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Ultrasound Image Enhancement Based on Automatic Time Gain Compensation and Dynamic Range Control

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Abstract

For efficient and accurate diagnosis of ultrasound images, appropriate time gain compensation(TGC) and dynamic range(DR) control of ultrasound echo signals are important. TGC is used for compensating the attenuation of ultrasound echo signals along the depth, and DR controls the image contrast. In recent ultrasound systems, these two factors are automatically set by a system and/or manually adjusted by an operator to obtain the desired image quality on the screen. In this paper, we propose an algorithm to find the optimized parameter values for TGC and DR automatically. In TGC optimization, we determine the degree of attenuation compensation along the depth by dividing an image into vertical strips and reliably estimating the attenuation characteristic of ultrasound signals. For DR optimization, we define a novel cost function by properly using the characteristics of ultrasound images. We obtain experimental results by applying the proposed algorithm to a real ultrasound(US) imaging system. The results verify that the proposed algorithm automatically sets values of TGC and DR in real-time such that the subjective quality of the enhanced ultrasound images may be sufficiently high for efficient and accurate diagnosis.

Key words: ultrasound imaging, time gain compensation, dynamic range

I. INTRODUCTION

The ultrasound imaging system is popularly used in clinical diagnosis because the imaging is non-invasive and can be performed in real-time. However, there exist several disadvantages relative to X-CT(X-ray computed tomography) and MRI(magnetic resonance imaging) images. One drawback is the presence of speckle noise in ultrasound images. The noise induces quality deterioration in the images and has a negative impact on clinical diagnosis. Accordingly, many algorithms have been developed to reduce the speckle noise. Another weakness is that the ultrasound imaging system requires significant interaction with the operator in order to obtain a desired image quality.

In the process of obtaining an ultrasound image, the operator needs to adjust several basic parameters of the system and the image quality is sensitive to their values. Among them, the parameters of time gain compensation (TGC) and dynamic range(DR) control are frequently adjusted by the operator. The TGC is employed to compensate for the attenuation of ultrasound echo signal along the depth and DR control determines the proper range to be displayed from a wide echo signal range. Hence, it is considered meaningful to automatically find the optimized parameter values for TGC and DR in real-time without operator interaction.

In a conventional ultrasound imaging system, an ultrasound image is composed of hundreds of scan lines, and each scan line represents an ultrasound echo signal sampled along the depth direction. The shape of the TGC curve, which controls the amplification applied to the echo signal as a function of depth, can be adjusted by allowing the user to set the gain employed within each of a number of discrete depth zones. Several algorithms have been proposed for compensating the intensity of ultrasound signals along the depth[1-5]. These methods focus on individual compensation of each scan line. Also, some algorithms are considered mainly at the radio frequency signal level, and thereby they can be implemented only through modification of system hardware.

Meanwhile, dynamic range is adjusted in order to determine an intensity range for the 8-bit display from a wide range of a log-compressed echo intensity. The relationship between a log-compressed ultrasound signal and the B/W ultrasound image after DR control is generally modeled as shown in Fig. 1. Here, the range of 8 bits is mainly due to the limitation of the display system. Based on the relationship given in Fig. 1,

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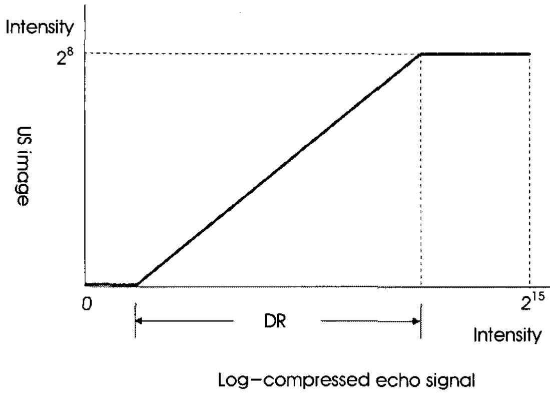


Fig. 1. Relationship between a log-compressed echo signal and the corresponding ultrasound image

we can adjust the image contrast by changing the value of DR. In this case, the initial point of the DR is usually fixed and only the width of the DR is changed. Note here that the image contrast affects the roughness of soft tissue and the edgeness near the tissue boundary, and thereby strongly affects the clinical subjective quality.

Many algorithms for improving the ultrasound image contrast have been developed. Among them, one method attempted to increase the contrast of lesions to the background by analyzing the frequency spectra [6]. However, this method is not appropriate for finding the optimal DR because it deals only with lesions rather than the overall image. A gray-scale image enhancement algorithm has also been proposed [7]. But this method does not find the optimal value of DR because the individual transformation of each pixel value cannot yield global transformation of DR, as shown in Fig. 1.

In this paper, we develop an algorithm to automatically find the optimal parameter values for TGC and DR control in real-time without operator interaction. The proposed algorithm is designed on the basis of a simplified model of an ultrasound imaging system without a loss of generality. Therefore, the algorithm differs from early attempts and can be easily applied to a practical ultrasound imaging system.

II. MATERIALS AND METHODS

The proposed algorithm consists of two procedures, TGC and DR optimization. As an input to the algorithm, we use the log-compressed digital echo data, whose intensities are not compensated along the depth. We perform the optimal TGC based on this input data. After compensating for the intensities of the input data along the depth, DR optimization is performed. Finally, the optimized parameter values for TGC and DR control are applied to the system.

A. TGC Optimization

It is known that ultrasound echoes attenuate exponentially with the propagation distance from the interface. Hence, log-compressed ultrasound echoes are linearly attenuated with the distance in a uniform propagation medium and the degree of attenuation of echoes depends on the medium. If we assume that the attenuation in an ultrasound imaging system occurs due to the one major medium in the body, the attenuation of log-compressed echoes can be modeled as a single straight line. In order to alleviate undesirable errors in estimating the straight line and achieve statistically reliable TGC results, we propose the following method.

Vertical Profiles and Straight Line Modeling

To estimate the attenuation, we need to measure the intensity variation of log-compressed echo signals along the depth. Due to the noise in the signals, however, it is generally difficult to estimate the attenuation. In order to overcome this problem, we use a vertical profile that represents the average of several neighboring echo signals. To obtain the profile, we divide an input echo image into M subsequent vertical strips. If each vertical strip consists of N scan lines, the corr- esponding vertical profile $v_k(n)$ can be written as

$$v_k(n) = \frac{1}{N}\sum_{m=1}^{N} u_k(m,\ n) \tag{1}$$

where $u_k(m,\ n)$ denotes the intensity at location$(m,\ n)$ within the kth strip and the range of k is 1 to M. As mentioned above, the attenuation function of log-compressed echoes in a vertical strip can be modeled as a straight line. This straight line can be estimated from the averaged vertical profile $v_k(n)$. In appro- ximating the vertical profile to a line profile, weadopt the least squares fit technique. Fig. 2. shows a vertical profile and its approximated line profile. We can see in the graph that the approximated line profile represents well the attenuation of the vertical profile along the depth.

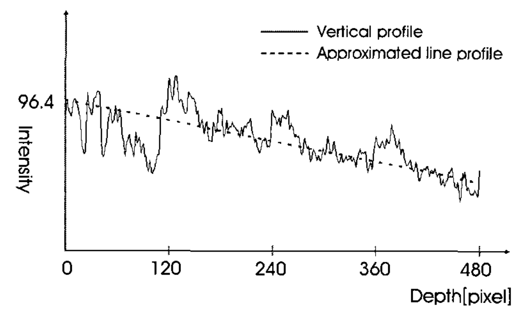


Fig. 2. Average intensity profile of a vertical strip along the depth in an ultrasound image(solid line) and its approximated line profile(dashed line)

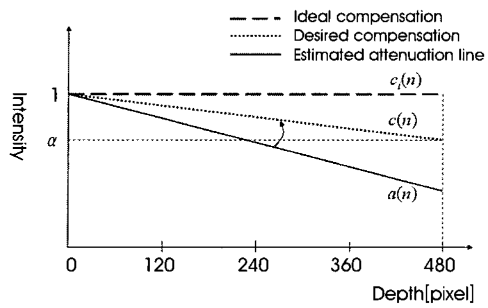


Fig. 3. Estimated attenuation line(solid) and desired attenuation lines(dashed)

Attenuation Compensation

In an ultrasound image, there can be local dark regions such as a vessel or mass in the scanned area. These dark regions may impair correct estimation of the attenuation line, because they tend to make the slope of the approximated line profile steeper. Therefore, to avoid this problem, we exclude the line profiles with steeper slopes in the process of averaging line profiles. The remaining line profiles are then averaged to obtain the attenuation line function for the ultrasound image. Note here that, before averaging, the value at the zero depth of the attenuation line is normalized to one.

Based on the slope of the attenuation line, we perform TGC or compensate for the intensities along the depth. Ideally, the TGC can be done by adjusting the slope of the attenuation line to zero. However, if we simply make the slope zero, the noise will be considerably amplified in the far field area. In a clinical sense, such noise amplification is not desirable. Therefore, in order to prevent the unwanted noise amplification in the far field area, it is necessary to alleviate the degree of TGC. Fig. 3 shows an estimated attenuation line $a(n)$ and two compensated versions of the ideal $c_i(n)$ and the desired $c(n)$. In the figure, a denotes the attenuation factor of $c(n)$ at the deepest point of the image and is used as an experimental parameter to set $c(n)$. Note that a becomes 1 for the ideal compensation of $c_i(n)$.

In practice, TGC can be performed by multiplying the intensities of echoes by a proper gain according to the depth. However, in the ultrasound image, the multiplicative model is changed into an additive model because the input echo signals are log-compressed in the system. Therefore, the image intensity $u_k(m, n)$ can be compensated as follows :

$$u_k^c(m,n) = u_k(m,n) + \beta(c(n) - a(n)) \tag{2}$$

where β denotes the original intensity value at the zero depth of $a(n)$ before normalization. We emphasize that TGC is simply accomplished by addition, as shown in Eq. (2).

B. DR Optimization

The ultrasound imaging system displays an 8-bit B/W image, which is selected from log-compressed echo data with a wide range of intensity by using a given DR. The DR optimization algorithm finds a value of DR that provides the best quality of 8-bit images displayed on the screen. The overall structure of the proposed DR optimization algorithm is given in Fig. 4. The proposed algorithm introduces a cost function as a quantitative measure representing the quality of the displayed image. The algorithm attempts to find the value of DR that minimizes the cost function. The cost function consists of two measures that are obtained on the basis of detected edges in a B/W image. For efficient calculation, we first obtain an edge map for the image in advance. We then calculate the measures only for edge regions described in the edge map.

Edge Detection

To detect edges, we adopt the coherent nonlinear anisotropic diffusion model, which is known to be effective in this regard[7]. In the adopted diffusion model, there is a large difference between the two eigenvalues of a pixel located onan edge. The edge direction is determined as one of the two eigenvectors, which represent the normal and tangential directions. However, for a pixel in a homogeneous region where speckle noise is dominant, there is a small difference between its two eigenvalues. Based on these characteristics of the eigenvectors, we generate an edge map that includes the edgeness and direction at each edge pixel.

Measures

The DR directly affects the boundary contrast and the roughness of soft tissue. Hence, we define two measures as

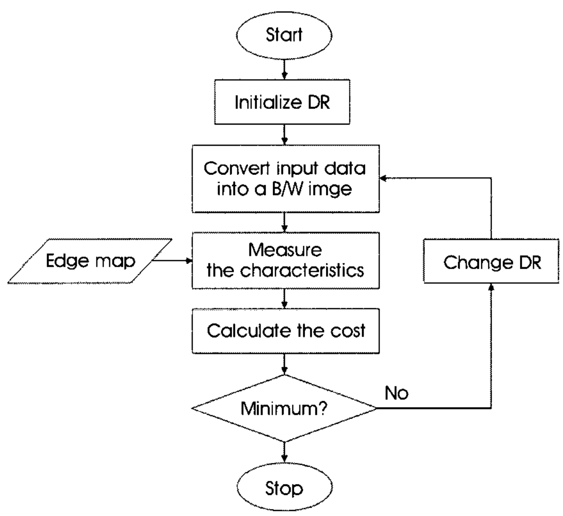


Fig. 4. Proposed DR optimization algorithm

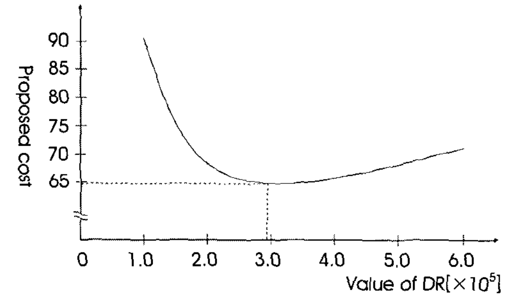


Fig. 5. Cost versus value of DR

global descriptors of the image, namely, edge contrast(EC) and soft tissue roughness(STR), based on the edge map. The edge contrast represents the intensity difference between two regions within a window. These regions are separated by the edge line centered at an edge pixel. The edge contrast can be obtained by the following equation :

$$\mathrm{EC} = \frac{1}{K}\sum_{(i,j)\in E}(|f(i,j)-m_A|+|f(i,j)-m_B|) \qquad (3)$$

where E denotes a set of edge pixels in an ultrasound image and K denotes the total number of edge pixels. m_A and m_B denote the averages of the intensity values of the two regions separated by the edge line in the window, respectively. Also, $f(i, j)$ represents the average of the intensity values of the pixels on the edge line. Here, the edge line passes through an edge pixel (i,j) and is located in a window centered at the edge pixel. This edge line is determined by using the direction vector of the edge pixel, and the intensity value of each point on the edge line is calculated by interpolating its neighboring pixels.

The soft tissue roughness is then defined as a standard deviation of the intensities of non-edge pixels in the image. Since soft tissue regions in the image are considered homogeneous, the soft tissue roughness represents the roughness caused by speckle noise, and can be represented as

$$\mathrm{STR} = \sqrt{\frac{1}{L}\sum_{(i,j)\notin E}\left(u(i,j)-\frac{1}{L}\sum_{(i,j)\notin E}u(i,j)\right)^2} \qquad (4)$$

Here, L denotes the total number of non-edge pixels and $u(i,j)$ denotes the intensity at pixel (i,j).

Cost Function

An ultrasound image with a high EC and a low STR is clinically desirable. By adjusting parameter DR, we attempt to obtain the most desirable image quality. In order to optimize DR, we define a cost function as a weighted sum of STR and the inverse of EC, i.e.,

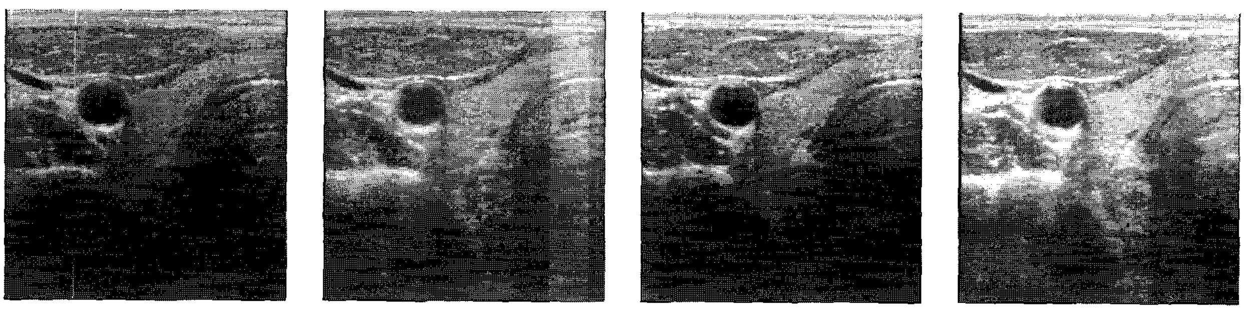
$$J(DR) = \frac{\lambda}{\mathrm{EC}+\varepsilon} + \mathrm{STR}. \qquad (5)$$

Here, ε is a constant of a small value that prevents the denominator of the first term of the right side of the equation from being zero, and λ is a weighting factor that is selected empirically.

The proposed cost J becomes lower as EC increases and/or STR decreases. Therefore, we can expect that it will have the minimum value when DR is optimized. Fig. 5 shows a plot of the cost versus the value of DR. To find a value of DR that minimizes the cost, we adopt the downhill simplex method, which reliably guarantees convergence within a capture range where the global minimum exists.

Ⅲ. RESULTS

To test the performance of the proposed algorithm, we implement the algorithm in a practical ultrasound imaging system, Medison Accuvix XQ, which includes a PC with a CPU of 2.0GHz and 1GB memory. The ultrasound imaging system takes the optimized parameter values of TGC and DR based on the proposed algorithm and displays enhanced images. The processing time to determine the parameter



(a) (b) (c) (d)

Fig. 6. Experimental results for Thyroid : (a) Initial image and the images after (b) TGC optimization, (c) DR optimization, and (d) TGC and DR optimization, respectively.

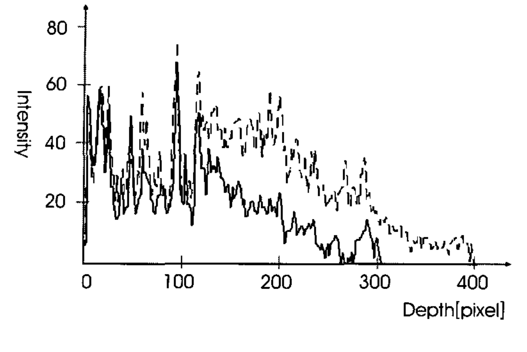


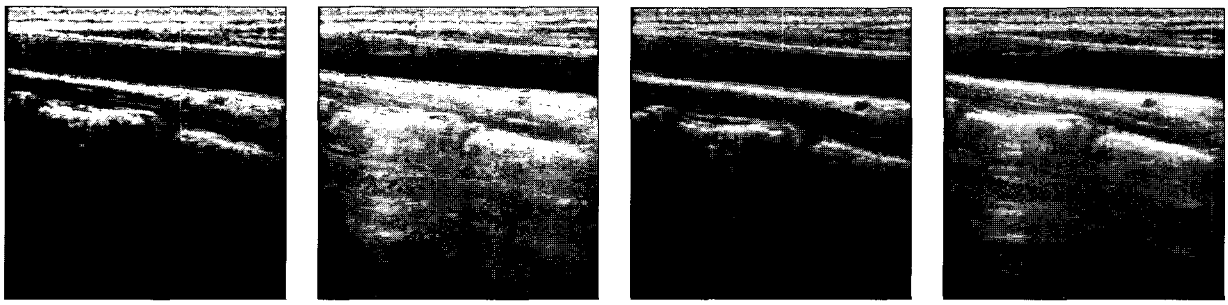
Fig. 7. Intensity profiles of the initial image (solid) and TGC optimized image (dashed)

values depends on the characteristics of the input image, but is approximately in a range of 100ms to 300ms without software code optimization. Since the optimization of TGC and DR is performed once during the patient scanning, the proposed algorithm can be considered fast enough for real-time processing. The variation of the processing time is mainly due to the number of iterations to minimize the cost in DR optimization. In the experiments, values of a and λ are determined on the basis of the opinions of clinicians. For thyroid and carotid, which are most frequently scanned in clinical diagnosis, a and λ are set to 0.95 and 1450, respectively. In the case of liver, a is set to 1.0 and λ is set to 1400.

Fig. 6 shows experimental results for the thyroid. Fig. 6(a) is an initial image in which the time gain is not compensated along the depth and the parameter of DR isnot properly adjusted. The result after TGC optimization by using the input echo signal is given in Fig. 6(b). Compared to Fig. 6(a), Fig. 6(b) shows a more homogeneous image, because it is compensated along the depth. Fig. 7 illustrates each intensity profile at the center of width in Figs. 6(a) and (b). The image after DR optimization is shown in Fig. 6(c). Examination of this image verifies that the contrast of the input image is improved and the structures are better visualized. Finally, the image after both TGC and DR optimization is shown in Fig. 6(d). As expected, the image shows superior quality over the other images.

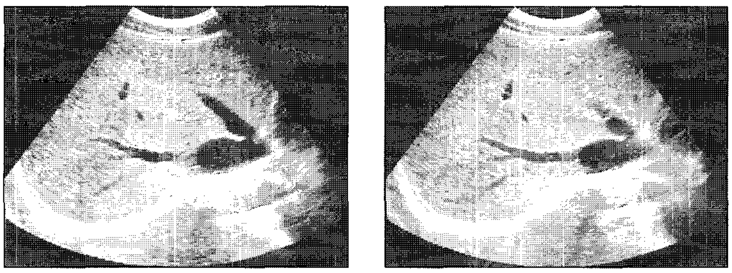
As noted above, the system first obtains an image, as shown in Fig. 6(a), by scanning a real body and determines the optimal parameter values of TGC and DR by using the proposed algorithm. The system then enhances the following images based on the obtained parameter values, as shown in Figs. 6(b)-(d). Hence, note that there exists a small scanning time interval between the initial image given in Fig. 6(a), from which the parameter values are determined, and the enhanced images given in Figs. 6(b)-(d).

Experimental results for the carotid are given in Fig. 8. We can see in an initial image given in Fig. 8(a) thatthe image



(a) (b) (c) (d)

Fig. 8. Experimental results for Carotid: (a) Initial image and the images after (b) TGC optimization, (c) DR optimization, and (d) TGC and DR optimization, respectively.



(a) (b)

Fig. 9. Experimental results for Liver: (a) Initial image and (b) the improved image after TGC and DR optimization.

intensity rapidly attenuates along the depth and the image appears rough due to high contrast. Meanwhile, Fig. 8(b) shows the image after TGC optimization. In this image, the intensity is compensated but the contrast is still too high. In the image after DR optimization, however, the contrast is adequate and the image appears smoother, as shown in Fig. 8(c). The best result, shown in Fig. 8(d), is obtained after both TGC and DR optimization.

Fig. 9 shows ultrasound images for a liver section. In this figure, a vertical column of pixels in the image does not represent the true vertical direction because the image is obtained using a convex probe and scan-converted. To apply the proposed algorithm, we use the image before scan conversion. Fig. 9(a) shows the initial image. In contrast to the initial images in the previous experiments, the initial image is already somewhat adjusted for TGC and DR. Fig. 9(b) demonstrates that the value of the TGC parameter hardlychanges and the contrast is improved slightly. Thereby, the overall image quality is maintained or slightly improved without introducing any deterioration. From this experiment, it is determined that the proposed algorithm is robust even for an initial image that has already been enhanced.

Ⅳ. SUMMARY

In this paper, we present an algorithm to optimize the TGC and DR parameters, which strongly affect the displayed image quality in ultrasound imaging systems. In the proposed algorithm, in order to find the optimized parameters for TGC, we divide the ultrasound image into consecutive vertical strips and obtain vertical profiles from the strips. We then approximate every vertical profile to a straight line, and estimate the degree of attenuation along the depth by averaging the straight lines. The parameter values for TGC are obtained by using the estimated degree of attenuation. For DR optimization, we define two measures of edge contrast and soft tissue roughness, which are functions of DR. For the optimal valueof DR, the corresponding image is expected to have high edge contrast and low soft tissue roughness. Hence, we define the cost function as a weighted sum of the soft tissue roughness and the inverse of edge contrast, which has a minimum value when DR is optimized. This cost function enables robust DR optimization. By applying the proposed algorithm to a practical ultrasound system, we demonstrated that it is possible to automatically optimize the parameters for TGC and DR control. We verify that the subject quality of the image with the optimized parameters is clearly improved for clinical diagnosis.

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