Proceedings of the 2nd EAA International
Symposium on Hydroacoustics
24-27 May 1999, Gdańsk-Jurata
POLAND

Recent Advances in Medical Ultrasound

Andrzej Nowicki

Institute of Fundamental Technological Research - Polish Academy of Sciences,
Świętokrzyska 21, 00-049 Warsaw, Poland
e-mail: anowicki@ippt.gov.pl

Imaging and Doppler ultrasounds creatively employing recent radar and sonar technologies are rapidly growing with numerous new exciting developments. Velocity sensing techniques can be as simple as C.W. Doppler up to the multi-channel MTI. High density, multi-elements linear array, phase array and annular array transducers benefit from the synthetic aperture and open a new era of a real time scanning of the biological structures with unprecedented quality. Flow velocity color mapping, power Doppler, tissue Doppler, contrast agents echo enhancement and other new, exciting developments will be discussed.

1. Introduction

Advances in ultrasonic equipment dedicated for medical diagnosis is closely related to the development of radar and sonar theory and technology. Historically, with a certain time lag ultrasounds were employing initially the A-mode, through M-mode to the recent sophisticated B-mode presentations with bearing indicated by one co-ordinate and distance by the other co-ordinate. Although the medical imaging ultrasounds rarely use the driving frequencies exceeding 10 MHz (comparing to GHz range of a typical airborne radar) the resulting axial resolution is similar to that of radar because of the difference of the order 10^5 in velocity of electromagnetic waves in air and sound in biological tissue the wavelength of both waves in the medium is comparable. Depending on the application the velocity can be simply measured using C.W. Doppler with no range resolution along the ultrasonic beam or in a MTI type processing with real time spectral analysis and color flow mapping of the moving targets superimposed on the grey scale images of the anatomical structures.

Real time ultrasonic imaging systems have been available for more than thirty years. Much has occurred in transformation of the basic architecture and applications of these imaging systems. The main developments were observed in the performance of scanning probes due to the quality of beamformers which are probably the most important components of the modern scanners, fabrications of wide band piezoelectric transducers and on-line image processing. The major technological advance pushing forward the whole ultrasound imaging was the introduction of the grey scale with up to 512 grey levels in commercially available instruments with digital scan converters. The advent of fast beam displacement and digital technology revolutionized B-mode allowing real-time imaging of all soft tissue organs with the perfect reproduction of their instantaneous movement. Frame-rates exceeding 100/s are not that uncommon now-days. Important field of ultrasonic diagnostic belongs to Doppler blood flow measurements and color flow mapping CFM.

2. Beam steering

Similarly to radar and sonar technologies three different scanning techniques were developed: 1. mechanical sector scanning - circular piston type

transducers are immersed in liquid and are mechanically wobbled thus the beam produced by the transducer is sectorially displaced with field of view of 30°-45°, 2. linear arrays - contain from 64 to almost 200 small individual rectangular transducers excited sequentially with short electrical pulses [2], 3. phased arrays - are more complicated than linear arrays. In general all individual elements of the transducers are smaller and are fired together with the pre-programmed time delay.

The basic functions of beamforming include: 1.generation, timing and apodization during transmission, 2. controlling of the time delays and signal processing during receiving, 3. apodization and summing of delayed received echoes. In general beamformers are to create the narrow and uniform (with minimal sidelobes) ultrasonic beams. Focusing and linear or sector steering are accomplished through the appropriate time delays between pulses exciting the individual elements of the single elements in the linear or phase array transducer.

The number of elements used in transmit mode depends on the selected focus range. Only one transmit focus can be selected for each transmitted pulse thus when more focal zones are selected then the overall frame rate decrease proportionally.

Fig.1. shows the simulated beam shapes for different steering angles [15]. The overall length of the arrays is D=20 mm radiating pulses with a center frequency of 3 MHz . Since the wavelength is equal to λ=0.5 mm the total aperture was 40 wavelength wide. The near field was then roughly equal to be 20 cm covering the diagnostic range for scanning heart and obstetrics. The beams were calculated for 64 elements array of size 0.625 λ=0.3125 mm (Fig.1a and b) and for 32 elements array of size 1.25 λ=0.625 mm (Fig.1.c). Beam shapes focused at the range of 8 cm (F=160 λ for 3 MHz). The beams are deflected by 0° and 30°. The beam width for the focused beams converge sharply in the focus and next diverge beyond the focal point. The amplitude is maximum in the focus. For 32 elements array the side lobes are visible at deflection of 30°. The individual element size has also a great influence on the overall beam shape and intensity. Similarly to photography the ultrasonic F-number is defined as a ratio of focal distance and the aperture of the array. For an aperture of 8 cm and transmit focus at 16 cm range an F-number of 2 is obtained.

The spatial resolution in homogenous media increases with decreasing F-number. As was said the multiple focal zones can be obtain at the expense of the frame rate. Thus a great deal of effort is now made to produce multiple transmit at sequential focal locations

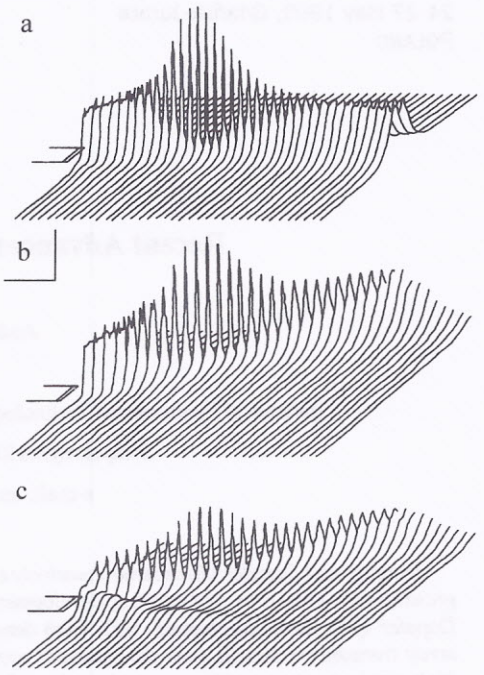

Fig.1. Beam pattern for 40 λ aperture array with 64 elements of 0.625λ width (a and b). focused at the range of 160λ deflected by 0° (a) and 30° (b and c).

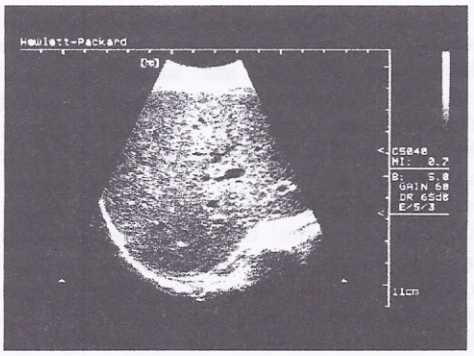

Fig.2. Sonographic image of liver and hepatic veins obtained with 5 MHz convex array.

The additional improvements in the image quality are achieved introducing 1.5D arrays - multi-row arrays with electronically controlled focusing in the elevation but with delay symmetry along the center line.

Another field of transducer design is dedicated to limited diffraction beam forming. [8]. The transmit and receive apodization (using smaller number or weighted beams) in the very near field is

also introduced. Another important image processing includes deconvolution of transmit beam patterns [6]. Introducing of originally radar developed synthetic aperture is also under the extensive study. With synthetic aperture receive processing the better performance might be achieved without actually using all real receive channels.

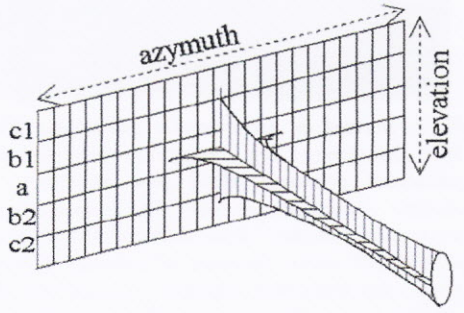

Fig.3. 1.5 D transducer array (5 rows of individual piezoelements in elevation).

3. Color Flow Mapping.

The relevant information gained with P.W. Spectral Doppler is limited to the small sample volume in the scan plane. Multigate Doppler incorporated into the 2D image provides the information on the flow distribution along a single scan line. When color-coding is added to the output of the correlator frequency estimator, instead of the flow velocity profile display a new type of presentation is gained. The hue and brightness of red and blue color are proportional to the velocity respectively towards and away from the transducer. This means also that the displayed color depends on the orientation of the blood vessels. The angle dependence as well as aliasing manifesting in rapid changes of color from bright red (yellow) to light blue or vice versa are the major drawbacks of the CFM Doppler shift extends over Nyquist sampling frequency range. Strictly speaking these are the velocity components along the sector line which are coded into the color. When beam/velocity angle changes during one scan when relating color does not necessarily reflects the actual velocity. However, observing the displacement of blood streams, e.g. in heart chambers the proper correction of the actual angle can be made enabling the exact measurements of maximum flow velocities and next the estimation of the pressure gradients.

The turbulent flow manifests itself by increased variance of the Doppler spectra and is coded in green. Color information in CFM is superimposed on standard grey scale 2D echo images with the priority for echoes signal rather than for flow signals.

CFM basic operation. - Over 100 serial gates are distributed along each scan line. Due to the echoes from stationary targets (e.g. vessel walls) being much larger than the echoes backscattered from blood the former must be suppressed in order to extract the correct estimation of the Doppler shift.

Stationary echoes cancellation was introduced to the medical ultrasound in the eighties [11, 12] and since then it has been applied to heart flow imaging, arterial and venous flow in peripheral circulation and recently for detection of local perfusion, tumor detection (localization of increased vascularization) and myocardial gradients. The suppression of clutter from stationary and slowly moving tissues is one of the major issues in color Doppler. If not sufficiently removed, clutter can destructively affect the ability of color Doppler imaging to estimate the Doppler mean shift. Because of the necessary speed and very short settling time second order IIR high pass are used. The preferred characteristics are of Chebyshev or elliptic ones with a very low cut-off frequency below $f_{prf}/20$ [17].

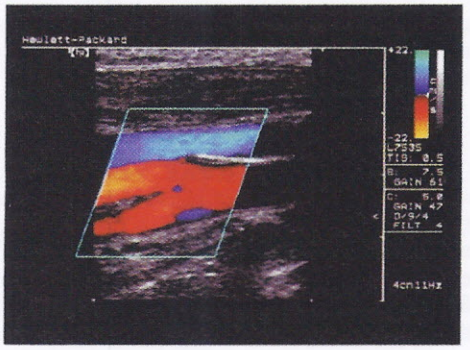

Fig.4. Color flow map of the image of carotid artery bifurcation (red) showing flow dynamics and jugular vein (blue). 7.5 MHz linear array.

The mean Doppler shift detection in most of the pulse Doppler color imagers is based on the autocorrelation algorithm. One of the problem with employing the mean as a parameter of choice is that random noise in the ultrasonic signal is similar to gaussian distribution of Doppler backscattered signal from the red cells. It obviously corrupts the Doppler output. Aliasing is distorting the directional and speed information as well. A novel color Doppler technique encodes power of the backscattered signal coding its intensity in color. That greatly improves system sensitivity and delectability of low velocity tissue flow. A disadvantage of power imaging is that it provides no

information about the speed or direction of flow but is angle independent and free of aliasing artefacts.

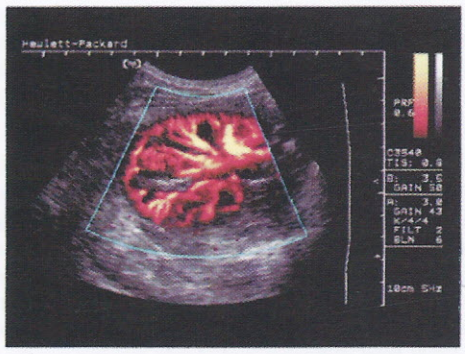

Fig.5. Renal flow perfusion displayed in power Doppler mode.

There is another new imaging modality based on the color Doppler - Doppler Tissue Imaging (DTI) provides the information on the velocity and acceleration of the tissue structures mostly used for myocardial tissue motion imaging [9, 13, 16]. With normal physiological contraction, the velocity of the endocardium is higher than that of the epicardium, and the estimation of the resulting velocity gradient across the heart wall can be of a great diagnostic value, especially in differentiation of the active heart wall regions from the passive one.

4. Intravascular Ultrasound

The developments of h.f. ultrasound and micromechanics stimulated a new era of catheter type intravascular imaging. The rotating transducers or electronic arrays of the overall external diameter not exceeding 1 mm were fabricated by several laboratories helping to obtain the astonishing images of the internal lumen of eg.coronary artery. Combined wit Doppler these system help to evaluate true coronary volumetric blood flow.

5. High frequency imaging

Microsonography at 30 MHz to 50 MHz adds new capabilities to the application of ultrasound in medicine. Microsonografic information about the skin at the penetration depth of a few millimeters can offer a non invasive diagnosis means for identifying skin cancers in incipient stage. It can be also used in the dermatology for diagnosis of skin diseases, for monitoring the process of lesions treatment [10].

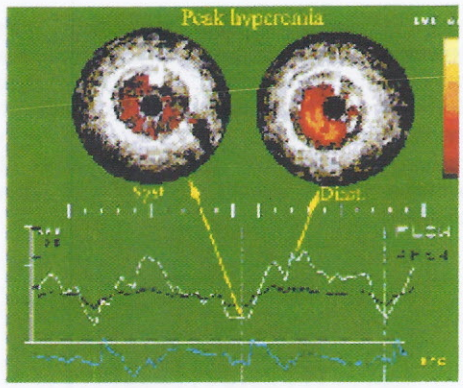

Fig.7. Normal an intravascular ultrasound (Du-Med, EndoSonics Corp.) image (top right) with the calculated coronary blood flow velocities in diastole superimposed over the cross-sectional area, color scheme - from dark-red (10 cm/s) to yellow (100 cm/s). Decrease of coronary flow in systole (top left)-lower velocities recorded over the arterial lumen. The calculated flow is plotted (in white) below, together with the instantaneous peak Doppler velocity (DOP - in blue). (Courtesy of Stéphane G. Carlier, Catheterization Laboratory, Thoraxcenter, Erasmus University, Rotterdam, The Netherlands).

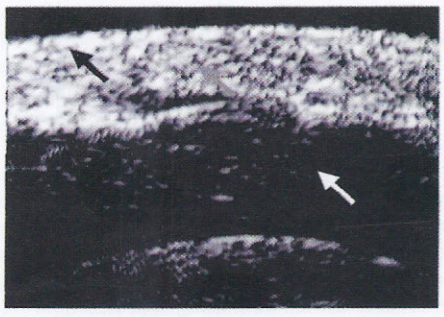

Fig.6. Ultrasound scan of healthy skin. Epidermal entry echo (black arrow), dermis (white arrow) subcutaneous tissue (grey arrow).

6. Elastography

Among the recent developments in ultrasonic tissue characterization the important place is reserved for elastography. The r.f. data from tissue ROI is acquired in two steps, before and after the compression. The tissue elastic signature is than examined and imaged comparing the a pair of precompression and postcompression r.f. echoes. As a result of processing (correlation) two sets of data, the local change of displacement of the tissue is computed. Knowing the value of the external force the estimate of the strain can be calculated.

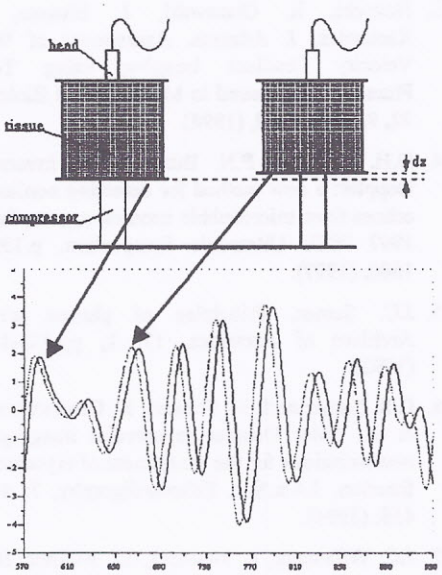

Fig.7. Echo-line from normal and compressed tissue are shifted in time, the time shift is proportional to the local strain.

7. Ultrasonic contrast agents

The recent hot topic in ultrasonography is echoes enhancing using so-called contrast agents. The evolution of ultrasonic contrast agents has progressed from naturally occurring gas bubbles in physiologic saline, indocyanine-green 1-5% [7] or rapidly i.v. injected glucose solution (any liquid injected through a small-bore needle would produce a contrast effect – cavitation at the needle (catheter) tip) to the developments of standarized agents. Some of these agents are widely approved like microcapsulated gas bubbles with the mean population diameter of few microns. Small encapsulated gas bubbles in a contrast medium react on an external oscillating pressure field with volume pulsation exhibiting a resonant behavior as a result of stiffness and inertia. The stiffness is that of the enclosed volume of gas which acts like a spring when the bubble is disturbed from its equilibrium radius. The inertia is principally due to the mass of the liquid surrounding the bubble, which oscillates with it. This effect is important because a resonating bubble effectively yields an increase in scattering cross-section which results in 20 to 30 dB increase of the backscattered ultrasound energy and thus greatly enhances the sensitivity of color imaging of weakly perfused organs. An approximation of the resonant frequency f_0 is given by Anderson and Hampton [1]

$$f_0 = \frac{1}{2\pi r}\sqrt{\frac{3\gamma P_0}{\rho_0}}$$

where γ denotes the adiabatic ideal gas constant, P_0 is the ambient fluid pressure and ρ_0 is the density of fluid, r is radius of the bubble

Depending on the magnitude of the ultrasound wave the pulsation will be:

1. Linear to the applied pressure

   linear scattering 0-50 KPascal

2. Non-linear to the applied pressure – the spectrum of the scattered ultrasound wave contains also higher harmonics of the ground frequency.

   harmonic scattering 50 -200 KPascal

3. Stationary and transient scattering

   transient power scattering 0.2 - 2 MPascal

The major improvements of ultrasound contrast agents were performed in the early nineties [3, 4, 5]. Some manufactures of ultrasonographs started to develop special software for contrast imaging and second harmonic imaging including pulse inversion [14]. In the second harmonics imaging the basic principle is that a narrow-band pulse is transmitted with a center-frequency of 2 to 3 MHz and the receiving filter has a center frequency of 4 or 6 MHz respectively.

Several agents are currently approved for clinical use, among them:

Echovist - (Schering AG, Berlin, Germany), granules consisting of agglomerated galactose microparticles mixed with galactose solution, *97% below 7 μm*. Opacifies the right heartside after intravenous injection, it does not cross the capillary circulation of the lungs.

Albunex - (Molecular Biosystems Inc. San Diego, USA) Transpulmonary echo contrast.

Levovist - (Schering AG, Berlin, Germany), transpulmonary echo contrast, based on galactose microparticles as a vehicle to deliver microbubbles of defined size. It contains palmitic acid, which acts as a surfactant, and forms a thin coat around microbubbles. This fatty acid not only enables the bubbles to survive the passage through the lungs but also provides a high degree of stabilization in the arterial vascular bed. Depending on the concentration used (200-300 mg/ml) signal enhancement of up to 25 dB is achieved. The duration up to 5 minutes.

EchoGen - (Sonus Pharmaceuticals, Bothell, USA) - a liquid perfluoro compound with boiling points below the body temperature. After intravenous injection the liquid starts boiling within the blood

and echogenic microbubbles are formed while liquid evaporates

References

1. A.I. Anderson, I.D.Hampton, Acoustics of gas-bearing sediments, J.Acoust.Soc.Am., 67, pp.1865-1889, (1980).
2. N. Bom, C.T. Lancee, F. van Zwieten, E. Kloster, J.R. Roelandt, Multiscan echocardigraphy, In: Technical description, Circulation, 48, pp.1066, (1973).
3. P.H. Chang and K.K. Shung, Second harmonic imaging and harmonic Doppler measurements with Albunex., IEEE Transactions of Ultrasonic, Ferroelectrics and Frequency Control UFFC, pp. 1020-1027, (1995).
4. C.C. Church, The effects of an elastic solid surface layer on the radial pulsation of gas bubbles. J. Acoust. Soc. Am. 97(3), (1995).
5. N. De Jong N, F.J. Ten Cate, C.T. Lancee, J.R. Roelandt, N. Born, Principles and recent developments in ultrasound contrast agents. Ultrasonics 29: pp.324-330, (1991).
6. S. Freeman, P.C. Li, M. O'Donnell, Retrospective dynamic transmit focusing, Ultrasonic Imaging, 17, pp.173-196, (1996).
7. R. Gramiak R.and P.M. Shah, Echocardiography of the aortic root. Invest.Radiol., 3, pp.356-366, (1968).
8. J.Y. Lu, H. Zou, J.F. Greenleaf, Biomedical ultrasound beam forming, Ultrasound in Medicine and Biology, 20, 5, pp.403-428, (1994).
9. W.N. McDicken, G.R. Sutherland, C.R. Moran, L. Gordon, Color Doppler velocity imaging of the myocardium, Ultrasound in Med. and Biology, 18: 651-654; (1992).
10. K. Mlosek, A. Nowicki, E. Szymańska, W. Jakubowski, W. Secomski, P. Karłowicz, J. Litniewski, M. Lewandowski, Ultrasound system for microsonography of the skin: instrumentation and clinical application (in polish), Ultrasonografia, 2, 67-76, (1998).
11. Nowicki, J.M. Reid, An infinite gate pulse Doppler, Ultrasound in medicine and Biology, 7, pp.41-50, (1981).
12. Nowicki, J.M. Reid, P. Pedersen, A.W. Schmidt H. Oung, On the behavior of instantaneous frequency estimators implemented on Doppler flow imagers, Ultrasound in Medicine and Biology, vol.16, pp.511-518, (1990).
13. Nowicki, R. Olszewski, J. Etienne, P. Karłowicz, J. Adamus, Assessment of Wall Velocity Gradient Imaging using Test-Phantom, Ultrasound in Medicine and Biology, 22, 9, 1255-1260, (1996).
14. D.H. Simpson, P.N. Burns, Pulse inversion Doppler: a new method for detecting nonlinear echoes from microbubble contrast agents, Proc. 1997 IEEE Ultrasonic Symposium, p.1597-1600, (1997).
15. J.C. Somer, Principles of phased array, Archives of Acoustics, 17, 1, pp.133-150, (1992).
16. G.R. Sutherland, M. Stewart, K. Groundstroem et al., Color Doppler myocardial imaging: a new technique for the assessment of myocardial function, J.Am.Soc. Echocardigraphy, 7: 441-458; (1994).
17. J.C. Willemetz, A. Nowicki, J.J. Meister, Bias and variance on the estimate of the Doppler frequency induced by wall motion filter, Ultrasound Imaging, vol.11, pp.215-225, (1989).