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Advances in ultrasound

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Abstract Significant advances have been recently introduced into various fields of technology, taking advantage of the use of new piezoelectric materials and the large diffusion of broadband transducers. Various types of modulation may be applied to the pulse characteristics, using single pulse, multipulse or multiline techniques, and resulting in improved spatial resolution and better penetration. Non-linear imaging uses the harmonics component, which is generated by tissues or by contrast agents. Different modalities can be used to separate harmonics from fundamental bands from the received signal. New Doppler modes have been developed, whereas grey-scale flow imaging allows the simultaneous imaging of blood flow and tissues. Compounding techniques improve the contrast resolution of tissues and reduce artefacts. If 3D techniques are now currently available, real-time 4D imaging has been recently introduced. Elastographic imaging is still under evaluation, but promising clinical results have been shown. Recent release of the DICOM specification has made the full integration of ultrasound to the PACS systems easier. All these advances indicate that the contribution and potential of ultrasound in patient management is still growing.

Keywords US · Technology · Doppler modes

Introduction

Engineering advances in ultrasound over the past decade have resulted in clinical technology that may permit physicians to use their ultrasound scanners in ways they have previously only dreamt about, giving ultrasound a major role in the diagnostic process and important advantages compared with other imaging modalities [1, 2].

The purpose of this review is to summarize some of the main advances that have been introduced recently into various fields of technology and are available for practical use by radiologists.

Advances in the signal transmission process

Advantages of broadband transducers

New piezoelectric materials with lower characteristic acoustic impedances and greater electromechanical coupling coefficients have been developed in the past years. Also, improvements in the understanding of the way small transducer elements vibrate, as well as in the construction of absorbing backing layers and quarter-wave impedance matching layers, have been achieved [3, 4]. For example, ceramics of variable thickness and shape, or multilayer transducers (Hitachi), are currently used. These technical advances have permitted a more accu-

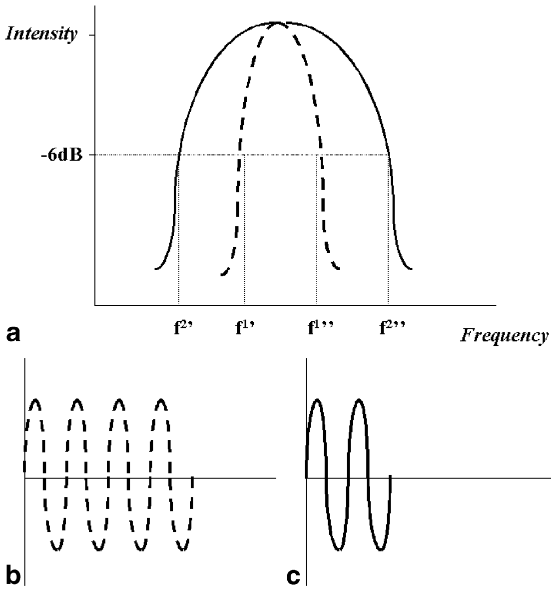

Fig. 1a–c Broadband transducers

rate shaping of ultrasound pulses, in terms of the control of transmission frequency, amplitude, phase and pulse length.

These developments have also permitted the widespread introduction of broadband transducers onto the market. From a theoretical point of view, the shorter the pulse length, the broader the pulse bandwidth (Fig. 1); therefore, the principal benefit of using a broadband transducers is the improvement of the axial resolution, due to shorter pulse lengths [3]. With broadband transducers, higher transmission frequencies give a better spatial resolution in the near field, whereas lower frequencies allow better penetration in the far field. In addition, as detailed below, harmonic imaging requires large transducer bandwidths, which respond to frequencies at least twice those of the transmitted pulse.

Moving to higher-transmission frequencies

The improvement in piezoelectric materials, the design of transducers, early digitisation and better analysis of received echoes with a lower level of noise allow the use of higher-emission frequencies for imaging. This results in higher axial and lateral spatial resolution while preserving good penetration. For example, it is now possible to evaluate the liver in adults with probes working at approximately 7 MHz.

High-frequency ultrasound (≥20 MHz) is another growing field, with increasing applications in dermatology, stomatology, ophthalmology, and the musculoskeletal field (Fig. 2) [5]. Original developments include, for example, measurement of very low blood flow velocities (<0.5 mm/s) in 100- to 300-μm-diameter vessels [6]. Very high-frequency ultrasound (≥50 MHz), including three-dimensional imaging, is also usable for the evaluation of the cornea and the anterior segment of the eye, allowing for precise biometry and delineation of pathological processes [6]; however, although many different dedicated transducers have been used for research, most of them are not yet commercially available.

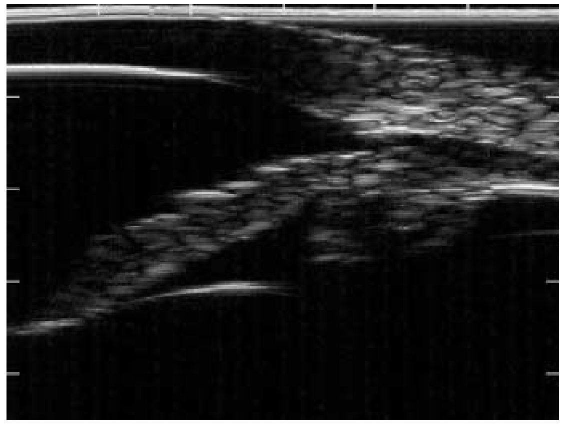

Fig. 2 Image of the anterior chamber of the eye, obtained at 20 MHz

Modulating pulses and lines

As mentioned above, the response of piezoelectric materials can be readily tailored, which allows modulation of the pulse characteristics, including the transmission frequency bandwidth, amplitude, phase and length. As a relative variable, phase can only be determined by comparison with a reference waveform, or with another pulse. This can be achieved by using multiple-beam formers, allowing a comparison of the phases for pulses from adjacent lines (coherent image formation; Acuson).

A wide variety of pulse-characteristics modulation processes have been introduced by commercial companies, but detailed specifications are often not published. We describe below some new processes that have been recently introduced which illustrate the great potential for improvements in the ultrasound technique. These new modalities can be classified according to the number of pulses and scan lines used.

Single-pulse techniques

With single-pulse techniques, modulation has been applied in the following ways:

1. To limit the overlap between the fundamental frequency and harmonic response by narrowing the frequency

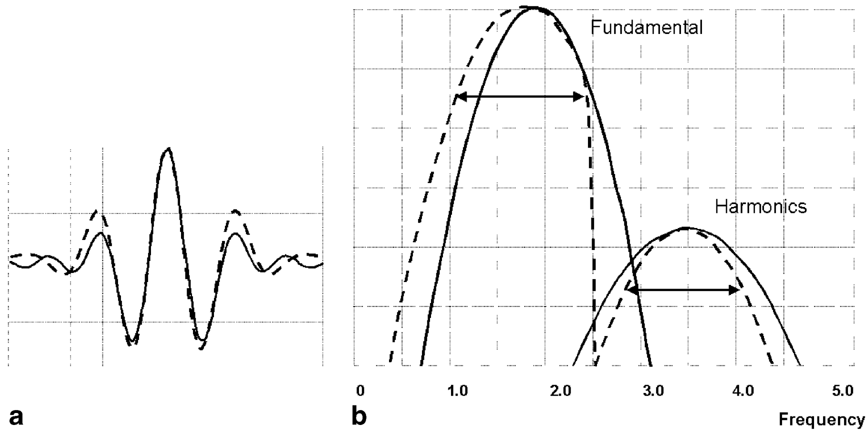

Fig. 3 Principle of precise pulse shaping (Acuson-Siemens)

spectrum. Usually narrowing the frequency spectrum of a pulse results in a longer pulse and therefore a degradation of the axial resolution. Subtle shaping of the transmit pulse allows better separation of the fundamental and harmonic spectra, preserving broadband harmonic imaging with excellent axial resolution (Fig. 3).

2. In chirped emission (Acuson), a long, specially shaped transmitter pulse (a chirp) varies in frequency and amplitude within the duration of the pulse itself. On receive, echoes pass through a filter that is an exact time-reversed replica of the transmitted chirp. At the output of the filter, the received signal from any small scatterer is “tall and narrow”, meaning that the image has very high axial resolution, increased signal-to-noise ratio and increased penetration

Multi-pulse techniques

In coded-emission mode (GE, Esaote), the scanner transmits not a single pulse but a sequence of 8–22 short, high-frequency transmission pulses that may have different phases and are modulated in a code sequence. Comparison between transmitted pulses and received signal shapes using matched filtering (decoding) is subsequently performed with a very high sampling rate (Fig. 4) [7]. This technique, which has long been used in radar and sonar, results in increasing image penetration without compromising axial resolution or increasing the transmitted peak pressures.

Multi-pulse techniques are based on the emission of consecutive pulses of opposite phase. In the first approach, called pulse inversion (ATL-Philips) or phase inversion (Siemens), two pulses of opposite polarity are transmitted along the same line. The subtraction process of the two signals results in a relative increase in the non-linear response from tissues by cancelling the response from structures which have small non-linear components (Fig. 5); however, like colour Doppler, these methods are susceptible to motion artefacts which allow

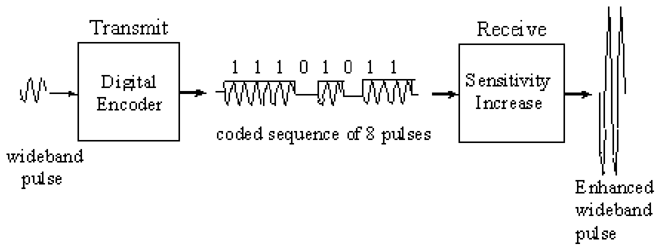

Fig. 4 Principle of coded emission (GE)

the fundamental signal to leak through and masquerade as a harmonic.

A complementary approach is called amplitude modulation (Agilent), in which the two pulses differ in phase and amplitude, one having half the amplitude of the other one. Prior to the sum being performed the received signal from the half amplitude pulse is multiplied by a factor of two allowing greater emphasis of the non-linear components.

More complex multiple-pulse techniques are now being introduced. For example, following the first two pulses of opposite phase, a third one with an inverted phase relative to the second can be used. This technique, called power pulse inversion (ATL-Philips) (Fig. 6), is based on power Doppler shift evaluation. As the time delay between pulses is very short, there is assumption that the target motion is constant during the acquisition. This technique seems to have some advantages in reduction in motion and flash artefacts and allows more complete separation between contrast and tissue. It is used mainly for contrast-enhanced sonography.

In the previously described methods, working in multiline nearly eliminates the frame reduction caused by using multiple pulses.

Multiline techniques

Using multiple beamformers allows the direct integration of the phase information from the signal received from

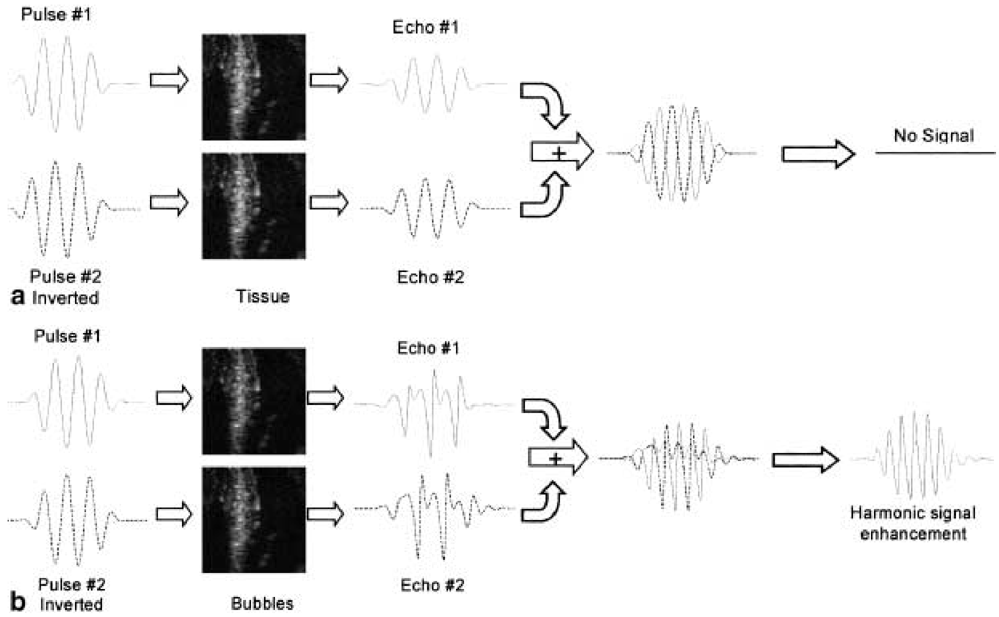

Fig. 5 Pulse inversion mode (ATL)

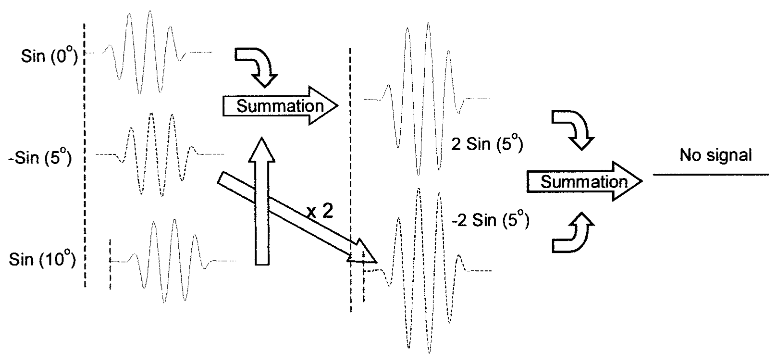

Fig. 6 Principle of power pulse inversion (ATL)

adjacent lines, and similar approaches as mentioned above have been developed with potential advantages. For example, the amplitude and phase information of the received signals from two adjacent lines can be added using dual receive beams, allowing the cancellation of the fundamental signal and the emphasis of harmonic signals, as used in pulse cancellation technique (Acuson-Siemens) (Fig. 7). This method offers the advantage of maintaining the frame rate and exhibits less motion artefact.

Improving spatial resolution

In the dynamic transmit focusing modality (Acuson-Siemens), edge elements have a longer pulse excitation than centre elements, making the US beam focus at two different points in the insonated field, improving lateral resolution. As the resulting wavefront has the characteristics of a short excitation pulse, the axial resolution is preserved.

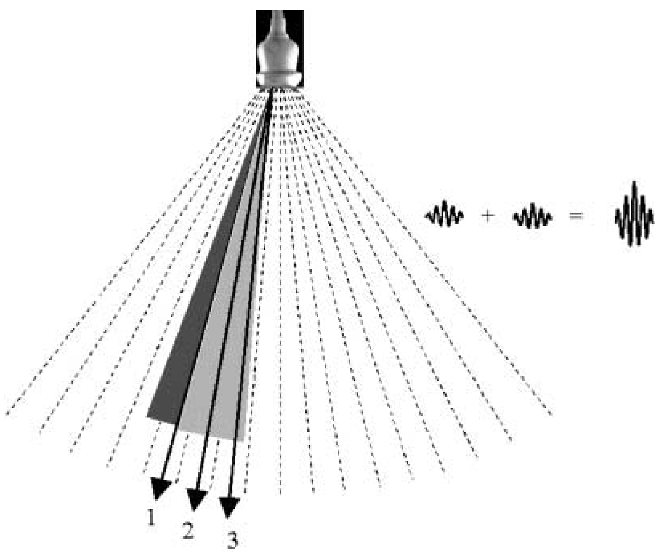

Fig. 7 Principle of coherent phase imaging (Acuson-Siemens)

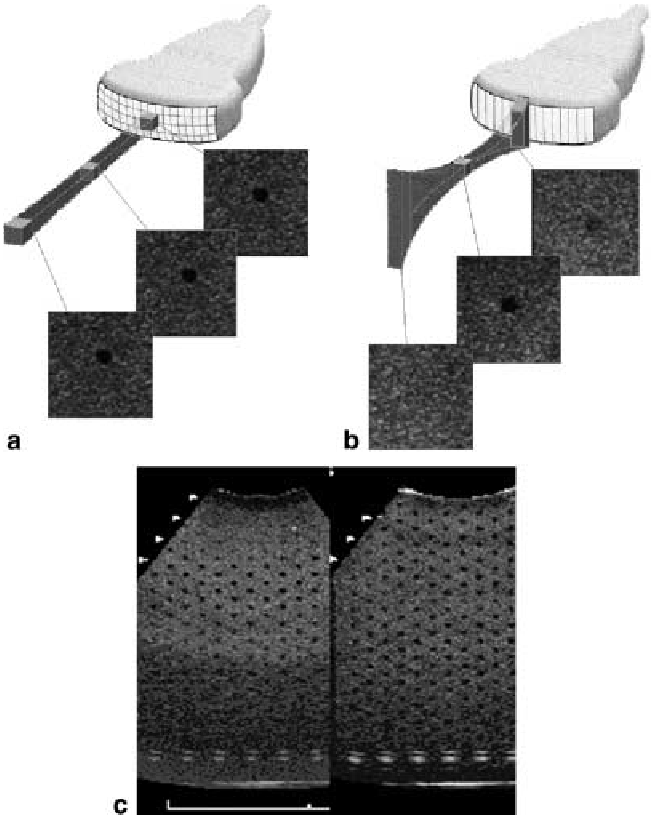

Fig. 8a–c Active matrix transducer (GE)

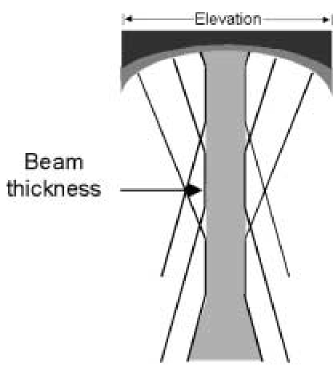

Fig. 9 Hanafy lens (Acuson-Siemens)

Improvement of the elevational resolution is also a challenge for ultrasound imaging. The emerging applications of contrast agent imaging and 3D imaging require thinner and more uniform image slice thickness for better performance. The use of so-called 1.5-D transducers (GE, Siemens), which are designed to include three to seven parallel rows of short elements, makes some progress in this direction (Fig. 8). The method allows focusing control in the z-plane, improves the spatial and contrast resolution, reduces partial-volume artefacts, but requires the management of a high number of channels [4].

Another approach is to add a special acoustic lens, a Hanafy lens (Acuson-Siemens) that uses a variable-thickness crystal to produce a narrow and uniform image slice thickness and, simultaneously, an extremely broad bandwidth pulse. The outer portion of the crystal resonates at the lowest frequency, and is focused, in both transmission and reception, at the deepest part of the image where the low frequencies also provide better penetration. The central portion is thinner, resonates at higher frequencies and focuses more superficially (Fig. 9).

Advances in the signal reception process

Non-linear or harmonic imaging

The observation of harmonics generated by contrast agents as described by Schrope and colleagues in 1992 [8] has focused research on the acoustic harmonics described, allowing marked improvement in standard use of ultrasound as well as for contrast agent imaging [9].

Basic principles and characteristics of harmonics

Ultrasound pulses contain a range of frequencies centred around the "zero-crossing" frequency of the pulse. When such pulses propagate through tissue, their frequency content is altered by numerous mechanisms. The best known of these is the more rapid attenuation of the higher-frequency components which causes a down-shift in the centre frequency of the pulse. Another mechanism, that of non-linear propagation, leads to a generation of higher frequencies. At any instant, the peaks of the pressure waves travel slightly faster than the troughs because of the different velocity of ultrasound propagation in compressed tissue when compared with relaxed tissue, and this gives rise to the generation of harmonics. Although the amount of harmonics that each slight pulse distortion generates at any given instant remains infinitesimal, the cumulative harmonic intensities increase as the pulse propagates through tissue (Fig. 10) [10, 11, 12, 13, 14, 15]. The amplitude of the harmonics is related to the non-linear parameter B/A, which is an inherent characteristic of the tissue [16].

Tissue harmonics intensity is virtually zero at the skin, and increases with depth up to the point where tissue attenuation overcomes this build-up and causes them to decrease again. At all depths, however, tissue harmonic intensity remains lower than that of the fundamental (Fig. 11). That tissue harmonic imaging allows the use of lower frequencies for transmission, because the generated harmonics that are used for imaging have shorter wavelengths. This mode benefits from a relative auto-

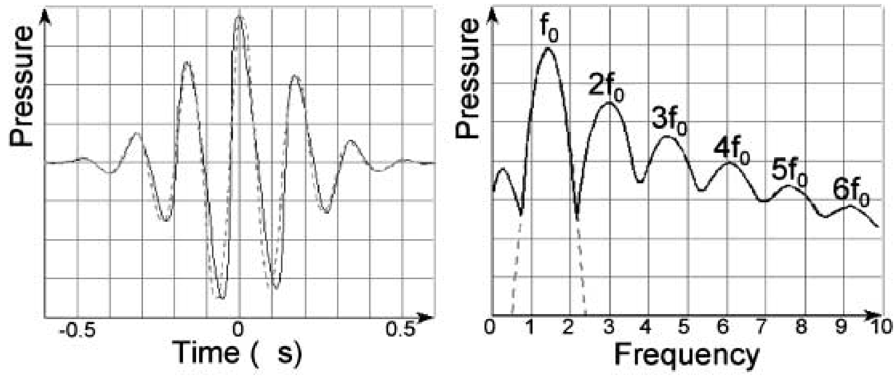

Fig. 10a, b Principle of harmonics generation

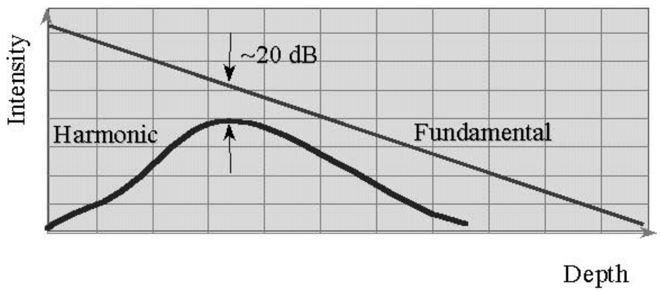

Fig. 11 Evolution of relative intensities of fundamental and harmonics signals (dB) with depth

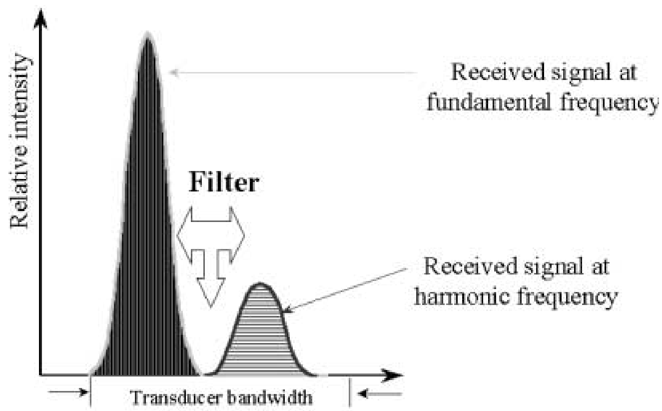

Fig. 12 Detection of the harmonic components with frequency filtering

focusing as the harmonics are generated mainly in the centre of the beam where the acoustic pressure is maximum. Additionally, the primary cause of image noise and clutter is the composition of the body wall, in which fat, skin layer thickness and hydration level are some of the principal causes of ultrasound beam distortion and scattering. In addition, lateral and slice thickness side lobes [17] as well as reverberation artefacts also contribute to generate image clutter; however, this distorted and scattered energy is much weaker than the transmitted energy and therefore generates much weaker harmonics. Finally, a tissue harmonic image contains minimal noise and clutter compared with fundamental imaging. This results in greater sensitivity of harmonic modes for lesion detection [18, 19, 20].

When using contrast agents, harmonics are related to specific interactions between ultrasound pulses and microbubbles. These applications have been extensively developed [20, 21, 22, 23].

Image reconstruction

Harmonic images are formed by utilising the harmonic components that are generated by tissue and by cancelling out the fundamental echo signals that are generated directly from the transmitted acoustic energy (Fig. 12) [2]. This imaging mode requires the use of wide-band transducers for correct transmission and reception over a wide band; however, the need for separation between the fundamental and harmonic components means that the bandwidth of the transmit pulse must be limited, which leads to lower axial resolution [3].

A receive filter can be used to filter out the pure fundamental signal and the image can then be reconstructed from the remaining harmonic signal. The harmonic filter must be matched to the harmonic band, wholly excluding the fundamental frequency band and therefore preventing corruption by noise and clutter from the fundamental.

Specific modalities can be used to improve the separation between the two frequency bands, to enhance the display of the harmonic image. These are detailed above, and include precision pulse shaping and multi-pulse techniques such as, for example, pulse/phase inversion and single-pulse cancellation. Recent clinical studies show significant improvement in the visualization of both normal and pathological tissues [24].

Complementary approaches have also been recently developed for contrast enhancement based on these two principles. One technique called ultraharmonics (Agilent) is a high-MI technique with a radiofrequency (RF) filter to remove tissue harmonic signal. The use of an RF filter between the second and third harmonics reduces unwanted signal from tissue. This implies the use of a large bandwidth transducer which responds to frequencies higher than second harmonics. Motion artefacts are minimized, and lateral and axial resolutions are improved. The use of subharmonics has also been proposed based on the fact that subharmonics are more specific for microbubbles.

Doppler techniques

In directional power Doppler (Toshiba, Esaote, ATL-Philips, GE) positive and negative flows are separated before the estimation of signal power. This allows the flow direction to be encoded in real time with a two-colour scale but does not provide any detail on haemodynamics within the vessel as the colour Doppler mode does; however, in addition to the better sensitivity to flow and lower dependence on angle of the power mode, the depiction of the direction of flow may be helpful in some cases.

It is also possible to calculate the Doppler shift observed between two consecutive acquisitions in both harmonic and pulse inversion modes, leading to harmonic power mode and power pulse inversion (ATL-Philips). These modes, which are under clinical evaluation, should give a better visualization of the harmonic response from microbubbles with a more effective cancellation of the fundamental.

Grey-scale flow-imaging techniques

Grey-scale flow-imaging techniques are based on B-mode and allow the simultaneous imaging of blood flow and tissues without a threshold decision and colour overlay, as is usually necessary in Doppler modes. B-flow (GE) works by comparing the backscattered signals from a pair of coded sequences emitted with a very short time gap. If blood cells have moved between the two pulses, then the two received signals will be slightly different, and if they are subtracted one from another, then there will not be perfect cancellation. The remaining signal can then be boosted and displayed as a representation of movement. Signal brightness is determined by blood echo strength and blood velocity but without linear relationship [7, 25, 26].

The Scieflow (Siemens) seems to be based on a similar principle, with an enhancement of echoes from mobile targets.

Grey-scale Doppler imaging

Dynamic flow (Toshiba) is based on a broadband Doppler technique, with a resolution similar to B-mode grey-scale imaging. It can be used with or without contrast agent. After two transmit pulses, the detection of blood cells and/or contrast agent bubbles is obtained from a specific algorithm which combines an adaptive filter and a digital motion artefact eliminator. Small vessels are better displayed, with a high sensitivity and temporal resolution (Fig. 13).

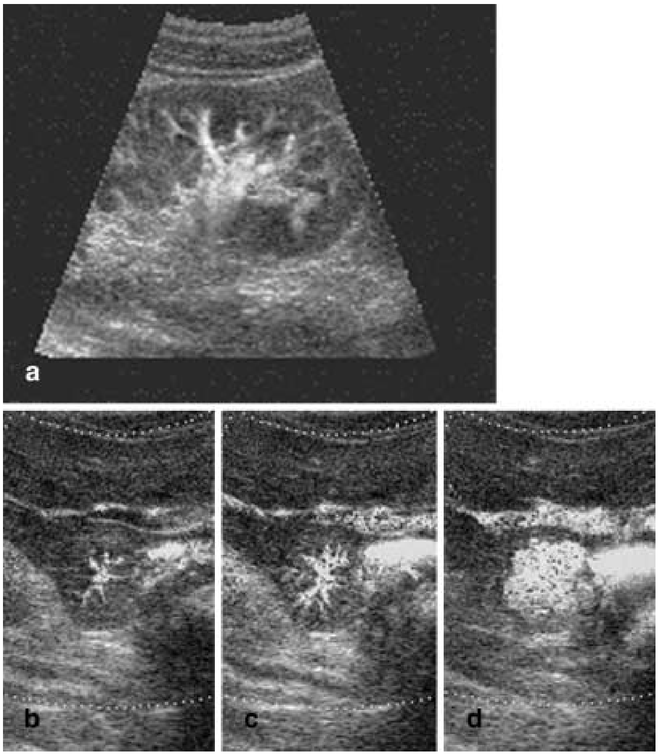

Fig. 13a–d Example of grey-scale flow imaging (Dynaflow technique; Toshiba)

Compounding, equalization, and extended field acquisitions

Transmit compounding

The principle of compounding is to combine images that have been obtained using different spatial orientations [SonoCT (ATL-Philips), Scieclar (Siemens), Sonoview compounding (Toshiba); Fig. 14]. The digital beamformer electronically steers transducer arrays at up to, respectively, nine and five steering angles, at real-time acquisition rates. Compounding improves the contrast resolution of soft tissues and lesions by reducing artefacts such as speckle, clutter and noise, without compromising other beneficial image characteristics such as spatial resolution. Compounding also reduces shadowing from strongly reflecting interfaces such as organ boundaries, fascial planes, vessel walls, tendons and ligaments, because they reflect only weakly at glancing angles.

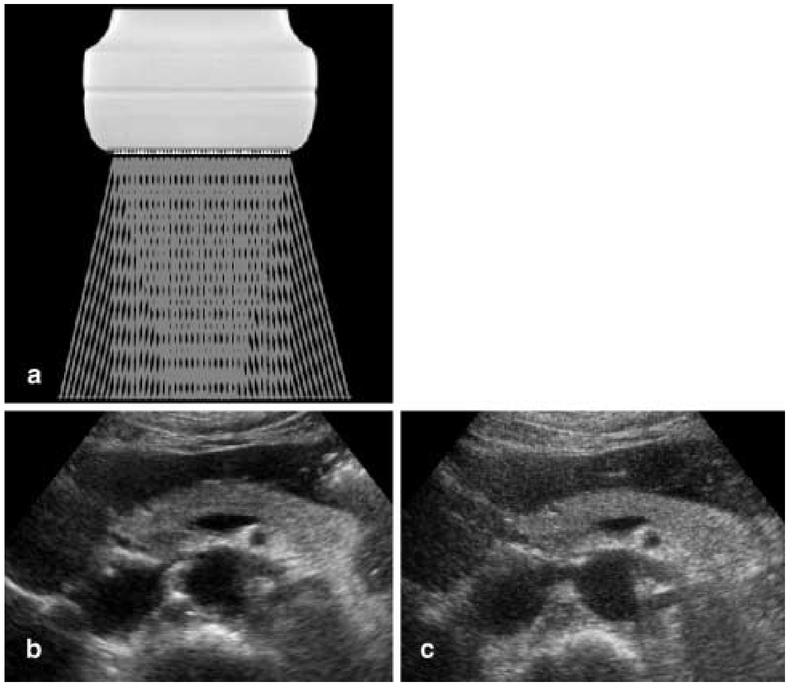

Fig. 14a–c Transmit compounding (SonoCT; ATL-Philips)

The simultaneous emission of two different frequencies (transmit frequency compounding; Acuson-Siemens) results in better contrast resolution in addition to the reduction of both electrical noise and speckle [3, 27, 28].

Image equalization

There is a variety of post-processing modalities, the benefit of which being to make images more uniform in two dimensions. Currently, there are two general approaches to this challenge. One approach, represented by digital imaging optimization (Toshiba) and automatic tissue optimization (GE), utilizes a histogram analysis of a small region of interest in an image and applies derived optimization factors that adjust contrast, brightness, and gain to create an optimized image. Another approach, tissue equalization (Acuson-Siemens), is based on regional and adaptive methods to analyse, using regional speckle statistics and thermal noise identification, and apply an algorithm to perform region-by-region adjustment to lateral gain, depth gain and overall gain. When using high-frequency transmission imaging, the adaptation of the gain vs the depth, associated with signal distortion reduction and dynamic range optimization in digital adaptative gain control (Esaote), also improves image homogeneity.

Extended-field acquisition

In extended-field acquisition processes, including Sciescape (Siemens), and more recently Freestyle (Acuson), Panoramic(ATL-Philips), LOGIQview (GE) and Panoramicview (Toshiba), the information regarding position is recovered directly from the ultrasound images themselves, allowing the reconstruction of a wide image by a progressive addition of data acquired during a hand sweep. These modes are available in greyscale and power modes (Fig. 15).

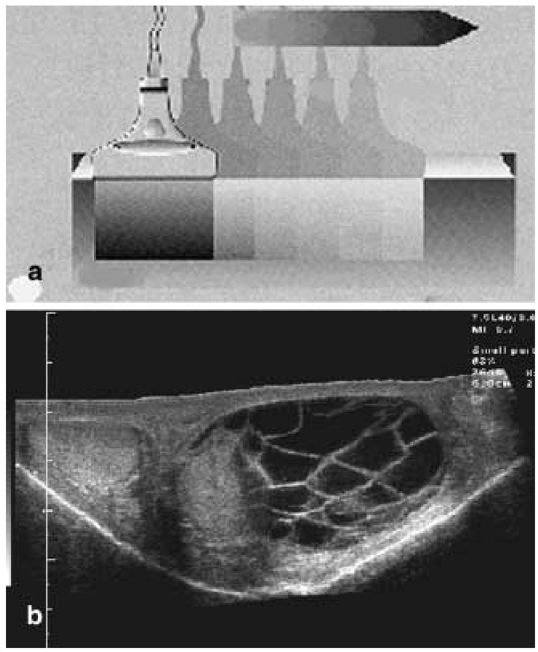

Fig. 15a, b Example of extended image (Sciescape; Acuson-Siemens)

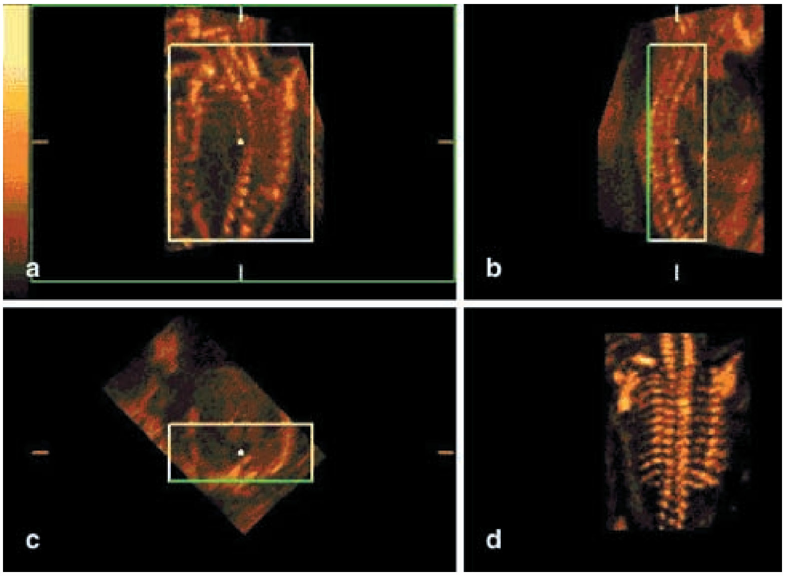

Fig. 16a–d Principle of 3D reconstruction. Example of a fetal spine evaluated in multiplanar views (Kretz-Medison)

3D imaging

Many companies now offer 3D capability options. The challenge and major differences come from the specific method used to locate the position of the slice within the volume [1]. In the future, 3D techniques should logically benefit from the continuous improvement of 2D greyscale imaging.

The main clinical application fields are gynaecology, obstetrics and cardiology [29, 30], but promising results have been obtained in other fields such as the neonatal brain, prostate, and from endoluminal ultrasound [31, 32].

3D ultrasound acquisition

Acquisition is generally achieved using a free-hand sweep. This requires a separate position sensor to encode the location of the transducer, which commonly is based on the use of magnetic field, with a small sensor being mounted on the transducer; however, such systems are susceptible to distortion of the local electromagnetic field by metallic objects (e.g. metallic bed, medical device) that can increase acquisition-related artefacts and positional measurement inaccuracy [1]. Position sensing during freehand acquisition may also be accomplished using an integrated 3D position sensing technology based on acoustic image data.

A third system consists of using a dedicated transducer which contains a mechanized drive allowing the performance wedge, linear or rotational scanning, and a position-sensing system. For each slice, the relative angle between slices is therefore exactly known, minimizing distortion in the resultant scan [1].

Finally, it is possible to acquire real-time 3D data by using beam steering and parallel processing with multidimensional, or mosaic arrays [30]. The 2D arrays have a similar number of rows and columns of elements, i.e. 64×64, giving 4096 elements. Although this approach is doubtless the most promising technique, issues of interference and computing/storage bandwidth need are very complex [1, 4].

3D reconstruction

User interface has been significantly improved during the past few years, and volume data can be easily displayed and manipulated on most current US units. Whatever the acquisition technique, some form of interpolation is still necessary to fill in any gaps in the volume between acquired images. Digitally stored volume data can then be displayed in a multiplanar array that simultaneously shows three perpendicular planes throughout the volume: axial; sagittal; and reconstructed coronal views. The volume can be explored by scrolling through parallel planes in any of the three views, and by rotating the volume to obtain an optimal view of the structure of interest. Data can also be displayed as true 3D images using various rendering algorithm, including maximum intensity projections and transparent and surface renderings (Fig. 16) [29].

Other potential improvements include 3D display by volumetric holography, or the creation of virtual reality models [30]. Easy storage and transfer of 3D volume acquisitions by the Web or PACS allow for later analysis, or teleconsultation by experts [33]. It is still not clear if such analyses will be available on non-proprietary workstations, i.e. those dedicated to general imaging.

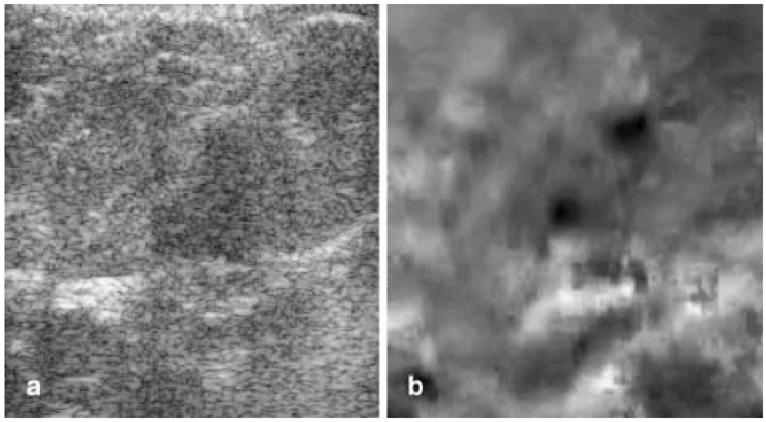

Fig. 17a, b Example of elastographic acquisition

ECG triggering and 4D imaging

Electrocardiogram triggering allows the synchronization of the data to the appropriate time of the cardiac cycle, either in a real-time acquisition or in a retrospective manner. Respiratory gating is also of interest in cardiac studies, especially for the trans-oesophageal approach. Both triggering techniques increase the acquisition duration. The development of ultra-fast, continuously rotating phased-array transducers could be an appropriate answer to these limitations. These modalities allow, for example, the display of the opening and closing of the cardiac valves [30].

Real-time 4D imaging (Kretz-Medison), obtained through fast updates of consecutive 3D image sets, has been introduced recently. The current highest frame rates range from 4 to 16 3D volume acquisitions per second, depending on the probe and the field size. This technique is available for grey-scale imaging as well as colour and power Doppler modes. This allows, for example, the visualization of fetal movements, the display of intravascular haemodynamics, a fly-through perspective from inside the common carotid looking toward the carotid bifurcation or a better guidance of the progression of a biopsy needle towards the target.

Elastographic imaging

Elastographic imaging was developed to objectively quantify the pathological changes related to the presence of abnormal tissue compared with surrounding tissue [34, 35, 36, 37, 38]. Manual palpation or subjective ultrasound palpation have been used for many years to provide a qualitative assessment of low-frequency tissue stiffness, but the need for direct imaging implies the development of ultrasound elastographic imaging.

The mechanical properties of soft tissues which depend on their content include elastic moduli (Young's modulus), the Poisson's ratio and the shear properties of tissue. These properties are totally different from the bulk elastic modulus that governs propagation of ultrasonic waves. For example, normal breast tissue is less stiff than that of fibroadenoma, and breast cancers showed shear moduli up to seven times higher than those of normal tissue.

The potential for imaging this high contrast has led to the development of ultrasound methods. For this different approaches have been developed:

1. The application of low-frequency vibrational energy and concomitant Doppler detection of wave perturbation
2. Imaging of local response to an applied load with direct estimation of parameters by comparison between pre- and post-compression data
3. Calculation of these parameters

Manual compression (freehand elasticity imaging) has been used to provide a flexible and real-time exploration of different tissues such as the breast. The acquisition of RF signals during compression allows the strain of the tissue, which is related to the stiffness of the tissue, to be imaged. This imaging method has been applied to different organs such as the breast, prostate, kidney, blood vessels and liver with promising results for differentiation between benign and malignant lesions or in the assessment of tissue changes following high-intensity focused ultrasound (Fig. 17).

Miniaturization of units and probes, transfer and archiving of US data

Miniaturization and introduction of portable units

New technology is now enabling the miniaturization of high-performance ultrasound equipment with the capability of doing complete exams; these include handheld ultrasound units, which have recently been introduced into the market by several companies [39]. The design, weight, battery life, choice of transducers, capabilities and modes vary considerably between units. In one system the ultrasound processing is incorporated into the probe, which is simply hooked into a personal computer or a laptop computer (Terason). Although a study of prototype units revealed a slightly lower display quality than larger ultrasound units, these hand-held units seem to be very promising, targeting bedside and intensive care unit markets.

Miniaturization is always very active in featuring new transducers such as endovascular, laparoscopic and trans-oesophageal probes.

Transfer and archiving of US data

Digital imaging has allowed consideration of ultrasound as any other imaging modality, and presently sonographic data can be easily archived and transferred by most PACS systems as CT and MRI images are. On the other hand, the worklist and/or the RIS can be easily integrated into ultrasound units.

The most recent release of the DICOM specification has made these advances easier. For example, this standard now integrates colour imaging; however, sound is still not available, which is a problem for Doppler mode.

Grey-scale and colour frames are now currently archived as DICOM data, which should prevent from compatibility problems when using units from different companies. This also includes image displays, specialized services and quantification. Ultrasound frames can be compressed, mainly using the JPEG mode. Dynamic clips can be transferred as frames series. Ultrasound data compression is mandatory to significantly reduce the volume of data storage, which might very rapidly increase. Compression modes include mainly MJPEG, MPEG and Wavelet, and allow preservation of the dynamic information.

Conclusion

In addition to the above-described advances, including many other research fields with promising developments, especially the direct use of RF data, arterial wall elasticity analysis, therapeutic aspects of ultrasound (thermal effect, targeted microbubbles), radiologists must keep aware of these new developments, which could open new diagnostic horizons and might be substitutions for radiating and high-cost imaging modalities.

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