaging, dynamic range (DR) is defined as the difference between the maximum and minimum values of the displayed signal to display and it is one of the most essential parameters that determine its image quality. Typically, DR is given with a fixed value and adjusted manually by operators, which leads to low clinical productivity and high user dependency. Furthermore, in 3D ultrasound imaging, DR values are unable to be adjusted during 3D data acquisition. A histogram matching method, which equalizes the histogram of an input image based on that from a reference image, can be applied to determine the DR value. However, it could lead to an over contrasted image. In this paper, a new Automatic Dynamic Range Adjustment (ADRA) method is presented that adaptively adjusts the DR value by manipulating input images similar to a reference image. The proposed ADRA method uses the distance ratio between the log average and each extreme value of a reference image. To evaluate the performance of the ADRA method, the similarity between the reference and input images was measured by computing a correlation coefficient (CC). In *in vivo* experiments, the CC values were increased by applying the ADRA method from 0.6872 to 0.9870 and from 0.9274 to 0.9939 for kidney and liver data, respectively, compared to the fixed DR case. In addition, the proposed ADRA method showed to outperform the histogram matching method with *in vivo* liver and kidney data. When using 3D abdominal data with 70 frames, while the CC value from the ADRA method is slightly increased (i.e., 0.6%), the proposed method showed improved image quality in the c-plane compared to its fixed counterpart, which suffered from a shadow artifact. These results indicate that the proposed method can enhance image quality in 2D and 3D ultrasound B-mode imaging by improving the similarity between the reference and input images while eliminating unnecessary manual interaction by the user.

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1. Introduction

Ultrasound B-mode imaging is useful for assessing a lesion in real time due to its ability to produce high resolution images. As a result, ultrasound B-mode imaging has become one of the most widely-used diagnostic tools for various clinical applications such as radiology, cardiology, and obstetrics and gynecology [1]. However, ultrasound imaging suffers from low reproducibility and its image quality is highly skill dependent because there are a large number of parameters to adjust for achieving a high-quality image. One of the most frequently-used parameters in ultrasound B-mode imaging is dynamic range (DR) which is defined as the ratio of the largest and smallest signal levels [2].

In medical ultrasound imaging, the total DR of detected B-mode signals can exceed 150 dB after low-noise amplification followed by depth-dependent amplification in analog front-end circuitries and receive beamforming. This signal range is too large to be properly presented on a display monitor for the human eye as its DR value is limited to the order of 30 dB [3]. Thus, in ultrasound B-mode imaging, the DR of the envelope signal is generally compressed by applying a nonlinear signal compression technique, such as log compression [4].

In log compression, weak signals that contain clinically-valuable information (e.g. boundaries of organs and margins of lesions) can be clearly displayed if the proper DR value is selected with strong signals from various specular reflectors, like the diaphragm or large vessel walls. However, the improper selection of the DR value can make an ultrasound B-mode image too flat or over contrasted. In addition, the DR value substantially modifies pixel intensities and image contrast. As a result, the outcome of ultrasonic image processing techniques is greatly affected based on the gray level of the pixel, such as texture analysis and

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segmentation [5]. In conventional ultrasound B-mode imaging, the user needs to manually adjust the DR value (e.g. 45–65 dB) depending on clinical applications and the patient's condition to achieve a high-quality image. While this approach may allow a user to select an optimal DR value, the image quality is highly dependent on the respective user's skill. Because this adjustment must be manually conducted, clinical productivity may be lowered.

To determine the proper DR value for ultrasound B-mode imaging, various methods have been proposed [2,3]. In [2], the cost function based on edge contrast and soft tissue roughness is utilized for selecting the DR value. While this cost function-based approach is able to calculate the DR value in real time, it is difficult to validate whether improved image quality is actually guaranteed with the associated high cost values. On the other hand, in [3], to choose the proper DR value and its corresponding compression map, a display signal is first generated. This display signal utilizes the full dynamic range of the envelope signals while eliminating noise level signals. Then, the obtained display signal is given to the user to choose an optimal DR value for reconstructing a high-quality ultrasound B-mode image. This approach is useful for analyzing the envelope signal to determine the DR value, however, it is difficult to adaptively determine such values while scanning.

In 3D ultrasound imaging, multiple 2D ultrasound images are first acquired using a 1D linear or convex transducer with free-hand scanning. Then, the acquired 2D images are combined with 3D scan conversion, and volume rendering or multi-planar reconstruction algorithms are applied to display 3D volume data. Alternatively, a 1D mechanical array transducer or a 2D array transducer may be used for enhancing the quality of 3D volume data. During 3D volume scanning, it is impossible to adjust the DR value. Thus, the automatic selection of the DR value becomes more important in 3D ultrasound B-mode imaging.

In this paper, a new method is presented, which automatically adjusts DR for B-mode ultrasound imaging. In the proposed Automatic Dynamic Range Adjustment (ADRA) method, the distance ratio between the log average and each extreme value (i.e., minimum and maximum values) of a predetermined reference image is used for manipulating input images to get similar brightness and contrast results. To evaluate the performance of the ADRA method, the correlation coefficients (CCs) between the histograms of the reference image and input images were computed with *in vivo* abdominal data.

2. Methods

2.1. Automatic Dynamic Range Adjustment (ADRA) method

Fig. 1 describes the block diagram of the proposed ADRA method. In ADRA, a reference image with a predetermined DR value is required. After removing the DC component from the beamformed data of the reference image, the envelope signal is extracted with the demodulation process (e.g. quadrature demodulation). From the envelope signal of the reference image, the log average is computed by

$$L_w = \exp \left(\frac{1}{N} \sum_{x,y} \ln(\delta + L_w(x,y)) \right), \quad (1)$$

where N is the total number of data points in the image, $L_w(x,y)$ is the envelope signal at a spatial location (x,y) and δ is a small value to avoid the divergence when $L_w(x,y)$ is zero [6]. The log average is the geometric mean, which is affected less by outliers than the arithmetic mean or median values [7]. This log average is first calculated with the full range of the envelope signal. Then, the ratio of the distance from log average to the minimum and maximum values is computed with the reference image.

In the proposed ADRA method, the minimum value is selected within the range of a conventional DR value (e.g. 60 dB). This process is described in the upper image in Fig. 2. The ratio of the distance is indicated by the letters A and B. The ratio between A and B is the key parameter from the reference image and will be used for adjusting the DR of input images. With the input image, the log average is computed in similar terms with the reference image. After calculating the log average for the input image, the distance from the log average to the maximum value of the envelope signal is computed. By using this distance, the minimum value of the input image is adjusted to have the same distance ratio with the reference image. As shown in the lower image in Fig. 2, MIN is the minimum value with the DR of 60 dB and the newly selected minimum value is MIN' that has the same distance ratio with the reference image. As a result, the input images are able to gain a new DR value, which is determined by the initial maximum value to the adjusted minimum value while having the same ratio with the reference image.

The proposed ADRA method is similar to the principle of the preserving bi-histogram equalization (BBHE) method that is commonly used for contrast enhancement in digital image processing [8]. Generally, histogram equalization (HE) is popular for enhancing the contrast of a given image; however, it is not suitable for consumer electronic products since the HE may change the original brightness of a given image [9]. For this reason, the novel extensions of the HE, such as BBHE, were proposed. In BBHE, an input image histogram is first divided into two sub-histograms based on the mean value of the input image, X_T , as shown in Fig. 3(a). After this separation process, these two sub-histograms are independently equalized as illustrated in Fig. 3(b). By doing so, the mean brightness of the equalized image can be positioned in the middle of the input mean value as well as in the middle gray level [8]. It should be noted that the BBHE has an effect of preserving the mean brightness during the enhancing contrast. Similarly, the proposed method merely divides the input image's envelope signal into two parts based on the log average, and then selects the new minimum value to have the same distance ratio with the reference image rather than equalization. This process will manipulate an input image similar to the reference image by having the comparable envelope signal distribution.

Fig. 4 shows the block diagram to describe how to apply the proposed ADRA method towards 3D ultrasound B-mode imaging. In 3D B-mode imaging, for example, the first frame of the consecutive 2D ultrasound images can be used for a reference image, and the distance ratio is then calculated as described above. After adjusting the DR of the rest of frames to have the same distance ratio with the first frame, the sequence of the processed frames may be used to complete 3D ultrasound image reconstruction. In this fashion, the DR of each frame can be different from each other. However, because the processed frames have the same distance ratio, their brightness and contrast can be consistent, leading to improved image quality in a 3D ultrasound image.

2.2. Experiment set-up and evaluation metrics

To evaluate the performance of the proposed ADRA method for 2D and 3D ultrasonic imaging, two independent *in vivo* experiments were conducted. The first was aimed to evaluate the performance of the ADRA method for 2D ultrasound B-mode ultrasound imaging and the second evaluated 3D ultrasound B-mode imaging. The former experiment used *in vivo* kidney and liver data from six volunteers that were acquired by an experienced radiologist with a C5-2 convex array transducer with a commercial ultrasound scanner (Accuvix V10, Samsung Medison, Korea). The reference images were selected by the radiologist for all the cases based on his clinical experience (e.g. brightness, contrast and image texture) to

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