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Adaptive Gain Control And Contrast Improvement For Medical Diagnostic Ultrasound B-mode Imaging System Using Charge-couples Devices

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ADAPTIVE GAIN CONTROL AND CONTRAST IMPROVEMENT FOR MEDICAL DIAGNOSTIC ULTRASOUND B-MODE IMAGING SYSTEM USING CHARGE-COUPLES DEVICES

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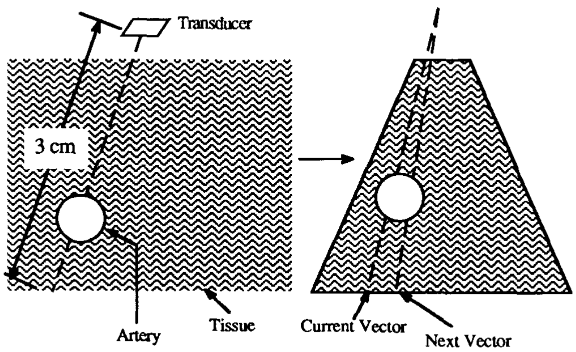
ABSTRACT

A B-mode ultrasound imaging system operates by transmitting a high-frequency sound pulse and measuring the strength and time of flight of the reflected signal. The ultrasound signal may be attenuated as much as 60 dB in its round trip to a target and must be normalized so that contrast is not lost in the far field. Manual controls for adjusting receiver gain versus depth are common in many clinical uses but are impractical in most surgical applications. The goal of this project was to design a circuit to automatically adjust the gain. A two-dimensional low-pass filter is used to approximate the returned signal's local average value which is a measure of the total attenuation, scattering and beam spreading experienced by the ultrasound energy. The attenuation is compensated by dividing the returned signal by its average value to remove the attenuation.

INTRODUCTION

The purpose of this project is to compensate the attenuation experienced by ultrasound energy in tissue during medical imaging procedures. A scanning radial sector ultrasound probe uses a pulse-echo technique to generate an image as shown in Figure 1.



Imaging in Tissue CRT Display

Figure 1. B-Mode Transducer Operation

A high-frequency acoustical signal is generated by a high voltage pulse to the transducer which is positioned on the tissue to be imaged. As the acoustical energy propagates in the tissue partial reflection of the ultrasound signal occurs at the interfaces of various materials. Reflectivity is a function of the difference between the acoustical impedances of the materials. Acoustical impedance is a function of velocity of propagation and density. In general, human tissue is a good transmission medium for ultrasound energy. Materials such as bone and metal are strong reflectors, while blood is a weak reflector.[1]

Between pulses, the transducer functions as a receiver for the echoes. Ultrasound energy is converted to electrical energy by the transducer, and the magnitude of the transducer output modulates intensities on the CRT. Since the brightness of the CRT display is modulated by the intensity of the returned echo, this method is termed B-mode operation.

The returned signal from a single pulse is amplified and peak detected. A 6-bit analog to digital converter (ADC) samples this signal and writes the sample values in one row of video RAM for display on a CRT. When the transducer moves to the next valid position, the pulse-echo (or transmit/receive) procedure repeats to form the next vector of the image.[2]

A target's depth is determined by the signal's propagation time; an average velocity of 1.5mm/us is assumed in body tissue. Typical imaging depths are 3 cm to 9 cm. Distance from the transducer to a reflector corresponds to radial distance on the CRT display.

At the frequencies used for medical imaging, typically 3 MHz to 10 MHz, the ultrasound signal may be attenuated as much as 60 dB in its round-trip to a target. (A typical rule of thumb is to approximate 1 dB/cm/MHz attenuation in body tissue.) As much as 40 dB of additional signal variation is due to the size and reflectivity of targets.[3] A logarithmic amplifier (log amp) is used to compress the possible 100 dB signal variation so that it can be displayed in the luminance range of a typical CRT which is approximately 20 dB to 30 dB. This compression can greatly diminish contrast in the far field; therefore the attenuation must be compensated so that deep targets appear with the same contrast as shallow ones. A static time-gain function cannot be used to compensate the attenuation because different tissues have different attenuation characteristics. Brain tissue is more attenuating than abdominal tissue, for example.

In many clinical ultrasound imaging systems the time-gain function is typically implemented by decomposing the image into range segments and providing controls for adjusting the gain applied to each segment. However, rapidly changing imaging conditions make manual controls impractical for use during surgery. As a parallel, consider the manual operation of a 35 mm camera compared to automated operation of aperture and shutter functions. Certainly, under static conditions a skilled photographer can obtain superior photographs with manual controls. However, with a moving subject and varying light field, even an amateur can obtain excellent results with an automatic camera. The ultrasound system described here is intended for this latter type of operation--varying image field, varying signal strengths, and an unskilled operator as would be the situation for an ultrasound system intended for use in the surgical suite.[3]

ATTENUATION COMPENSATION METHOD

Figure 2 is a sketch of a typical voltage output of the transducer after being compressed by a logarithmic amplifier and then peak detected. This signal would form one vector of the image. It consists of high-frequency local variation representing scene detail. An average of the high-frequency signal over a short period of time, that is the local average, forms a slowly varying signal. Due to attenuation, the low-frequency signal decreases with time (or range) and so does the peak-to-peak magnitude of the higher-frequency local variation. A large local change such as the dip at 30 us would be due to a target such as the blood vessel depicted in Figure 1. When this signal is presented to the ADC input, the local variation covers a large number of quantizing levels in the near field but only a small number in the far field, and so, contrast in the far field is lower. Additionally, the signal in the far field is at a lower level than the near field, so the near field will appear much brighter on the CRT. If the intensity variation due to attenuation can be reduced, the local contrast can be enhanced to make better use of the ADC quantizing levels.

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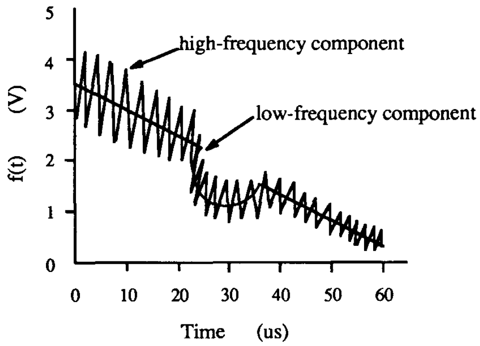

Figure 2. Peak-Detected Transducer Output

The basis of the system described here is that the local average of the detected signal is a good measure of the attenuation that the ultrasound signal has experienced. A returned signal for one vector is a function of time:

$$f(t) = a(t)\, v(t)$$

where v(t) is the signal containing scene information if there were no reduction due to attenuation and $a(t)$ is an attenuating function of the form $\exp(-2Bt)$. Once the local average has been estimated, the attenuating function $a(t)$ can be easily removed:

$$f_2(t) = \frac{f(t)}{\hat{\mu}(t)} = \frac{a(t)v(t)}{\hat{\mu}(t)} \approx \frac{a(t)v(t)}{\hat{a}(t)} \approx v(t)$$

where $\hat{\mu}(t)$ is the estimated local average. Then a uniform gain and offset can be applied to the signal in order to bring the signal levels within the range of the ADC input.

Figure 3 is a graph of the desired result of this processing. Here it can be seen that the magnitude of the high-frequency variation has been boosted in the far field and that the signal level no longer falls off with time. This will enhance contrast in the far field by making more effective use of the ADC quantizing levels.

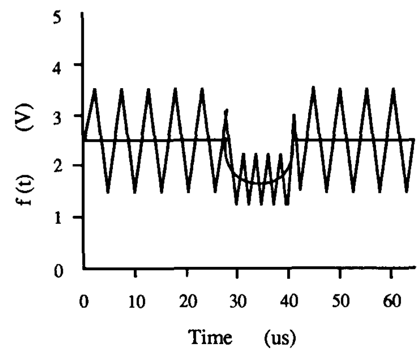

Figure 3. Peak-Detected Transducer Output After Processing

However, finding the local average for only one vector as in this one-dimensional example will not always give a good estimate of the attenuation function. For the vector shown in Figure 2, for example, the local average will follow the big dip at 30 us created by the target. Averaging the local average across several vectors will reduce this problem. This two-dimensional averaging is depicted in Figure 4. Here, the gain applied at time t will be a function of the samples within a window of N percent of the vector length along the vector and M previous vectors. This effectively creates a sliding window where the local average is calculated for the N x M area.

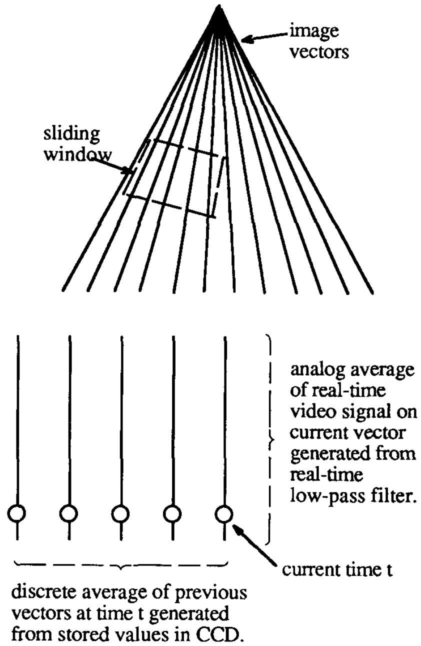

Figure 4. Two-Dimensional Local Average

The alternate method considered for compensating the attenuation involved creating several non-sliding windows at increasing depths in the image. The local mean would be calculated in each region using a digital signal processor, and this information would be used to control the gain in that region. Interpolation of the gain function across region boundaries would be required to eliminate harsh intensity changes across the boundaries. This approach proved effective in computer simulation, but because of the high sample rate of the ultrasound imaging system (4 MHz), the hardware required for a real-time implementation of this scheme would be unnecessarily complex and expensive.

Since the local average value is the DC component of the signal, it can be isolated with low-pass filtering.[4] A two dimensional filter is required to implement the sliding window described to estimate the average over a local area. The size of the window will be controlled by the cutoff of the filter. A two-dimensional, separable, first-order filter has a frequency response given by the product of two one-dimensional filter responses:

$$\left|H(f_x, f_y)\right| = \left|\frac{V_\mu(f_x, f_y)}{V_{in}(f_x, f_y)}\right| = \frac{1}{\sqrt{1+(f_x/f_{cx})^2}}\,\frac{1}{\sqrt{1+(f_y/f_{cy})^2}}$$

Giving:

$$H(s_1, s_2) = \frac{V_\mu}{V_{in}} = \frac{\omega_{cx}}{s_1 + \omega_{cx}}\,\frac{\omega_{cy}}{s_2 + \omega_{cy}}$$

The equivalent sampled data filter has a two-dimensional Z transform[5]:

$$H(z_1, z_2) = \frac{V_\mu(z_1, z_2)}{V_{in}(z_1, z_2)} = \frac{g_1}{1 - e^{-g_1} z_1^{-1}}\,\frac{g_2}{1 - e^{-g_2} z_2^{-1}}$$

Since g_1 and g_2 control the cutoff frequencies of the filter, they control the effective size of the local area over which averaging is

done.[4] By defining a new intermediate variable, $V\mu'$, the two-dimensional filter can be broken up into two one-dimensional filters in cascade:

$$H_1(z_1,z_2) = \frac{V_{\mu'}(z_1,z_2)}{V_{in}(z_1,z_2)} = \frac{g_1}{1 - e^{-g_1} z_1^{-1}}$$

$$H_2(z_1,z_2) = \frac{V_{\mu}(z_1,z_2)}{V_{\mu'}(z_1,z_2)} = \frac{g_2}{1 - e^{-g_2} z_2^{-1}}$$

Using the first filter along the scan direction eliminates the need for it to have a recursive architecture since it is in real-time. Therefore, the first low-pass filter can be implemented with a simple passive circuit. The second filter must be recursive since it must average information from the current vector with information from the previous ones, requiring that the previous vectors be stored. Figure 5 is a block diagram of this two-dimensional filter.

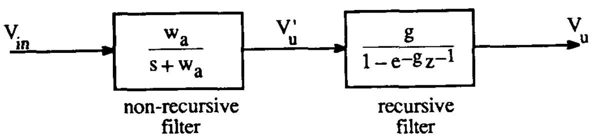

Figure 5. Two-Dimensional Low-Pass Filter

PASSIVE LOW PASS FILTER

The first filter is required to return the average over a length of time which is short relative to the length of the vector. For a typical image depth of about 37 mm, the returned signal would be 2 x 37mm/1.5mm/us = 49.3us. To average over 5% of this signal would require a time constant of about 2.5us. By experiment, a value of 2.2us was found to give the best result, and the second-order system in Figure 7 offered a response which more closely approximated the local average shown in Figure 5.

$$H(s) = \frac{V_{\mu'}}{V_{in}} = \frac{\frac{1}{LC}}{s^2 + s\frac{R}{L} + \frac{1}{LC}} = \frac{\omega_c^2}{s^2 + 2\omega_c\, d\, s + \omega_c^2}$$

The pole frequency was $f_c = 1/(2 * \pi * 2.2\mu s) = 72.3$ kHz, and the damping factor was selected to be d = .707 to avoid gain greater than one near the pole frequency.

The CCD selected for this circuit was the CCD321B-3 video delay line from Fairchild Weston which is rated for operation at video frequencies and for start/stop mode operation. This delay line consists of two independent 455 bit analog shift registers, which are operated in multiplexed mode in this circuit, providing 910 bits of analog delay at a sample rate double the clock rate. Figure 6 shows the complete AGC system.

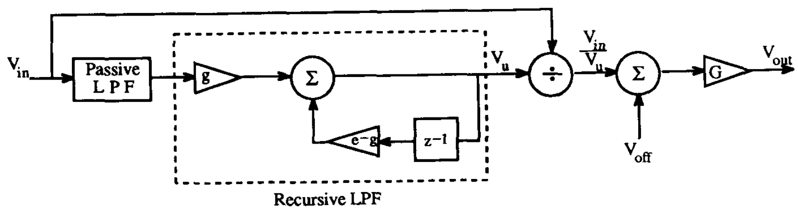

Figure 6. Block Diagram of AGC System

RECURSIVE LOW-PASS FILTER

The recursive low-pass filter is required to have the transfer function

$$H(z) = \frac{V_\mu}{V_{\mu'}} = \frac{g}{1 - e^{-g} z^{-1}}$$

and so must implement the function

$$V_\mu(M) = g V_{\mu'}(M) + e^{-g} V_\mu(M-1)$$

where g controls the cutoff of the low-pass filter and M is the current vector number.

By experiment, it was found that $0.2 < g < 0.3$ gives good results. With exp(-g) of the current vector, up to 7 of the previous vectors will have a significant effect on the current value of the local average. The output changes rapidly enough to approximate the attenuating function but sufficiently removes variations due to large targets. Larger values of g do not remove enough of the variation due to targets and therefore boost the gain too much in the center of a large dark target. Smaller values of g make the average too flat to approximate the attenuating function, and therefore, dividing by the average does not provide sufficient gain in the far field.

The single vector delay is implemented with a charge coupled device (CCD) delay line, which operates by storing minority carriers in a depletion region at the surface of a semiconductor. The size of the depletion region is defined by clock potentials. Charge is moved about by moving the potential well in which the charges are stored. Charge can be injected at one end of the device, moved along it, and detected at the output.

CONCLUSION

The attenuation experienced by ultrasound energy is severe at frequencies useful for medical imaging. The average value of the peak detected signal returned from an ultrasound transducer is a good approximation of the attenuating function. This average value can be found easily by low-pass filtering the peak detected signal. By dividing the peak detected signal by its average value, the attenuation is removed, and contrast is improved in the far-field. Obtaining the average in a local area, instead of just averaging along a vector, is necessary in order to remove large local average swings due to large targets. Merely averaging across vectors was not effective because not enough of the high-frequency scene detail could be removed by the recursive averaging filter alone.

BIBLIOGRAPHY

[1] Herrick, Tom and Alan Bryant. "Automated Gain Control for Medical Diagnostic Ultrasound Imaging." Paper presented at the Midwest Circuits and Systems Symposium, August, 1990, Calgary, Alberta, Canada.

[2] Richard, William David. "A Microcomputer Architecture for Opthalmic Ultrasonography." Diss. UMR, 1988.

[3] Herrick, Tom. "Implementation of Real-Time Automatic Gain Control for Ultrasound Imaging Systems." Paper presented at Computers and Ultrasound Imaging Symposium, Drexel University, Philadelphia, PA, 1986.

[4] Narendra, Patrenahalli M. and Robert C. Fitch. "Real-Time Adaptive Contrast Enhancement." IEEE Transactions on Pattern Analysis and Machine Intellegence PAMI-3 (1981): 655-661.