SYNCHRONOUS DYNAMIC FOCUSSING : A NOVEL ULTRASOUND IMAGING TECHNIQUE

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ABSTRACT

In real-time ultrasound linear scanners, focussing of the acoustic beam is generally achieved by selection of a single focus, improving the image resolution over a restricted zone along the penetration depth. A novel dynamic focussing technique, capable of continuously controlling the focus position synchronously with the echo return, is presented. The new approach is based on merely phase equalization of the signal delay along the array channels, associated with a dynamic variation of the array aperture. Negligible degradation of the directivity pattern due to the uncompensated group delay is ensured by a proper system design. The required phase distribution is instantaneously created at the taps of a delay line, individually connected with the array elements, fed by a properly controlled variable local oscillator. High quality images obtained by a prototype linear scanner are presented.

INTRODUCTION

In ultrasound echographic systems for medical diagnosis, the image quality largely relies on the capability of focussing the acoustic beam over the depth of exploration. Electronic focussing of array transducers is generally achieved by delay lines controlling the delay of individual signals at the array elements, so as to equalize the differences of geometrical paths from the desired focal point to the array. However, owing to the limited focal depth, extending over a restricted portion of the beam, a single focus provides a distorted image outside the focal zone. An undistorted image needs 'dynamic' focussing, i.e. variation of focus in synchronism with the arrival time of echoes received from different depths. Dynamic control of the delay profile synthesizing the electronic lens requires, however, rapid switch of the delay lines, and can produce undesirable switching noise during the transients. The approach used in many commercially available equipments consists in providing a number of selectable single focuses, which can also be combined in a 'multizone' focussing mode. In this operation mode, the penetration depth is divided in a number of zones, each of which is imaged by a different transmission burst and using the most appropriate focus both in transmission and in reception. This technique, however, requiring M acquisition cycles for M zones, reduces the frame rate by a factor M, thus not allowing imaging of very rapidly moving structures, and giving rise to a troublesome image persistence effect also when scanning a quasi static structure.

In previous papers [1, 2] a new technique for ultrasound beam steering/focussing has been introduced by the authors. The effect of phase delay and of group delay of signals was separately investigated. It was evidenced that the directivity pattern is prevailingly affected by the accuracy of phase equalization of the signal carrier, while compensation for group delay can be provided by coarse delay increments. The use of phasing techniques other than delay lines can thus release from the need of controlling a great number of delays a fraction of a wavelength apart.

In this paper the above considerations have been applied to focussing of the acoustic beam in a linear scanning equipment. Since moderate values of delay are needed, focussing has been obtained by merely phase equalization, without the use of delay lines. A phasing technique has been developed that allows continuous variation of the focus in synchronism with the echo arrival time. This technique, briefly referred to as SDF (Synchronous Dynamic Focussing), is described, and experimental results in vitro and in vivo are reported.

BASIC OPERATION

The maximum delay, t_{max}, needed for focalization depends on the focal length, F, and on the array aperture, D = Nd, with d the interelement spacing

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and N the number of elements in the array. Using the paraxial approximation, t_{max} is given by

$$t_{max} = \left(\frac{D}{2}\right)^2 \frac{1}{2vF} \qquad (1)$$

where v is the propagation velocity in the medium. According to the above mentioned approach [1, 2], this delay must be compared with the width, T, of the echo pulses to be processed. If t_{max} is contained within a suitably low fraction of T, then phase equalization can replace delay equalization along the array with no significant degradation of the directivity pattern. The echo pulse width, T, is determined by the Q of the piezoelectric transducer as well as by the frequency dependent absorption of the intervening medium and by electronic preprocessing. Experimental evidence indicates that typically a minimum width of three or four cycles at the operation frequency, f_o, is to be expected. As a consequence, it was estimated that efficient focussing can be achieved by merely phase equalization provided that the delay t_{max} (1) does not exceed approximately one cycle. This condition, however, cannot be fulfilled for every value of F in the range of interest (typically up to 180 mm) if a constant aperture, D, ensuring focus to be positioned in the near field of the array, were used. The array aperture is therefore changed as F is varied, taking on values D_F such that

$$\left(\frac{D_F}{2}\right)^2 \frac{1}{2F} < \lambda_o \qquad (2)$$

with $\lambda_o = v/f_o$ the acoustic wavelength. This condition ensures that the focus F lies within the near field of the aperture D_F. In practice, D_F can be varied in a discrete manner in increments of twice the interelement spacing, d.

The dynamic variation of the array aperture is a technique independently used in echographic systems: by mantaining the ratio F/D_F approximately constant, a quasi uniform beam width and depth of field can be obtained. This second feature can be reasonably compromised with condition (2) in order to achieve a quasi uniform beam shape.

SDF DESIGN

In order to implement phasing of echoes along the array channels, an heterodyne technique has been used. A phase-controlled coherent reference oscillation is mixed with the echo signal for each array channel, and the low-pass or the high pass component is selected, depending on the system design. In such a manner, the phase impressed to the reference oscillator is communicated to the signal. Various technique can be employed to control the phase of the reference oscillator. A set of discrete phase values in the range 0^o - 360^o with the desired increment can be generated in a number of ways, and programmably selected to provide the phase shift closest to the desired value. Phasing profiles corresponding to a number of focuses can be synthesized by such discrete techniques.

A technique has been introduced which allows the phase variation to be controlled in a continuous manner, thus providing the capability of a continuous variation of the focus.

The delays to be equalized for a focal length F_1 are given by

$$n^2 \frac{d^2}{2vF_1} \qquad n = 0,1,2,.. \qquad (3)$$

where n=0 corresponds to the central element of the array (an odd number of elements is assumed). The corresponding elementary phase increment comes out to be

$$\Phi_1 = 2\pi f_o \frac{d^2}{2vF_1} \qquad (4)$$

with f_o the frequency of the echo signal.

Consider a tapped delay line with tap delays given by (3). A local oscillator at frequency f_o, fed serially at the delay line input, provides at the parallel output taps a set of oscillations shifted in phase by the desired amounts. By mixing the individual tap outputs with the corresponding signals from the array, phase equalization is performed, and summation of signals gives rise to constructive interference for focus F_1.

When the focal length is changed to a value F, the required phase increment becomes

$$\Phi = 2\pi f_o \frac{d^2}{2vF} \qquad (5)$$

Assume that the frequency of the oscillator feeding the above delay line is changed to a value f such that

$$f \frac{d^2}{2vF_1} = f_o \frac{d^2}{2vF} \qquad (6)$$

i.e.

$$f = f_o \frac{F_1}{F} \qquad (7)$$

This change of frequency produces at the delay line taps the phase distribution required by the new focal length. Hence, by controlling the frequency of the reference oscillator in accordance with (7) the phase profile can be changed according to the desired focal length.

Consider now that the frequency law (7) is synthesized by a VCO in synchronism with the arrival time of echoes from different depths. Continuous frequency variation causes slightly different frequencies to be present at the taps of the delay line; however, it can be demonstrated that the phase error due to the spread of frequency over the delay line can be contained within negligible values. Then, continuous control of VCO frequency according to the hyperbolic law (7) produces continuous synchronous focussing of the acoustic beam.

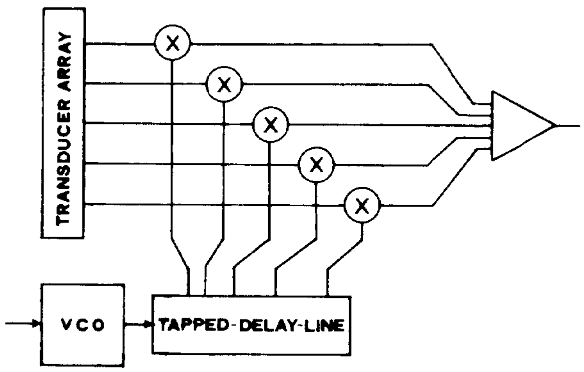

FIG. 1 Schematic diagram of SDF system.

The schematic diagram of the SDF system is shown in fig.1. A typical VCO control waveform, synthesizing a range of F from approximately 35 mmm to 150 mm, is shown in fig.2.

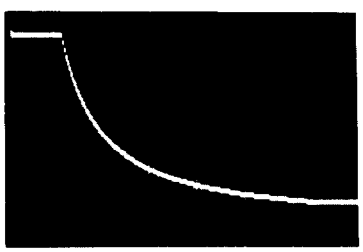

FIG. 2 VCO control waveform (40 μsec/div).

In the following experiments a modification of the above technique has been used, allowing VCO frequency higher than those given by (7). In this manner the low-pass components at the mixer outputs can be selected; the phase equalized signals after mixing are summed, and finally a second mixing with the VCO carries back the resulting signal to the original frequency for subsequent processing.

EXPERIMENTAL RESULTS

A commercially available linear scanner has been modified in order to experiment the new technique. The equipment possesses a selection of four focal distances, each focus being the same in transmission and in reception. The four focuses can be individually operated, allowing a frame rate of 24 images/sec, or combined in a multizone focussing mode, reducing the frame rate to 6 images/sec. The operation frequency was 3.5 MHz, and the maximum aperture of the array during reception was 24 mm, with 1.5 mm interelement spacing.

Signals from the array channels were taken out from the equipment, processed by the experimental SDF system, and fed back to the final processing chain of the equipment.

Two images of an AIUM tissue phantom, obtained by the equipment when one single of the four focuses is selected at a time, are shown in fig.3. The lack of resolution and the distortion outside the focal zone are apparent. These images must be compared with the image shown in fig.4, obtained with SDF in reception and a single focus in transmission at approximately 100 mm. The good resolution and beam uniformity along all the penetration depth can be remarked.

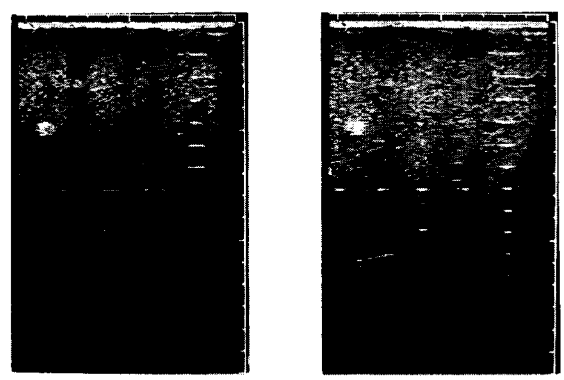

FIG. 3 Single focus images of a tissue phantom for two different focal distances (10 mm/div)

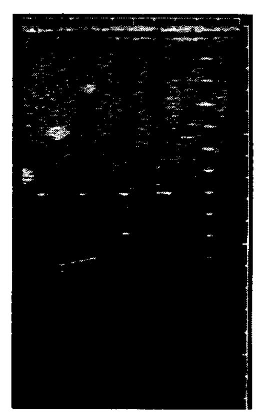

FIG. 4 Image with SDF in reception and 100 mm transmission focus (10 mm/div).

In spite of the fact that a single transmission focus is used, the image quality appear to be comparable with that obtained when operating the equipment in the *multizone* focussing mode (fig.5, left), where four focuses both in transmission and in reception are used at the expense of a low frame rate. For an homogeneous comparison, the image obtained using four focuses in transmission and SDF in reception is shown in fig.5 (right).

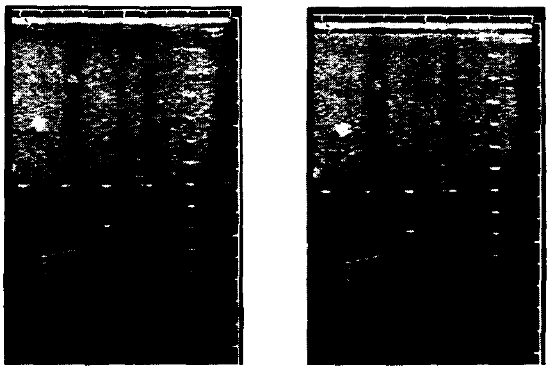

FIG. 5 Images obtained with four focuses in transmission.
Left: Four focuses in reception.
Right: SDF in reception.

Preliminary clinical evaluation has confirmed the superior quality of the images obtained with the SDF technique. As an example, images of a liver showing the hepatic vein are compared in fig. 6, obtained with SDF and a single transmission focus (on the left) and with the same single focus both in transmission and in reception (on the right).

FIG. 6 Images of a human liver obtained with SDF (left) and with a single 100 mm focus (right).

CONCLUSIONS

The new technique of synchronous dynamic focussing, based on continuous phase equalization of signals along the array channels, has demonstrated a high degree of effectiveness, offering superior performances with respect to techniques employing selection of a single focus or multizone focussing. SDF technique can thus advantageously replace in all operative conditions the conventional processing based on delay lines.

SDF system design depends on relationship (7), and hence on the center frequency of the return signal. Corrections due to variation of the center frequency, due to the frequency dependent absorption at different depths, can be easily taken into account in designing the VCO frequency control waveform.

Sharpness of focalization is strongly affected by the synchronism of VCO frequency with the corresponding depth of the reflecting target. Departure from the estimated propagation velocity can give rise to imperfect focalization. Change of velocity in different tissues can be taken into account by providing a choice of VCO frequency vs time behaviour, in such a manner that the focussing rate matching more closely the real propagation velocity in the scanned tissue can be selected by the operator.

Finally, individual control of phase at each array element can afford further improvement to the technique. In fact, dishomogeneity of tissues traversed by the sound path can cause removal from the phase distribution predicted for an homogeneous medium. Future developments of the technique can be enviseaged providing a sort of adaptive control of phasing at each array element on the basis of a comparative evaluation of signal phase with respect to a reference array element.

REFERENCES

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