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Computers in Ultrasonic Imaging

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This article describes the role of computers and digital electronics in state-of-the-art diagnostic ultrasound scanners. An overview of the computational requirements is provided, and limits on color flow image frame rates are discussed. The new scanner architectures emerging may be used to extend current limitations of ultrasonography, making features such as automatic phase aberration correction, speckle reduction, and tissue characterization available.

KEY WORDS: color flow, computers, Doppler, gray scale, ultrasound.

ULTRASONIC IMAGING has become an essential tool in diagnostic radiology. Major advances in hardware design over the past four decades have taken ultrasonic imaging from A-mode (amplitude) displays on an oscilloscope to the high resolution gray scale and color flow images common today. The use of computers, or more generally, digital electronics, is one of these advances, and it continues to impact ultrasonic imaging.

State-of-the-art ultrasound scanners present a large range of computational tasks. One of the simplest tasks performed by digital electronics is the generation of text and graphics, such as patient information and color bars, for video display. These tasks are a trivial computational load. More "computer power" is associated with the functions of echo signal data acquisition and processing. Some tasks, such as signal processing for flow information or beam forming for array transducers can be very computationally intensive, if performed in software.

Ultrasound scanner manufacturers have produced imaging systems that acquire and analyze ultrasonic data, and display the resulting images at real-time image rates. This article will identify some of the major computational requirements in ultrasonic imaging and how these requirements are met. Some of the research applications of computers in ultrasonic imaging are also presented.

COMPUTATIONAL TASKS AND REQUIREMENTS

In the simplest case of real-time ultrasonic imaging, data are acquired using a mechanical sector scanner, which is a fixed-focus transducer mechanically steered by the imaging system to produce the familiar pie-shaped sector image. At each orientation of the transducer, an acoustic pulse is launched and echoes are detected. Under scanner control, the piezoelectric element of the transducer is repositioned to produce pulse-echo data along consecutive lines of sight into the body. Data from each line of sight, or A-line, must be amplified, corrected for depth-dependent signal losses, filtered, envelope detected, logarithmically compressed, converted from analog to digital data, and stored in random access memory, though not necessarily in that order. These functions are performed in what is commonly referred to as the scanner front end. Data from the front end must be interpolated between A-lines, transformed to be compatible with video timing, and encoded into gray levels; this is the function of the scan converter. All of this signal processing must be performed at real-time rates (30 frames per second) for display.

The use of array transducers adds significantly to the complexity of this process. Consider, for example, a conventional phased array system. As many as 256 array elements are

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excited to produce the transmitted pulse, which is the coherent sum of pulses from each element. The excitation time for array elements is varied to focus and steer the transmitted pulse. The acoustic beam can be focused at multiple transmit zones by using several pulses, one for each focus. Following each pulse, a sufficient time must elapse to detect all echo signals in the desired depth of field. Another transmit beam is then created at a different depth by again exciting the array elements with slightly different timing. The echo signals detected by each element are delayed and summed in a similar manner to again vary the focal properties and steer the array, this time on receive. These techniques require further image processing to smooth the banding in the image gray levels that can occur with multiple foci. All of these signal delays and sums, with the associated image processing, must occur at "real-time" frame rates.

Color flow imaging is another area that requires significant signal and image processing. A color flow image maps the encoded Doppler signals that indicate flow (using color overlays) onto a gray scale image of the surrounding tissue anatomy. Autocorrelation techniques are commonly used to estimate flow velocities for color flow imaging. With this approach, the echoes from one location in the body from one transmitted pulse are compared with that from successive pulses and the phase differences are analyzed. The magnitude of the phase difference for echoes from a given location is proportional to the flow velocity at that location. Noise generated by system electronics and by moving structures is a significant problem in color flow imaging. To minimize color image noise, color flow imaging uses at least three pulses to estimate the flow velocity for each pixel. The short pulse durations used for high spatial resolution gray scale imaging are not well suited for color flow imaging. More accurate estimates of average flow velocity are obtained by using narrow band pulses that are several cycles long. To allow simultaneous data acquisition for gray scale and color images, the long duration pulses for flow analysis are interleaved with the short duration pulses for high resolution gray scale.

The summary above illustrates the magnitude of computational tasks currently involved in state-of-the-art ultrasonic imaging. Whereas the exact solutions implemented by different manufacturers vary, the general approach to this real-time signal processing problem is outlined below.

SYSTEM AND COMPUTER ARCHITECTURE

It would seem that the computational requirements of a real-time ultrasound scanner would prohibit signal processing on a standard (serial) computer architecture. However, the central processing units (CPU) found in current ultrasound scanners are the same as those found in small personal computers.

This raises two questions regarding computational requirements. First, how can signal processing and image formation be performed at real-time imaging rates by a relatively small microprocessor? This problem has been addressed in several different ways. First, a distributed microprocessor architecture relegates independent tasks to different processors to reduce the computational requirements of any single processor. System functions are divided into independent subsystems that can run concurrently on separate processors. Subsystems are interconnected and share common signals via signal busses, just as the backplane in a computer interconnects the CPU, memory, disk controllers, etc. The subsystems are arranged with the system CPU acting as a master with several slave CPUs in the subsystems. Independent tasks, such as the signal processing needed to generate gray scale and color flow images, are performed concurrently on independent circuits. Second, some computationally intensive tasks are handled by digital signal processing (DSP) chips that perform specific tasks very quickly. For example, the Hilbert transform used for calculating echo signal envelopes for gray scale imaging and the autocorrelation process used in color flow imaging are performed with DSP chips. Third, ultrasound scanners appropriately mix the high accuracy of digital signal processing with the high speed of analog signal processing. Compared to digital signal processing, analog circuits have the advantage of higher processing speeds, but the disadvantages of higher sensitivity to noise and lower stability.

In some cases, such as analog delay lines for beam forming, the use of analog circuits presents another drawback. The circuit that performs the analog delay also acts as a low pass filter and limits the available bandwidth of the processed echo signals. The reduction in system bandwidth degrades spatial resolution in the image. A recent improvement in VLSI (very large scale integration) circuitry has led to application specific integrated circuits (ASICs) that place very specific computationally intensive tasks on a dedicated chip. For example, ATL (Advanced Technologies Laboratories, Bothel, WA) has recently implemented an ASIC design for the delay lines used in beam forming. The reported result[1] is a reduction in the number of circuit elements required to perform specific functions and, because digital delays have comparatively high bandpass, an increase in the available system bandwidth. These features increase the system reliability and decrease the power requirements, size, weight, and eventually the cost of the imaging system. Consequently, spatial resolution, and therefore image quality, increase. A research group at General Electric (Corporate Research and Development Center, Schenectady, NY) has also developed similar circuitry.[2]

The second question regarding computational power is whether or not more computing power can increase frame rates in color flow imaging. High color flow frame rates are desirable in applications such as cardiology where visualizing flow around rapidly moving heart valves can be diagnostically important. For small, peripheral regions of interest, color flow images can be formed at real-time rates. For deeper structures, frame rates must be reduced significantly.

For example, with a 6 cm depth of field and a region of interest (ROI) of about 2 cm by 2 cm, defined by the duration of a time-gate and the number of A-lines included, most scanners can process color flow information and display it at frame rates in excess of 20 frames per second. Varying the length of the ROI typically does not decrease the frame rate, because the limit in this case is imposed by pulse-echo transit time and not signal processing requirements. The same 2 cm by 2 cm ROI can be processed for color flow images with a much larger depth of field, such as 20 cm, to show that the frame rate is reduced significantly. This reduction is not due to lack of processing power, because the same size region is being processed. Instead, the reduced frame rate is due to the added pulse-echo time for the deeper depth of penetration. The processing limit on frame rate can be observed when viewing a full width but short duration ROI with a 6 cm depth of field. In this case the frame rate is reduced, relative to the first example, because of the pulse-echo time associated with the added number of interleaved long duration pulses for acquiring color flow data. These frame rates are typically high enough for real-time imaging. In this case, extending the length of the ROI in a fixed depth of field reduces the frame rate. This reduction is due to insufficient computational power.

The ultimate limit on image frame rates is imposed by the time of flight of the ultrasonic pulse used to probe the tissues. In such cases, increasing computational power will not improve frame rates. However, for regions larger than a few square centimeters, signal processing rates can limit image frame rate. In those cases an increase in computational speed will result in an increased frame rate.

FUTURE TRENDS

Future ultrasound scanners are likely to rely even more on digital electronics and ASIC devices. The initial development costs and low production quantities make these a relatively expensive alternative currently found only in the top of the line scanners. However, this technology will likely be found in the full line of future scanners.

The increase in computational power provided by these circuits opens the way to several exciting possibilities. Significant improvement in image quality is available through *phase aberration correction.* In the process of image formation it is assumed that the speed of sound in all soft tissue is 1,540 m/s. In reality, speeds in soft tissue vary from about 1,470 m/s (fat) to about 1,650 m/s (muscle). As the ultrasonic beam passes through tissue layers of inhomogeneous composition or thickness, the phase front of the sound beam becomes distorted. This is particularly important for phased array systems where the time delays used to steer and focus the beams assume a constant speed of sound.

The current trend in phased array technology is toward larger apertures for improved resolution. However, phase aberration reduces system resolution and the peak pressure amplitude causing a loss of echo signal dynamic range. These effects become more severe as the array length is increased. Therefore, the phase aberration caused by tissue inhomogeneity impose an upper limit on the resolution available with large aperture array transducers.

The electronic time delays used in phased array systems for steering and focusing the beam can also be used to correct for this beam distortion. An example of the image improvement available by aberration correction is shown in Fig 1. Techniques for phase aberration correction that operate at near real-time frame rates (~ 0.1 s/image) have been reported[3] and may be available on future ultrasound scanners.

A *reduction in the image speckle* is another area of image processing where high-speed circuitry is necessary for real-time scanning. Speckle is common to ultrasonic imaging and refers to the granular appearance of homogenious tissue regions. It is the result of constructive and destructive interference of pressure waves at the phase-sensitive detector. Speckle reduction strategies involve the coherent sumation of independent image data. The simplest technique, available on most high quality scanners, simply sums consecutive frames. Small motions of the transducer or patient during scanning, for example due to breathing, provide slightly different images and an "average" image that is somewhat smoother. Frame averaging is effective at reducing temporally-varying electronic noise but has very little effect on temporally-invariant speckle noise. Independent images must be combined to obtain the full effect of speckle reduction. One technique[4] uses subsections of a phased array transducer to provide statistically independent views of the same tissue region. Image data are summed either in space (spatial compounding) or in frequency (frequency compounding) to reduce the coherence that produces speckle and provide improved detection of relatively large, low contrast targets such as tumors.[5] Speckle reduction via spatial or frequency compounding trades off spatial resolution for increased image contrast. Therefore, speckle reduction methods are important for diagnostic imaging of large, low-contrast targets. An example of the smoothed versus original image is shown in Fig 2.

Another approach to speckle reduction[6] relies on the statistical properties of the image data to reduce image speckle. If the statistical properties of pixels in a local region are consistent with a random scattering medium, then pixels in that region are smoothed. Regions that do not fit the random model, such as the boundaries of organs and vessels, are unmodified in the resulting image. All these speckle reduction methods use information about the

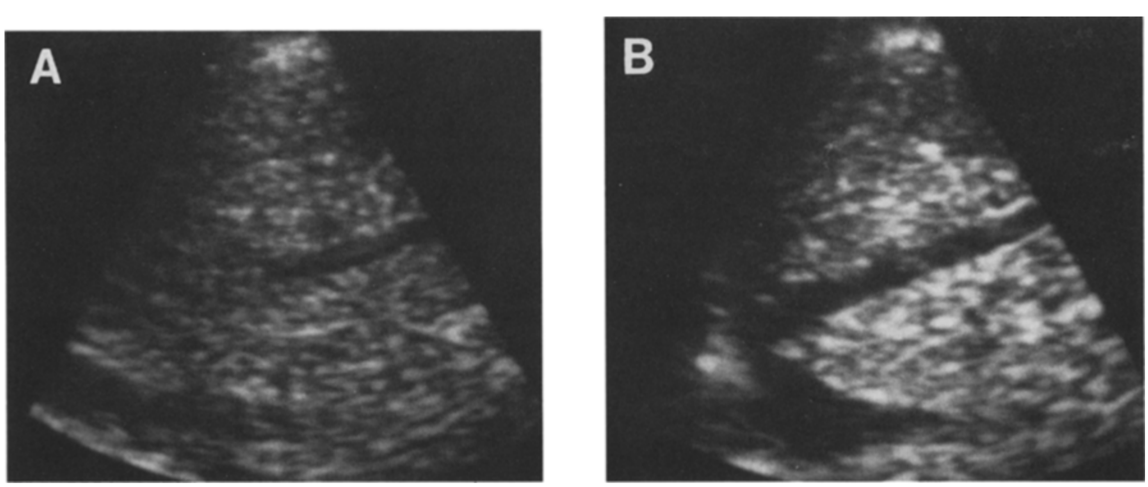

Fig 1. Images of a normal liver obtained in a slightly obese patient, acquired with a phased array scanner. The image on the left was acquired with normal signal processing, as done on a conventional scanner. The image on the right was processed, in real time, to reduce the effects of phase aberration. Note the improved detectability of the hepatic vein in the corrected image. (©1990 IEEE.[3])

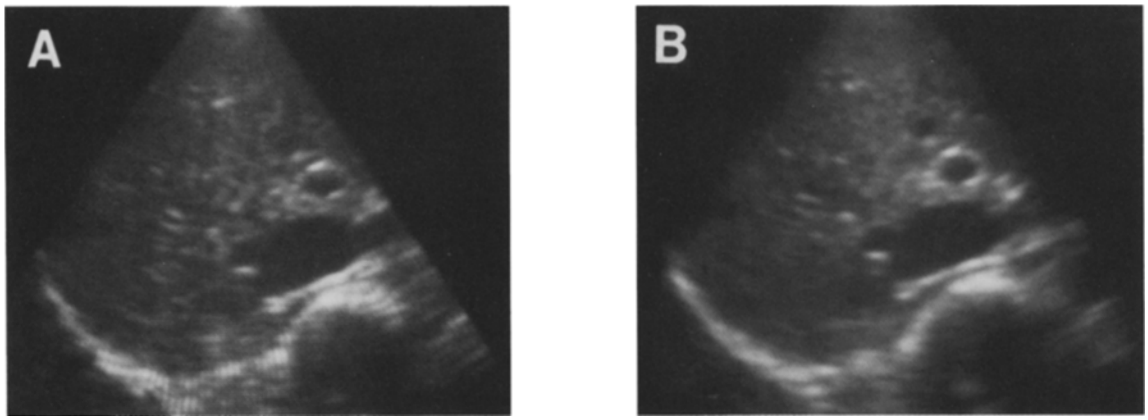

Fig 2. The image in (A) is that of an in vivo liver acquired with a 32 channel phased array imaging system. In (B) is an image acquired with the same system using spatial compounding to provide speckle reduction. Note the improved contrast and reduced vessel "fill in" in the compound scan relative to the simple scan.

physics of ultrasound-tissue interactions that is not normally used in conventional gray scale imaging.

Other methods of using detailed physical information about tissue, either modeled or measured from the data, are called *tissue characterization* or *quantitative ultrasound.* The motivation for tissue characterization research is to provide diagnostic information that allows for a more direct interpretation of the image data.

Several recent approaches to tissue characterization involve selecting several parameters extracted from the echo signal that are physically descriptive and sensitive to the changes in tissue microstructure that indicate disease. Previous experience with multiparameter ultrasonic analysis has shown these approaches to be very promising.[7,8]

In our recent work, we exploit acoustic scattering theory to estimate the size, shape, orientation, impedance difference, and number of scattering sites from the radio frequency (rf) echo signal spectrum. By separating these properties, we obtain information that describes spatial variations in these physical parameters. Using this analysis on data acquired in vitro[9] and in vivo,[10] we are developing methods to identify the microanatomy responsible for scattering in various tissues. If proven to offer diagnostic information, such information may lead to the development of task-specific imaging devices, or add-on hardware, as used for color flow imaging.

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