PRINCIPLES AND TECHNIQUES OF ACOUSTICAL IMAGING

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INTRODUCTION

The ability to get good images when using ultrasound as a source of data is due to a certain number of suitable conditions which are found in biological tissues. Most of the tissues, other than bones, have a density ρ approaching that of water, and the propagation speed of the ultrasound waves c in this media is practically constant ($\simeq$ 1540 m/s), so that accurate depth ranging by echo delay time is achievable in pulse echo systems. Moreover, the relative uniformity of the acoustical impedance ($Z = \rho c$) of the tissues to be explored explains why it is possible to perform an exploration in depth. Low-level reflections occur between different soft tissues ($\Delta Z << \overline{Z}$), so that most of the acoustic energy is transmitted through the different interfaces and it is available for imaging deeper structures. The depth of penetration is mostly limited by the absorption of ultrasound waves which increases with frequency, and not so much by the number of interfaces which may be high. Frequency of the order of 3 MHz are generally used to achieve exploration over 20 cm.

I. THEORY OF IMAGING

Development of imaging techniques involve to choose some model for the propagation of ultrasound in a 3 D biological structure. The acoustical properties of tissues enable us to adopt as a simple model, the one of the linear acoustics in a fluid. The acoustic field is derived then from a potential Φ which obeys for a homogeneous medium, and when neglecting absorption, the wave equation :

$$\Box \Phi = 0 \qquad \text{where} \quad \Box = \Delta - \frac{1}{c^2} \frac{\partial^2}{\partial t^2} \tag{1}$$

The presence of non homogeneous properties associated with the explored tissues will modify this wave equation by introducing an ensemble of source terms which are a function of the local variations of density, compressibility and attenuation and which may be considered as an ensemble of scattering centers.

$$\Box \Phi = S (\Phi, \delta\rho, \delta\chi, \delta\alpha) \tag{2}$$

The purpose of acoustical imaging is to reconstruct the map of all these scattering centers with the best accuracy. This is why we choose to submit the object to be exa-

mined, to a known acoustic beam $\Phi^\circ(\vec{r},t)$, and from the acoustic field $\Phi(\vec{r},t)$ that we observe, we try to extract the source terms S.

If we accept the slight disturbing characteristic of the non homogeneous properties presented to the incident field Φ° by the tissues submitted to the beam, it is possible to simplify the equation (2) by keeping in the development in the source terms with respect to Φ° only the term of order 0.

$$\square\ \Phi = S\ (\ \Phi = \Phi^\circ,\ \delta\rho,\ \delta\chi,\ \delta\alpha) \tag{3}$$

This classical approximation of a one diffusion process [1] allows then to completely separate the effect of different sources. The ensemble of scattering centers creates a potential Φ^1 which, added to the irradiation potential Φ° , will give the value of the observed potential $\Phi = \Phi^\circ + \Phi^1 \quad (\Phi^1 << \Phi^\circ)$

$$\square\ \Phi^\circ = 0 \tag{4'}$$

$$\square\ \Phi^1 = S\ (\ \Phi^\circ,\ \delta\rho,\ \delta\chi,\ \delta\alpha) \ = s\ (\vec{r},t) \tag{4"}$$

Such an approximation which is equivalent to disregarding the effect of the non homogeneous properties on the potential diffracted by each source is certainly inaccurate in the areas containing some bone structure or air pockets because their impedance is extremely different from the impedance of most of the biological tissues. However, it is a very convenient hypothesis to the extent that the desired acoustic data can be acquired outside the irradiated domain in terms of the acoustic field Φ^1 , and this, through a linear equation (4") which allows one to find a solution under the form of an equation of convolution because of its invariance properties (in a free space) :

$$\Phi^1\ (\vec{r},t) = s\ (\vec{r},t)\ \circledast\ g(\vec{r},t) = \iint s(\vec{r}_\circ,t_\circ)g(\vec{r}t|\vec{r}_\circ t_\circ)d\vec{r}_\circ dt_\circ \tag{5}$$

The effect of each point source can be made evident through the term $g(rt\,|\,r_\circ t_\circ)$: Green function associated with the equation (4") which is nothing else than the perturbation Φ^1 created by a point source located in $\vec{r}_\circ$ excited by an impulse $\delta(t_\circ)$. The theory of diffraction [1] teaches us that in a free space this function can be written under the form :

$$g(\vec{r}t|\vec{r}_\circ t_\circ) = \frac{1}{4\Pi R}\ \delta\ (t - t_\circ - \frac{R}{c}) \quad \text{where } R = |\vec{r} - \vec{r}_\circ| \tag{6}$$

This is the classic choice of the solution of delayed potential of equation (4") which expresses the fact that each point source creates a perturbation which travels at a speed c under the form of a diverging wave front.

The seeking of the reverse convolution operation which will allow to recover the ensemble of the source terms $s(\vec{r},t)$ is going to be made easier by the two following remarks :

a) on the one hand, it is not necessary to know $\phi^1(\vec{r},t)$ in the total space : the Huygens principle and its various formulation [1] teaches us that the value of the field ϕ on a given plane π located outside the source area is sufficient to determine in all points in the half space $z > z_\pi$. Such a fact will allow us to only make a two-dimensional spatial study, limited to the exploration of a single plane π :

$$s(\vec{r},t) = \phi^1_{\vec{r}\in\pi}(\vec{r},t) \otimes g^{-1}(\vec{r},t) \qquad (7)$$

We shall see later [II-III] the fundamental part played by the location of this reception plane relative to the irradiation source ϕ° (transparence imaging or echography).

b) On the other hand, the operation consisting of going back to the different source terms, starting from the field $\phi^1(\vec{r},t)$ observed on the plane π, can be seen as a true reversal of the time progression, which would bring back each of the diverging wave fronts to its initial source ; utopian operation that can be replaced rightly with a reconstruction of $s(\vec{r},t)$ in delayed time, because only the spatial relationship of the source terms is of interest to us. This operation of deconvolution which must associate to the diverging wave front coming from a source located in $\vec{r}_\circ$ the distribution $\delta(\vec{r}_\circ)$ can then be understood as a parallel processing of the data, similar to the one that is sequentially performed by scanning a plane π with a spherical ultrasonic transducer (for example, a spherical cup of piezoelectric ceramic) of "infinite" aperture and of radius $z_\circ - z$ ($z_\circ$ and z are the depth coordinate of the source and of the plane. This type of transducer, because of its spherical geometry, delivers an infinite signal every time that a source located at the distance $z_\circ - z$ aligns itself on its axis. Mathematically, this spherical geometry can be considered as equivalent to a spherical correction of delay associated to a plane transducer. Such a correction of delay can be written under the form of a convolution by a distribution $g^{-1}(\vec{r},t)$, which is equal to

$$g^{-1}(\vec{r},t) \propto \delta\left(t - T + \frac{R}{c}\right) \qquad (8)$$

where T is a time delay required to provide a causal characteristic to the function $g^{-1}(\vec{r},t)$.

The acoustic image of the object under study is then directly related to the spatial portion of the function $s(\vec{r}',t)$ which is equal to :

$$s(\vec{r}',t) = \iint_{\vec{r}\in\pi} \phi^1(\vec{r},t - T + \frac{|\vec{r} - \vec{r}'|}{c})\, d\vec{r} = \phi^1(\vec{r}',t) \otimes g^{-1}(\vec{r}',t) \qquad (9)$$

A parallel processing of this kind, although being worth considering at first, once the maping of the field $\phi^1(\vec{r},t) = \phi^1(x,y,t)$ is obtained by means of numeric processes, is however excessively cumbersome. On the one hand, the amount of data to store is tremendous, that is to say : a three dimension table (two for the space and one

for the time) where the number of data by dimension is larger than a hundred ; the frequency of the temporal sampling and the pitch of the spatial sampling being tied to the duration and the shape of the irradiation pulse. On the other hand, the numerical decoding of such a vast quantity of data is still an operation much too cumbersome, which could not be conducted and still maintain real time.

It is possible to consider replacing the digital processing with a purely analogic processing which would be performed in real time, by means of acoustic lenses. It is well known that a lens of this type is able to perform in real time the desired decoding operation. Its behavior is one of a time delay device able to reverse the order of arrival of the data conveyed by the various wave fronts coming from the object. It transforms, in this way, the diverging wave coming from a point source in a converging wave focusing at the point image. This stigmatism allows us to produce, in a natural way, the image of the object, plane by plane. However, this kind of processing is limited because of the difficulties encountered in the manufacturing of either quality ultrasound retinas or wide aperture acoustic lenses. The most successful lens imaging system is the Green camera [2] which uses a sophisticated acoustical lens and a system of rotative prisms, which permits a fast mechanical scanning of each plane of the object (10 frames per second) in front of a linear array of 192 ultrasonic transducers.

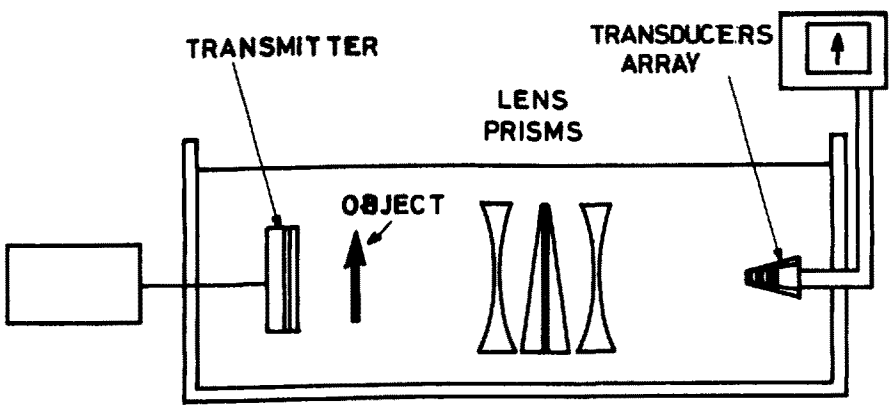

Fig. 1 - Real time acoustical imaging with a linear array and mechanical scanning of the focused image using rotative prisms

Besides these classical imaging methods, different attitudes have been retained in the field of acoustical imaging.

- The first attitude, originating from the spectacular results obtained in coherent optics, consists of illuminating the entire object with a monochromatic signal. This is the domain of Acoustical Holography.
- A second attitude consists of coming back to the use of brief signals and working

in echography, but replacing a global illumination of the object with a more selective illumination by only illuminating one plane, one line, or even only one point of the object. This increase in selectivity produces a better image quality but at the same time increases the image acquisition time.

II. ACOUSTICAL HOLOGRAPHY

The visible advantage of a monochromatic illumination is primarily found in its invariance properties which entail that in each observation point, the same monochromatic structure is found, of which the complete knowledge then requires only two parameters : the amplitude A and the phase ψ , which, because of the linearity of piezoelectric transducers, can be easily obtained.
It must be noted that in monochromatic lighting, the temporal information associated to each point source, is then replaced by a simple phase information. The spatio-temporal form $g(\vec{r}t|\vec{r}_o t_o)$ (6) which was associated with each source in pulsed mode is reduced here to the spatial form :

$$\frac{\exp\left(ik\left|\vec{r}-\vec{r}_o\right|\right)}{4\pi\left|\vec{r}-\vec{r}_o\right|} \qquad (10)$$

The recognition of each of the point sources is then made through the signature in phase of the perturbation that it retransmits, and the deconvolution process requires now only to work in the spatial domain (which was not the case in (9)). In the case of Fresnel approximation, such a deconvolution process is nothing more than a Fresnel transform which may be realized by various methods.
Some are purely analogic, which after having transferred the phase and amplitude information of the acoustic waves under a form permitting their detection by means of an optical coherent beam, are using optical decoding processes [3,4,5,6] .
Among the various solutions offered, mention must be done of the levitation effect of a liquid gaz interface due to the ultrasonic radiation pressure (Fig. 2). The optical reconstruction performed in real time with the diffraction of a laser beam by the distorded interface, realizes the required Fresnel transform. This decoding process is however degraded by some important aberrations due to the large difference between the optical wavelength and the acoustical wavelength. Some other decoding methods, purely digital, associated to the exploration of the detection plane by one or more transducers, allows us to avoid these aberrations at the expense of the data acquisition time of these images [7,8] .
It must be understood that until now, even if some good images of simple test patterns have been obtained, the results obtained when observing "in vivo" organs have been disappointing, despite the use of reconstruction processes apparently ideal (large aperture, sufficient sampling, high dynamics).

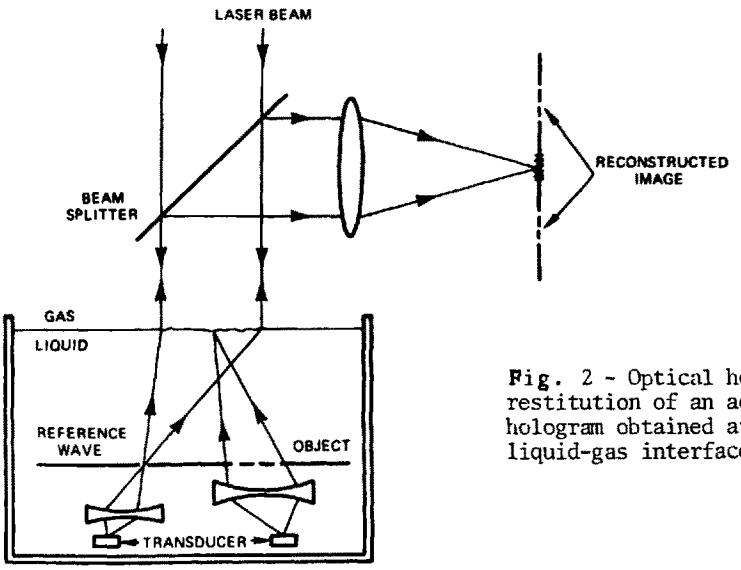

Fig. 2 - Optical holographic restitution of an acoustical hologram obtained at the liquid-gas interface.

This constatation can, in fact, be explained by noticing that in acoustical holography, the over all illumination of the object implies that the recognition of each of the point sources will be made in the presence of an important noise coming from the ensemble of the other points of the object. This interfering noise (speckle noise) decreases in practise the resolution power in a drastic way especially when the studied object contains many scattering centers. Besides the use of a monachromatic signal which can only provide locally two informations (A and ψ) no matter what the complexity of the object to be studied, is not a good choice for a source illumination when studying organs of complex structure. The decoding of the coordinates of each of the sources in the presence of speckle noise is made more efficent when the "signature" of each of the sources on the detection plane carries more information. By enlarging the detection aperture, one can hope to gain access to a larger quantity of information. However, in acoustics, the difficulties encountered to manufacture ultrasound retinas of large dimension limit, in fact, the apertures used, to some dimensions of the order of a few hundred wavelength. In the best cases, the $2N^2$ data acquired in acoustical holography, by means of a matrix retina sampled by N^2 transducers (where N as of today is no greater than 300 [9]) are not enough to reconstruct with accuracy the three dimensions of an object of reasonable size. It must be noted here that this problem is not as acute in optics where the apertures used contain a much larger number of wavelength ($\simeq 5.10^4 \times 5.10^4 \ \lambda^2$) which allows us to obtain much richer information on some objects that, it must be pointed out, are not truly three dimensions objects. In optics, in most cases, only the surface of the objects diffuses the light, and the information that it delivers and that we try to decode, as only in fact a bi-dimensional characteristic ; whereas in acoustics, the relative transparency of organs

to ultrasound beams of a few MHz makes them true tri-dimensional objects.
In front of these findings, the use of a temporal coding richer for illumination appears to be required in acoustics. Besides the "radar type" signals having a high compression ratio ($BT \gg 1$, B being the bandwidth and T the signal duration) which is able to provide a large quantity of data, but which requires some very cumbersome temporal decoding techniques, it has become evident that the easiest way to enrich the acoustical information was to come back to the use of brief illumination signals, while working then in echography.

III. ULTRASOUND ECHOGRAPHY

To work in the echographic mode consists in choosing for a reception plane, a plane identical to the transmission plane, while using a brief signal for illuminating the object. One gains additional information to locate the different scattering centers, taking into consideration the arrival time of each of the wave fronts diffused by each of these sources.
For an infinitely short duration signal emitted at the time $t = 0$, the signature of a given target located in $\vec{r}_o\ (z_o, \vec{m}_o)$ at the level of the reception plane ($z = 0$) varies except for an amplitude factor according to :

$$\delta\ (\ t - z_o/c - |\ \vec{r} - \vec{r}_o\ |\ /c) \qquad (11)$$

An expression that can be made simpler by considering it in a Fresnel paraxial approximation according to :

$$\delta\ (t - 2z_o/c - |\ \vec{m} - \vec{m}_o\ |^2/2z_o c) \qquad (12)$$

and which explicitely demonstrates that it is possible to go back to the coordinates of a point source not only by means of the radius and the curvature center of the wave front that is associated with it ($\rho = 2z_o c, \vec{m}_o$), but also by taking into account the arrival time of the beginning of this wave front at the level of the reception plane ($\tau_o = 2z_o/c$), time which is dependant, in an univocal way, of the depth of the target. This information is to be compared to that obtained in transmission imaging where for a fixed distance between the transmitter and the receiver equal to L, each of the sources is observed through an expression of the form

$$\delta(t - L/c - |\vec{m} - \vec{m}_o|^2\ /2z_o c) \qquad (13)$$

where the arrival time L/c of the various wave fronts on the reception aperture is independant from z_o. It is an information not as rich as in echography, which only gives the distance z_o through the wave front curvature. The signals coming from the different scattering centers arrive together at the reception plane and introduce a

speckle noise very undesirable for the decoding process. In echography however, the natural decoding of the z coordinate brings an important decrease of this noise, which is coming from a slice of the object and which is the thinnest when the illumination signal is the shortest. In fact, experimentally, the illumination signal $i(t)$ is far from being similar to the infinitely short duration of the illuminating signal $\delta(t)$ and the signature of the target at the level of the reception plane must be considered as being the convolution of the distribution (12) with $i(t)$.

$$i\ (t - 2z_o/c - |\vec{m} - \vec{m}_o|^2 /2z_o c) \qquad (14)$$

where : $i(t)$ has a form and a duration directly in relation to the characteristics of the emitting source. It is possible to obtain some illumination time in the order of the microsecond, when one uses well damped transducers ; this compared to the 200 μ s required for exploring a depth of 15 cm gives an idea of the amount of data that can be obtained.

It must be well understood that the possibility of using the echographic information in order to visualize small dimension structure ($\simeq$ mm) is a specific acoustical property. It is the very slow velocity of the ultrasound, as compared to the electro-magnetic waves, that permits us to realize without difficulties the separation of echos coming from very close targets (1,3 μ s for a distance of 1 mm between two targets).

A - "Parallel" echographic processing.

The choice of an echographic exploration mode, although it appears desirable, brings in acoustics a certain number of problems. The fundamental problem is, of course, related to the decoding operation that must be performed in order to reconstruct with accuracy the mapping of the scattering centers. If one wants to take advantage of the natural decoding of the coordinate z_o, brought forth by the echography, real time decoding processors must be developped that perform a deconvolution operation at the same time the echos are coming back, by means of a time delay nucleus (8) which is adapted at all times to the cuvature of the observed wave fronts. A decoding process of this type can be perceived as a real "zoom effect" that will be accomplished for example by means of an acoustical lens with ultrafast variable focus, located on the way of the reflected beam. One would observe then, at the level of a single detection plane, the sequence of the images of each of the planes illuminated by the acoustical exploring beams ; the rapid change of focus ($\Delta f/\Delta t \simeq 3$mm/$\mu$ s) permitting us at all times to focus on the image of the plane of which the echos are arriving on the detection plane. If it is true that such a decoding operation is perfectly utopian when classical acoustical lenses are used, one can imagine substituting the delay effect of such a lens by the action of a delay line network connected to a transducer array which will be used as the imput face of this lens. The ensemble of the delay electrical signal then will excite a second tranducer array which, acting as the output face of the lens, would deliver in a second propagation medium, an acoustical beam which,

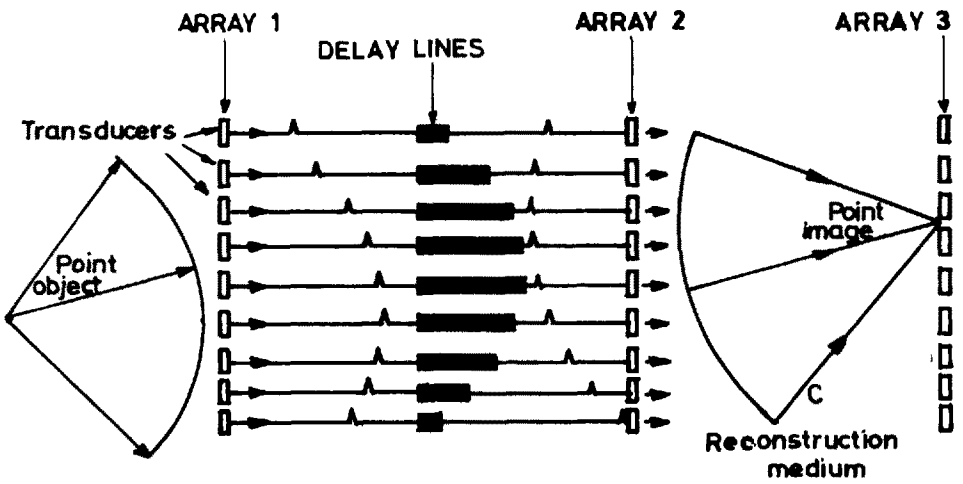

Fig. 3 - Electronic lens with delay lines and 2 transducers arrays, the third array is used to observe the acoustical image.

through the propagation effect, will create the acoustical image of the plane that is to be observed (Fig. 3). The instantaneous focus of such a system depends on the one hand on the distance z' at which one seeks to restitute the images of each plane of the object and, on the other hand on the depth $z = ct/2$ of the target observed at the time t

$$f = \frac{z\,z'}{z' - z} = \frac{ct\,z'}{2z' - ct} \qquad (15)$$

The quadratic correction of delay that must be applied by the electronic delay lines in order to simulate this focusing will be a function of the time according to an hyperbolical law

$$\Delta\tau\ (\vec{m},t) = -\frac{\vec{m}^2}{2cf} = -\frac{\vec{m}^2}{c^2}\left(\frac{1}{t}\right) + \frac{\vec{m}^2}{2cz'} \qquad (16)$$

This solution which would permit access to the tri-dimensional structure of an object during a single ultrasound "shot", no matter how attractive it looks, is, as of today, too complex technologically to be carried out. Whether it is the making of a very well damped two dimensional transducer array or the numerous use of analog controlled delay lines (for example, CCD type) while being extremely expensive, do not have a performance satisfying the requirements of an echography (high dynamic > 80 db), or also the display of tri-dimensional information, the technological problems that remain to be solved are enormous. In fact, until now, the echographic experiments have always been limited to a much more selective exploration of the objects, by illuminating either one plane or one line of the object during each ultrasonic "shot". The limitation to 2 dimensions of this type of parallel processing has been investigated by TORGUET and

BRUNEEL [10,11]. The principal interest of this parallel processes is the speed in forming an image ; each B mode echographic plane (the B plane is determined by the direction of the array and the direction of the ultrasound beam) is obtained in less than 300 µs. However, if this image rate seems interesting, one must think that in the case of a complex structure, the decoding of the coordinates of each target among the noise coming from other targets located at the same distance of the array, can deteriorate the resolution power in a significant manner.

B - Sequential echographic processing.

Until now, the sequential exploration processes of B mode echographic planes have given the results more exploitable by physicians. In these technics, one uses an even more selective illumination of the object. An ultrasound "shot" allows the formation of the image of only one line of the object and not of all the plane. To form the image of a line, after having selectively illuminated a section of the object with a relatively narrow ultrasound beam, one applies a focusing process which during the reception phase is going to select among the echos, the ones that comes from the line that is to be displayed. The image of a slice in B mode echography is then obtained sequentially, line by line, during a period that can last, in the case of the manual scanning of a transducer, some 10 seconds (Fig. 4). In the case of the electronic scanning of a transducer's array, this period can be reduced to less than a 1/50 of a second.

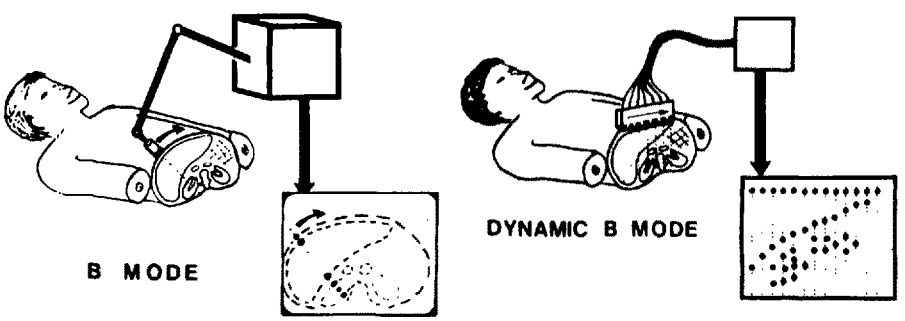

Fig. 4

In fact, practically all the echographic research has been carried out on the development of the sequential exploration systems. On the one hand, the decoding processes to set up are much simpler to make than in the parallel processing technics. On the other hand, the selective illumination by a relatively narrow beam largely reduces the spatial interference noise encountered with the other technics, in the case of the visualization of objects having a complex structure. In other words, by accepting N shots in order to form the image of a plane, one substitutes the deconvolution operation that was to be performed during a single ultrasound "shot" in the "parallel" technics,

with a "deconvolution" operated in a sequential manner which reduces, in a substantial way, the problems of interference noise.

Besides the technics using a single aperture transducer, one tries to have it focus along a caustic axis long enough [12,13], but which must be moved manually or mechanically in order to get the image of the B mode echographic plane, one has tried, obviously to use transducers' arrays, associating with the electronic scanning, the synthesis of a focusing aperture. The problem to solve then is the making, behind the transducer array, of the electronic filters that are adapted to the curvature of the diverging waves coming from the target under illumination. Such a filter which simulates the action of a cylindrical transducer with a varying focus, must bring a cylindrical correction of time delay to the signals delivered by the aperture transducers, then through the summation of these delay signals, must only produce a strong signal for each of the targets being studied (Fig. 5).

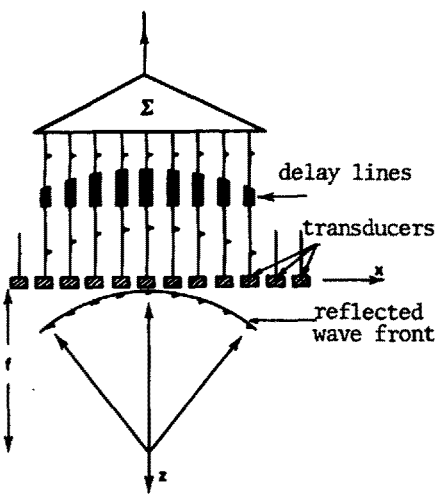

Fig. 5 - Electronic focusing during reception with delay lines.

At this time, many dynamic focusing processes have been developped that are using rather than time delay lines made of analogical shift registers (which as we know are not satisfactory enough to be used in echography), some digital time delay lines which permit one to obtain only a limited number of focal distances during an ultrasound "shot", but which, because of the relatively small aperture retained, will however give a good covering of the various focal zones. Moreover, in these systems, the use of time delay lines during the transmission can permit, when only a limited number of transducers is used, a scanning of the plane by sectors rather than linearly(Fig.6). By applying a transmit impulse slightly shifted in the time from one transducer to the other, one orientates the angle θ of the transmit beam according to a law :

$$\Delta\tau_T(x) = (\text{tg } \theta)\ x/c \qquad (17)$$

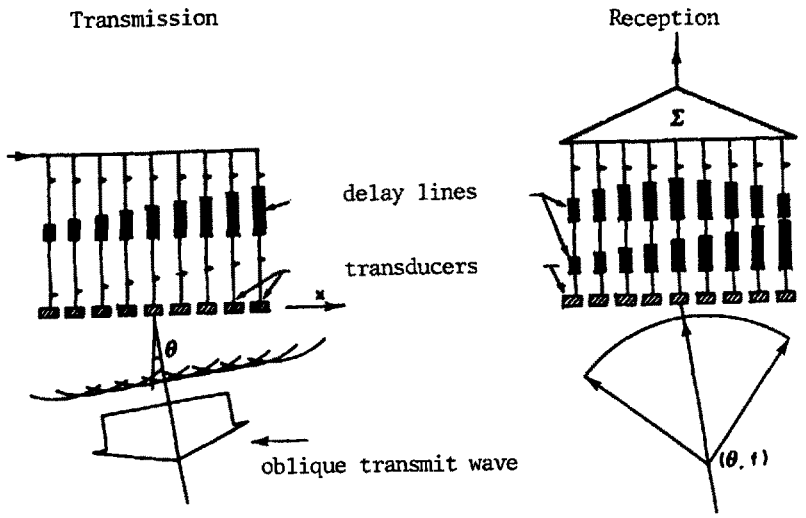

Fig. 6 - Electronic beam steering at transmission and oblique focusing during reception.

that can be changed from one "shot" to the other. One uses then during the receiving, an oblique focusing process which combines simultaneously the linear delay corrections and the quadratic delay corrections needed to recognize the target oriented at the angle θ (Fig. 6).

$$\Delta\tau_R(x) = \frac{tg\theta}{c}\, x - \frac{1}{2\,cf}\, x^2 \qquad (18)$$

Such a device has been developped for the first time by THURSTONE [14] that uses an array of 32 transducers, which allows the focusing on 5 different areas, but requires having at its disposal a large choice of time delays. It is a device still very complex and expensive.

More recently, in order to limit the number of time delay lines used, one has become interested in the linear scanning of the arrays containing a larger number of transducers (approx. 100). By limiting the number of transducers located in the focusing aperture at less than 20, an analogic multiplexing then allows the restriction of the number of time delay lines to a reasonable value [15]. However, in this last realizations, the focusing aperture remains relatively small. In the technic quoted in reference, the use of 16 transducers at 3 MHz with a pitch of 1 mm, limits the angular aperture of reception at less than 10 degrees for targets located at 10 cm, which means that a lateral resolution of less than 3 mm at 6 db cannot be obtained.

In absence of a simple and economic way for synthetizing large focusing apertures with

time delay lines, it has been thought that the focusing possibilities which are specific to the monochromatic waves and which since FRESNEL have been developped in optics, are in fact especially interesting in ultrasound echography. One knows that when considers a monochromatic wave of pulsation ω , the time delay concept can be systematically replaced with the one of phase shift. One can then, as in holography, choose to recognize the target under study, solely by means of the signature in phase of the waves that it generated. The most total electronic decoding process will consist then in substituting the quadratic correction of delay apllied behind the transducers, with a quadratic correction of phase under the form of :

$$\Delta \psi (x) = \omega \Delta \tau (x) = - \alpha x^2 \qquad (19)$$

The ability to define this law of phase modulo (2 Π) permits, in addition, the restriction of the number of dephasing networks required for the focusing. This "Fresnel" focusing mode, similar to the one obtained in optics with the kinoform lenses, appears to be extremely interesting in B mode echography where the signals used are necessarily brief. In this case, if the quadratic correction of phase made electronically is independant from the pulsation ω, that is to say that the coefficient α is a constant, the echos of the target will be decoded by such a lens on an appreciable depth of focusing. The quadratic correction of phase is in fact equivalent to a delay correction being different for each of the frequency components of the reflected signal.

$$\Delta \tau (x) = \frac{\Delta \psi (x)}{\omega} = - \frac{\alpha}{\omega} x^2 \qquad (20)$$

Therefore, one creates, by means of a single array of dephasor networks, a curvature of the delay law and a focal length that will be different for each of the echographic frequencies.

$$f = \frac{1}{2 \alpha c} \omega \qquad \leftrightarrow \qquad \Delta f = \frac{1}{2 \alpha c} \Delta \omega \qquad (21)$$

One reaches then, a system able to focus on a large depth, without having the need to modify the reception network in proportion to the echo return provided that the echographic signal as a wide spectrum $\Delta \omega$. It is an undeniable advantage of this focusing mode, which also can be used for the transmission in order to make a very narrow ultrasound exploring beam on a considerable depth [16]. The double focusing, at the transmission as well as at the reception, improves greatly the lateral resolution of the system. Even if this technique looks attractive, the making of frequency non-dependant dephasors is not an easy process. In fact, such dephasors can be perceived as dispersive lines, with a linear frequency dependance of the phase velocity $v_\phi = \beta \omega$. It must be noted that a "chirp" convolver (as the one used in radar technics) is able to realize this operation. However, high dynamic (80 db) convolvers of this type are difficult to obtain. Therefore, a strong phase sampling of the Fresnel law (19) is necessary ; only the phaseshift of 0 or Π can be obtained easily on wide frequency

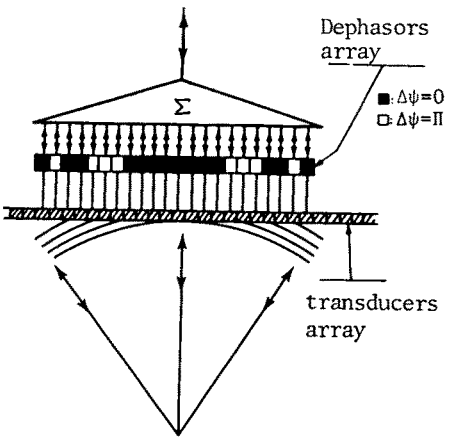

Fig. 7 - 2-state Fresnel focusing

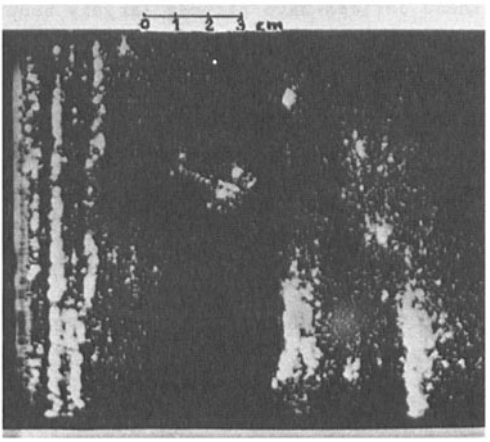

Fig. 8 - B echography of a 12 weeks fetus. Details of the vertebral column.

bandwidth. Each transducer is thus connected with a simple electronic inverter. It is a question of either keeping the signal as it has been received from the transducer, or processing it with an inverter, which will shift its phase of Π (Fig 7). The simplicity resulting from the use of electronic inverters instead of time delay lines, allows one to obtain easily very large focusing aperture sampled by an important number of transducers. We have developped such a Fresnel focusing technique from a linear array of 160 transducers [16, 17]. The focusing aperture is obtained from a group of 64 transducers. Electronic scanning of this aperture along the array allows us to obtain 25 frames of 160 lines per second with a lateral resolution of the order of 2 mm.

The clinical examination technique is simple and consists of putting the transducer array in contact with the skin by means of a coupling jelly and by changing its orientation in order to systematically explore the area of interest. When the probe is in a fixed position, the image observed on the screen is a slice of the organs in motion : for example, the motion of the liver or the kidneys when breathing, the change of caliber of the vena cava, the beats of the abdominal aorta, as well as the motions or the heart beats of the fetus. The excellent lateral resolution obtained allows us to observe some details that in practice were not easily distinguishable. On very young fetuses (8 to 12 weeks), it is possible to watch cardiac motion and to see vertebrae that are separated by less than 2 mm (Fig. 8).

In order to conclude this study on acoustical imaging, we must notice that, compared to the deceiving results of acoustical holography, echographic devices have proved their greater versatility specially after the development of the dynamic B echography.

These devices have already largely benefited from Fresnel focusing and holographic concepts to improve the quality of imaging and their lateral resolution particulary.

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