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Advances in transducers and techniques for diagnostic ultrasound

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Until recently, diagnostic imaging systems have remained stable with familiar modes to be measured. Emerging technologies are changing the measurement landscape rapidly. New developments are reviewed as well as those in research to examine the possible measurement challenges of the future. Improved focusing and steering are available with 1.5 D, multiple subaperture arrays, fully populated and sparse 2D arrays, and CMUT arrays. Encoded excitation pulses provide unique consideration. High frequency and special purpose arrays stretch the present limits of measurement bandwidth and resolution. Special auxiliary pulses and pulse sequences are being proposed for controlling, manipulating and destroying contrast agents and specially formulated therapeutic contrast agents. Improved harmonic imaging can involve multiple phased pulses as well as types of encoding. Active tissue characterization methods may include a means for tissue deformation as well as aberration correction. Imaging systems are undergoing a dramatic revolution in terms of their architecture, complexity, miniaturization and the role of software in image formation. These developments will be examined in terms of field distribution, pressure levels, pulse sequencing, waveforms and measurement

1. Introduction

This brief study examines the changes to the acoustic measurement landscape by recent developments in imaging systems and transducers. Familiar one-dimensional arrays are being replaced by 1.5D, 2D arrays, CMUT arrays and high frequency arrays. Both temporal and spatial encoding can dramatically alter the waveforms and beams being transmitted by imaging systems. Unique pulse sequences are commonly used for enhancing harmonic imaging and the viewing and controlling of contrast agents. In the future adaptive arrays may alter transmit sequences in real-time to correct for tissue effects. The growing field of tissue characterization is already demonstrating the improvements offered by additional signals. Finally, the architectures of the imaging systems are undergoing change.

These topics will be examined from an acoustics output measurements perspective. In particular, the effects of changes in field distributions, waveforms, pulse sequences, acoustic output levels and measurement requirements will be discussed.

2. Beamforming and Transducers

The mainstay of ultrasound imaging has been the one dimensional array. Although disguised in many shapes and sizes, this type of array has the common characteristics

of electronic focusing and steering in the azimuth (x-z) plane and fixed focusing by a mechanical lens in the elevation (yz) plane. In terms of acoustic output, this type of array has interesting limitations. The maximum pressure amplitude that these arrays can produce in water occurs close to the distance where both the elevation and azimuth focal lengths are coincident. To a first approximation, the pressure focal gain at this distance can be estimated under linear conditions as

$$p / p_0 = \text{ActiveArea} / (\lambda F) \quad (1)$$

where p_0 is the pressure at the transducer face, *ActiveArea* is the area of the active elements, λ is the wavelength, and F is the focal length. Here $F = F_{\text{azimuth}} = F_{\text{elevation}}$ for coincident foci. When these focal lengths are not coincident in the majority of cases, focal gains are far less, and multiple focal peaks occur along the beam axis. Once measured water values are derated, these multiple peaks cause highly spatially and amplitude variable effects.

Under ideal conditions, a two dimensional (2D) array has the capability to focus and steer to any point in space. In this case, the pressure focal gain of Equation (1) can be realized for a large range of practical focal lengths. Under these conditions a single focal peak occurs on axis; derated values will track this peaks. Focal gains increase as focal length is decreased.

Imperfect or intermediate realizations of 2D arrays are sparse, cross, and 1.5D arrays. Sparse and cross arrays use fewer numbers of active elements to reduce electronic requirements; however, Eq. (1) shows that the corresponding reduction in active area will substantially lower the achievable focal gain. An intermediate variant is the 1.5D array (Wildes et al., 1997) which accomplishes crude elevation plane focusing electronically. Along the elevation (y) axis the aperture is severely undersampled; consequently beam and sidelobe, grating lobe control is less than that for a 2D array and limitations on focal range can result. To first order, the active aperture of a 1.5D array is similar in size to that of a 2D array and coincident azimuth and elevation electronic focusing is possible; so the chief benefits of a 2D array are realized in terms of pressure focal gain over a more restricted focal length range.

The technical challenges associated with manufacturing 1.5 D and 2D arrays have been met. While a 1D array may have n active elements and n electrical connections and channels, a 1.5D array would have $(2-10)n$ electrical connections, and a 2D array, n^2 connections. 1.5D arrays have been available for several years and pioneered by General Electric (Wildes et al., 1997). In 2002, Philips introduced a fully populated real-time 2D array (Savord and Solomon, 2003) through a unique approach involving micro-beamformer ASICs in the transducer handle.

Another potential 2D array technology is called “CMUT” (Capacitive Micromachined Ultrasound Transducer). With this approach (Oralkan, et al., 2002), arrays are constructed from arrangements of tiny capacitor drumheads using standard silicon integrated circuit technology. 1D CMUT arrays have produced clinical images (Wong et al., 2003). The transduction mechanism and overall active area are less efficient than those for a PZT array but they provide adequate sensitivity for imaging. The

comments above for the acoustic output of 1D and 2D arrays apply to CMUT arrays with the constraints of a possibly smaller absolute source pressure level (p_0) and a slightly smaller active element area.

3. High Frequency Transducers and Measurements

High resolution clinical and tissue characterization applications are driving the need for high frequency transducers (Foster et al., 2000). Intravascular transducers in the 20 MHz and 50 MHz range have been around a number of years, as well as ultrasound microscopy applications extending more commonly from 50 MHz to several hundred MHz, though GHz frequency have been achieved. Newer efforts have concentrated on real-time high resolution imaging systems. Because the manufacture and design of high frequency transducers pose many challenges, most imaging work has been with single transducers, mechanically scanned. For faster real-time imaging, work has progressed on high frequency arrays (Ritter et al., 2002). For example, Ritter et al. have described a 30 MHz center frequency, f_c , 48 element composite array. With a fractional bandwidth greater than 50%, it has a 1.5 mm elevation aperture, a 4.8 mm azimuth aperture, and a elevation focal length of 7.5 mm. The one way full width half maximum azimuth beamwidth in the focal plane can be described as

$$FWHM = 1.2\lambda F / L = 1.2\lambda F \#, \quad (2)$$

where F and L are azimuth focal length and aperture and F# is F number. For this array, FWHM=0.092 mm.

What characteristics would a hydrophone need to adequately measure the acoustic output of this array? Both the frequency and small beamsize pose measurement problems for currently available hydrophones. The suggested current bandwidth requirements are $f_c / 20$ (Harris, 1998) at the lower end and $8f_c$ or 40 MHz, whichever is higher at the high end. For measurement of higher frequency transducers, an upper frequency hydrophone limit has not been established. Assume a modest $3f_c$ upper limit is needed to capture the third harmonic.

For many years, a conventional membrane hydrophone had a spot size of $500 \mu m$ and a useable bandwidth of 20 MHz. On the other extreme a prototype high frequency membrane hydrophone was developed at Hewlett Packard Research Labs (Lum et al., 1996). This hydrophone had a design goal spot size of $37 \mu m$ (actual diameter not measured but determined to be $< 100 \mu m$) and a -3 dB bandwidth greater than 150 MHz. The required hydrophone diameter for measurement in the focal plane of this transducer can be shown to be related to the F# and the FWHM for this array,

$$d_h = 0.52\lambda F / L = 0.52\lambda F \# = 0.431FWHM . \quad (3)$$

For this 30 MHz array, this diameter is $40 \mu m$. Larger diameters would cause spatial averaging (Lum et al., 1996). Then, the suggested bandwidth requirements would be approximately 1.5 MHz to 90 MHz. The Hewlett Packard hydrophone may meet some of these requirements and was used to measure this array (Huang et al., 2001),

but it was never made available commercially. Currently available piezoelectric hydrophones have difficulties meeting these specifications simultaneously. Even though it may be technologically possible to meet them, costs involved in manufacture would be high. Calibration methods at frequencies above 60 MHz are still being developed, as well as alternate technologies for measurement. Another possibility on the horizon is compensating the pressure measurement for the hydrophone response rather than having the hydrophone meet rigid flatness requirements over extremely wide bandwidths.

4. Advanced Signal Processing

In order to achieve better signal to noise, higher temporal resolution, faster frame rates and greater flexibility, spatial and temporal encoding of transmit waveforms have been proposed and implemented. A guiding principle here has been the application of matched filters on receive to unscramble the encoded waveforms. If the output y of a filter h to an input x is

$$y(t)=x(t)*h(t),$$

(4)

where $*$ denotes convolution, here in the time domain. An ideal matched filter has the characteristic,

$$y(t)=x(t)*h(t).$$

(5)

Since the matched filter reverses the input in time,

$$y(t)=Ax(t)*x^*(-t).$$

(6)

This process can be visualized graphically as $x^*(t)$ being slid, one time increment at a time, past $x(t)$ and their product summed. The device for accomplishing this decoding is called a matched filter correlator. Popular binary codes are the Barker and maximal length (M)-sequences (Carr et al., 1972). A characteristic of these codes is a correlation peak after decoding equal to an amplitude M which is the number of individual binary bits in the code. Note that the transmitted waveform is M binary units long but the correlation peak is compressed to one binary unit in width. On either side of this peak are time sidelobes so that the signal to noise ratio is the peak to the highest time sidelobe.

Another major type of encoding applied to ultrasound is the chirped waveform which has been used widely in radar systems (Lewis, 1987). This waveform usually consists of an instantaneous frequency that changes linearly within the pulse as either an upchirp (increasing instantaneous frequency) or a downchirp (decreasing instantaneous frequency). Matched filtering also creates a high correlation peak related to the time-bandwidth product.

A way to eliminate time sidelobes is to use Golay codes. These complementary codes require two or more transmissions so that the total number of encoded signals are

decoded and summed on receive in such a way as to cancel the sidelobes completely. An excellent discussion of the advantages and implementations of binary, chirp and Golay codes for medical ultrasound imaging can be found in Chiao and Hao (2003).

While these types of codes can be considered as temporal, it is also possible to create spatio-temporal codes for sending multiple encoded beams out simultaneously. These beams hit targets and their pulse-echoes are decoded not only in time but direction (Shen and Ebbini, 1996). To accomplish this sort of magic, orthogonal codes are preferred. Orthogonality means that the particular codes used do not interfere with each other and pass the information for the selected beam independently of the codes for other beams. Orthogonal Golay codes can serve this purpose.

We now examine what effect these encoded signals have on acoustic output and its measurement. As explained in Chiao and Hao (2003), it is advantageous to use a number of cycles for each binary code time slot to improve signal to noise. In general, the transmitted temporally encoded pulses are long and often strange looking compared to traditional transmitted pulses, and they are of lower amplitude. Still it is possible to have substantial time average power W and reasonable derated I_{spta} values due to the larger duty cycles. In this case it is the Thermal Index (TI) limit rather than the Mechanical Index (MI) limit that will come into play first. Spatio-temporally encoded pulses present interesting measurement and alignment challenges as there may be few, if any, simple focusing peaks or recognizable landmarks for the multiple beams.

5. Tissue Characterization and Correction

Tissue characterization is a growing research area which seeks to invert or determine tissue properties through an analysis of tissue backscatter or of other measured parameters for purpose of diagnosis. Some characterization techniques employ an extra excitation field such as audio frequency vibrator for sonoelasticity (Gao et al., 1995). This extra vibratory field falls outside the usual range for acoustic output measurements and regulations. As more active tissue characterization methods are developed, means of characterizing their acoustic output will be necessary.

To correct for the aberration of wavefronts in tissue, time delay and amplitude corrections need to be applied on both transmit and receive for maximal focusing improvements. Complete real-time phase aberration correction involves the transmission of a wavefront adapted to correct for propagation distortion through tissue. If measured in water, this transmitted adapted wavefront may focus poorly and be quite different than its properties and acoustic output levels in real tissue. Furthermore, if this process were carried out in real-time, transmit waveforms would change from frame to frame. Should this correction process be realized, considerable work will be required to modify present measurement techniques to characterize the acoustic output levels of these adaptive arrays adequately.

6. Pulse Sequences for Contrast Agents and Harmonic Imaging

The widespread application of harmonic imaging has brought about special signal processing techniques to enhance images. Examples of these techniques are pulse inversion and amplitude (power) modulation (Averkiou, 2002). Pulse sequences for harmonic imaging involve transmit pulses repeated in the same direction (typically 2-

3 pulses, one per acoustic line) and differing in amplitude or phase from each other. Summing and scaling methods are used to emphasize spectral regions such as the second harmonic. In addition, pulse shaping is used to minimize unintentional second harmonic transmission. In terms of acoustic output measurement, the pressure waveforms and fields used fall within the present levels and pose no special problems as long as the different pulse shapes and amplitudes are tracked.

Contrast agents also generate harmonic frequencies but many types of transmit pulse sequences are applied for purposes other than second harmonic image enhancement. Contrast agents behave differently with various types of insonification fields that differ in their spectral content and amplitude. They are also delicate and can easily fragment as demonstrated through time streak photography by Chomas et al. (2000).

A number of specialized transmit pulse sequences have been devised to obtain different effects. Contrast agents can be fragmented by the first insonification pulse. Transmit pulses may be given at long triggered intervals to control the destruction of contrast agents over longer periods of time. To enhance the appearance of contrast agents, a second insonification transmit pulse of a different amplitude and frequency can lead to a variety of effects (Frinking et al., 1999). Pulse sequences can also be designed to emphasize certain bubble harmonics or sub-harmonics (Phillips, 2001).

An exciting emerging area is the application of designer contrast agents for therapeutic applications. Agents are engineered to have special coatings or attachment mechanisms for seeking out and bonding to sites of injury or disease (Lindner, 2001). Once bonded, these agents can be imaged. Another twist is to have the agents carry a drug payload which can be released by insonification at the desired site. These agents can also be guided or manipulated by radiation forces from ultrasound beams (Morgan et al., 1997).

In summary, sequences of transmitted pulses may differ in interval (duty cycle), amplitude and/or frequency to resonate and fragment contrast agents. These may be combined with pulses discussed above for harmonic imaging signal processing. For therapeutic agents, in addition to imaging pulses are control pulses for manipulation, enhancement, and drug delivery. So far, proposed pulse sequences for contrast agents have usually been complicated combinations within present measurement capabilities; however, this does not rule out that new ones outside present limits may be proposed. Current cases that employ sequences of standard pulses should fall under conventional measurement methods if the sequencing is known.

7. Measurements for New Imaging Architectures

A remarkable trend in ultrasound imaging is the emergence of portable systems. Through a variety of techniques, miniature imaging systems with good image quality as well as other modes such as color flow imaging have become commercialized. A review of these systems can be found in a special issue of the Thoraxcentre Journal (December 2001).

After fifty years of the same beamforming, ultrasound imaging systems with new architectures are starting to appear. An example of one of these new systems is the zone-based imaging approach (Mo et al., 2003). Designed to overcome the transmit

roundtrip delay limitation of sequential acoustic lines, the zone-based imaging systems has only a few (3-5) very wide transmit beams rather than a hundred or more narrow ones used in conventional imaging. This simplification speeds up the frame rate by over an order of magnitude. Special receive processing includes storing real and quadrature signals from all elements and implementing real-time receive beamforming with digital signal processors. Temporal binary encoding has been reported on prototype systems (Mo et al., 2003).

Our discussion of measurements will focus on portables and the zone-based architecture. In principle, the acoustic output measurement of these small systems should be the same as those of larger systems. Because of extreme miniaturization, many parts of the imaging system are no longer accessible unless special measurement test points for synchronization and acoustic measurement purposes are designed in. For the zone-based system, wide transmit beams pose no major problems. Acoustic output measurements should be straightforward. Earlier comments apply to temporal encoding, if used.

8. Conclusion

Comments for the measurement aspects of the individual technology and signal processing improvements discussed have been consolidated in the table below. More information about the technology advances can be found in Szabo (2004).

Measurements Comparison Summary Table

Changes	Fields	Waveforms	Sequences	Output levels	Hydrophones
2D Arrays	More focused	Same		Higher in water	Adequate
High Frequency	Scaled down	Scaled down	Same	Derated Lower	Need new
Encode	Multibeam!	longer	Complicated	TI limited	Adequate
Tissue Char.	Uneven wavefronts	Same	Same	Weaker Focusing in H ₂ O	Adequate
Contrast Agents	Same	Variations	Complicated	Generally similar	Adequate
Imaging Architec.	Similar / complicated	Same or longer	Similar or complicated	Same	Adequate

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