Time gain compensation in orbital ultrasonography

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Abstract
Time gain compensation (TGC) is a setting applied in diagnostic ultrasound imaging to account for tissue attenuation. By increasing the received signal intensity with depth, the artifacts in the uniformity of a B-mode image intensity are reduced. The purpose of TGC is to normalize the signal amplitude with time, compensating for depth.
When the image is displayed, similar materials should have similar brightness, regardless of depth; this is achieved by "Linear-in-dB" Gain, which means the decibel gain is a linear function of the control voltage. Gain is expressed in dB, a logarithmic ratio of the output power relative to the input power. Gain can be calculated by subtracting the input from the output levels when both are expressed in dBm, which is power relative to 1 milliwatt.
The TGC creates uniformity in the brightness of the echoes when used in conjunction with the overall gain. The best approach is to center all the TGC settings before adjusting the overall gain. After adjusting the overall gain, the TGC can then be adjusted to compensate for attenuation at specific depth. Gain is a uniform amplification of the ultrasonic signal that returns to the transducer after it travels through the tissue.

Keywords: Orbital Ultrasound, Time gain compensation (TGC)

Introduction
Ultrasonography is an essential method for medical imaging. Compared with other imaging techniques, ultrasound is inexpensive, harmless, and enables easy real-time imaging for medical diagnosis [1].
Ultrasound produces multi-perspective images with high resolution, allowing the assessment of the morphology, orientation, internal structure, and margins of lesions, specifically in dense glandular structures and predominantly fatty tissue [2]. The amplitude and intensity of ultrasound waves decrease as they travel through tissue, a phenomenon known as attenuation [3], and the degree of ultrasound attenuation depends on the type of the tissue and ultrasound wave [4]. The attenuation coefficients vary among different tissues, and the attenuation is affected by the frequency. The higher the frequency, the greater the attenuation [5].
Ultrasound attenuation occurs during the forward propagation of the body and the return of sound waves. Therefore, the pulse-echo intensity of deep tissue is significantly lower than its initial intensity [6]. A way to overcome ultrasound attenuation is time gain compensation (TGC), in which signal gain is increased as time passes from the emitted wave pulse [7].
This correction makes equally echogenic tissues have a similar appearance despite originating from different. So, rather than brightening the monitor, the image on the screen is whitened by a uniform margin, as though the returning signal is stronger than it is, to make it easier to see. The depth dimension of the image is divided into horizontal (linear format) or radial (sector format or curved linear array format) strips, each of which is connected to a separate amplifier stage with a variable gain. These gain controls can be adjusted manually to boost the gain independently in each strip zone. The net effect of these gains is that they provide a means to increase gain with depth in a stepwise manner in order to offset the effects of absorption.

Adjustments in time gain compensation (TGC) are made to approximate a uniform background level throughout the field of view. TGC sliders are used to adjust the gain in specific areas of the image (near-, mid-, and far-field). Most sonographers adjust these to left-of-center for the near field (top), and slowly move to right-of-center as image quality decreases deeper in the image [8]. The attenuation coefficients of different tissues can be assessed automatically, and adaptive TGC can be achieved by means of software. The key of this algorithm is the automatic estimation of the attenuation coefficient. The methods of extracting the attenuation coefficient mainly include time domain and frequency domain approaches [9].

Methodology

The fundamental principle of ultrasound imaging is reflection of ultrasound waves from tissue in the path of the beam. Piezoelectric crystals in the transducer of the scan head produce pulses of ultrasound with frequencies >2 MHz used for ultrasonic imaging. The ultrasound beam is transmitted through the skin. When the beam comes in contact with a tissue interface (e.g., skin-subcutaneous fat, fat-muscle, and muscle-bone), it is partially reflected back to the transducer as an echo signal. Thus, the transducer has a dual function of transmitting the ultrasound and receiving it. The echoes are converted into signals for processing by the transducer. The strength of each reflected wave is represented by a dot, and the position of the dot represents the depth from which the echo was received. The dots are combined to form an image [10].

The amount of sound reflected is dependent on the changes in acoustic impedance between two tissue interfaces. Acoustic impedance is the product of tissue density and acoustic velocity [11]. Air has almost no impedance, while fat and muscle have impedances of 0.138 $g \cdot cm^{-1} \cdot s^{-1}$ and 0.170 $g \cdot cm^{-1} \cdot s^{-1}$, respectively, and bone has a relatively high impedance of 0.78 $g \cdot cm^{-1} \cdot s^{-1}$. Homogenous zones with relatively uniform acoustic impedance are free of echoes. Because the acoustic impedances of fat and muscle are similar, there is a weaker echo for the fat-muscle interface than for the muscle-bone interface. For example, the software for a relatively new portable ultrasound that converts ultrasound images to body fat percentages (Body View software, Intel Metrix, Inc., Livermore, CA) assumes an acoustic reflection coefficient of 0.012 for the fat-muscle interface, but 0.22 for the muscle-bone interface [12].

The procedure for ultrasound scanning is simple. Gel is placed on the head of the transducer and/or the skin at the site to be measured. This creates a close bond between the transducer and skin reducing artifact and making it easier to move the transducer over the skin. With the ultrasound on, the transducer is slid across the measurement site without loss of contact with the skin. A scan takes only a few seconds. Once scanned, the image on the monitor can be saved for analysis.

Tissue thickness measurement is accomplished with electronic calipers. Identification and placement of the two caliper points defining the boundaries of the tissue to be measured requires practice to improve the objectivity of the measurement [13].

A basic orbital US screens the orbital fat, evaluates and measures the extraocular muscles, and assesses the optic nerves [14]. Targeted exams include topographic echography in which a mass may be isolated on the B-scan and then its dimensions were measured using the A-scan, 30° off sagittal plane tests to evaluate the subarachnoid optic nerve fluid versus inflammation, quantitative evaluation in which the A-scan uses sonar reflectivity to evaluate a lesion's tissue properties, and kinetic echography during which the physical pressure of the probe is used to characterize the compressibility of a lesion in question [14-16]. As a result, the Trans ocular and orbital use of A-scan, B-scan, and Doppler US has persevered as a fast, inexpensive, and valuable tool in the evaluation of orbital pathologies [17-18]. Tissue changes in orbital pathologies are irregular and high internal reflectivity in contrast to a low reflectivity in the rare disorder of acute inflammatory orbital myositis. However, the muscle reflectivity has never been related to the stage of the disease, and we postulated that a high reflectivity might be typical for inactive disease caused by the acoustic interfaces produced by fibrosis within the muscle, whereas lymphocytic infiltration and edema would result in low reflectivity (i.e., low echogenicity, as in acute orbital myositis). The orbital fat volume can be measured on fat images. (Figure 1).

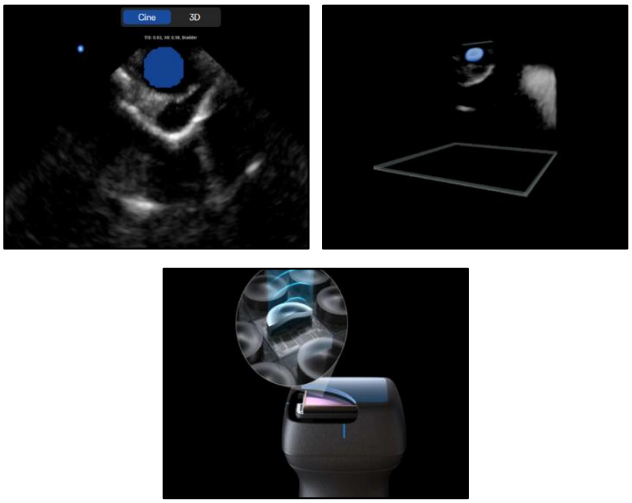

Fig 1: Large Intra-Orbital Abscess with delineation and 3D configuration using TGC extended to extraconal space using Butterfly IQ+ Handheld Ultrasound.

Orbital ultrasonography instructional diagram (Created with BioRender.com) (19-21)

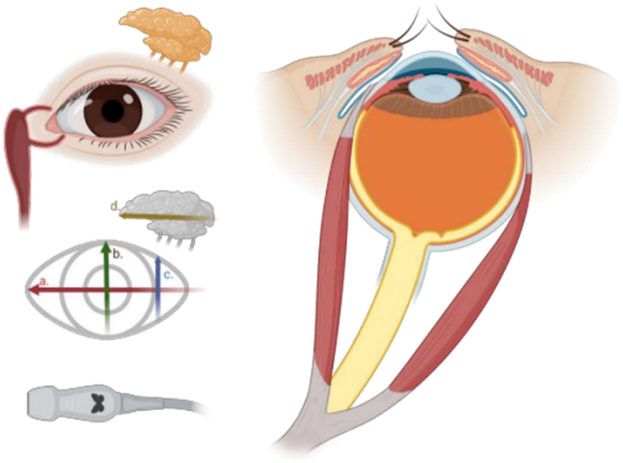

Fig 2

Standard operating procedure for orbital ultrasonography:

1. Obtain approved consent and authorization form

2. Clean and prepare the site of interest and ultrasound probe
3. Apply sterile coupling agents on ultrasound probe headpiece
4. Connect the probe to mobile device and launch the app
5. Select the appropriate preset to start scanning
    a. Horizontal linear scan (medial orientation) - adjust depth and ΔTGC
    b. Vertical linear scan (superior orientation) - adjust depth and ΔTGC
    c. Lateral vertical oblique scan (Ossoinig technique + doppler) - optic nerve assessment
    d. Horizontal linear (Lacrimal gland) scan - volumetric and ΔTGC
6. 6) 3D scan/Cine recording of orbit

Discussion

TGC is an approach available to imaging system users to manually adjust for the changes in echo-amplitude caused by variations in beam-formation along the beam axis and by absorption. Better image improvements can be obtained by analyzing the video data and adaptively remapping the gain in an image in a 2D sense. At least two different approaches have appeared in literature (Melton and Skorton, 1981); Hughes and Duck, 1997). The first method senses differences in RF backscatter and adaptively changes TGC gains. The second analyzes each line of video data to read just the intensity levels as a function of time, based on an algorithm, and it leads to an image renormalized at each spatial point. This last approach is more suitable for imaging systems because it can be accomplished in software without major hardware changes.

Ultrasound energy is progressively attenuated as it travels through tissue. Therefore, signals returning from reflectors at a depth are weaker in strength. By selectively amplifying the echoes from greater depths, using the TGC method or depth-gain compensation (DGC), equal reflectors at unequal depths are displayed as structures of equal brightness on the monitor. TGC is preset to a large degree, and the operator can make fine adjustments if necessary. The TGC control is presented as a series of sliders arranged in a vertical fashion on the control panel. Each of the sliders adjusts the amplification of the returning US signals at a specific image depth.

After TGC of the signal has occurred, additional signal processing is performed. The algorithms for signal processing performed differ among ultrasound processors and are closely held proprietary information. In general, some form of demodulation of the radiofrequency (RF) signal is performed to obtain an envelope of the RF signal, which is used to produce a B-mode image. In addition, processing can include threshold suppression to eliminate signals that are below an operator-specified threshold. Leading edge detection, peak detection, and differentiation are additional methods that can be used by processors to improve image quality [22].

Conclusions

In clinical application, non-invasive ultrasound detection does not rely on the operator's experience and gives patients a certain degree of comfort and safety. Additionally, the adaptive TGC technology realized by software can reduce the complexity of hardware and effectively compensate the attenuation signal, which lays a foundation for real-time diagnosis and is beneficial to the development of portable medical equipment. The process and its output are conducive to the accurate classification of the state of the mass and the differentiation between benign and malignant tissues in clinical applications, improving specificity and reducing biopsy rates.

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