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U1S S1031 S1032 S1945 S2139 S2190

(56) Documents Cited
GB 1583057 A US 4512197 A US 4469977 A
US 4269067 A
IBM Technical Disclosure Bulletin, Vol 21,No 8,Jan 1979, R J Von Gutfeld and SS Wang, p 3441-3442

(58) Field of Search
UK CL (Edition L) **G1G GED GEEA GES GET , H4J JCX**
INT CL^5 **G01N 29/24 , G10K 15/04 , H04R 17/00 23/00**
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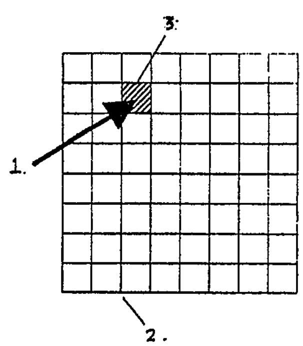
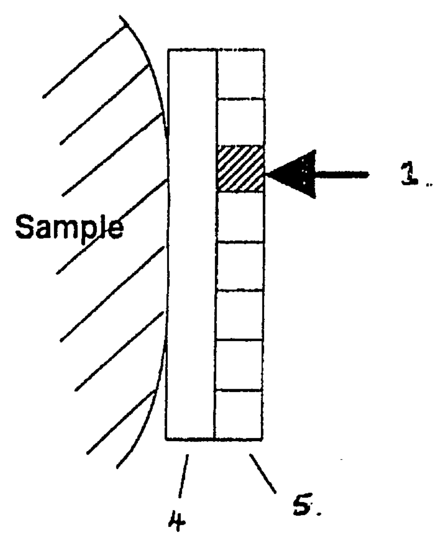
(54) Optically controlled ultrasound array

(57) An optically controlled ultrasonic array comprises method and means of optically addressing and activating a slab of a suitable material which can be pixelated or continuous and which acts as a remotely controlled ultrasonic array. The slab is addressed and activated by a scanning optical beam or by an optical array, and replaces hard wired ultrasonic arrays reducing or eliminating transducer wiring and solid state switching. The scanned addressing and activation into acoustic oscillation of areas on the slab, effective pixel elements, is remote and faster than can be achieved by electronic switching of individual transducer elements on a conventional piezoelectric transducer array, eg. by optically switching electrical parts of the elements or by optically activating photo-sensitive cells which will generate the electrical signals necessary to energise the elements. The optical control can make use of a photo-thermal or photo-acoustic transduction process to generate acoustic waves. The system can be used for medical diagnosis or treatment as well as non-destructive testing or ultrasonic treatment, including plastic welding.

Diagram 4. Optically Controlled Ultrasound Array

Suggested embodiment:





At least one drawing originally filed was informal and the print reproduced here is taken from a later filed formal copy.

The claims were filed later than the filing date within the period prescribed by Rule 25(1) of the Patents Rules 1990.

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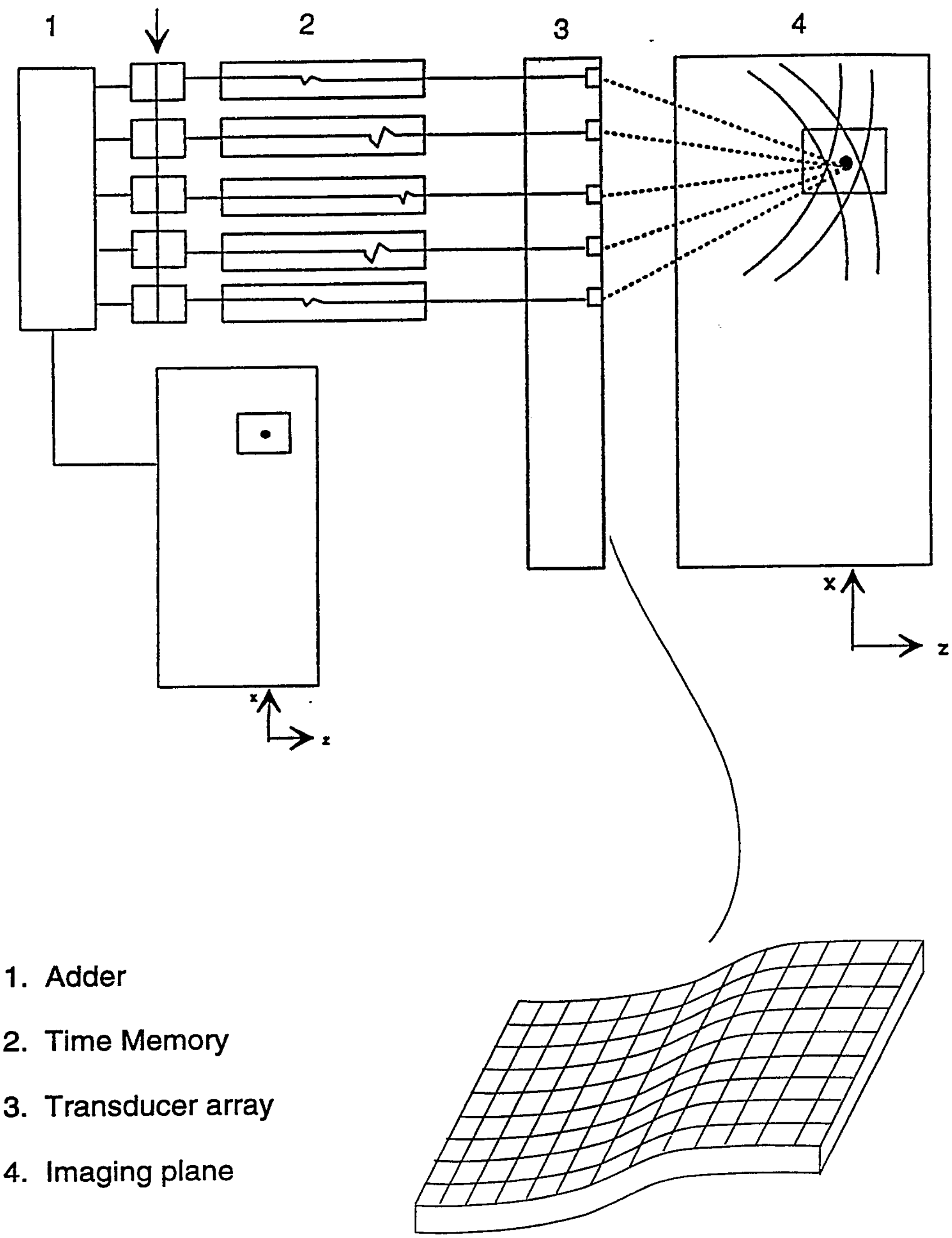


1. Adder
2. Time Memory
3. Transducer array
4. Imaging plane

Diagram 1: Quasi-real-time synthetic aperture imaging (cf ref (4))

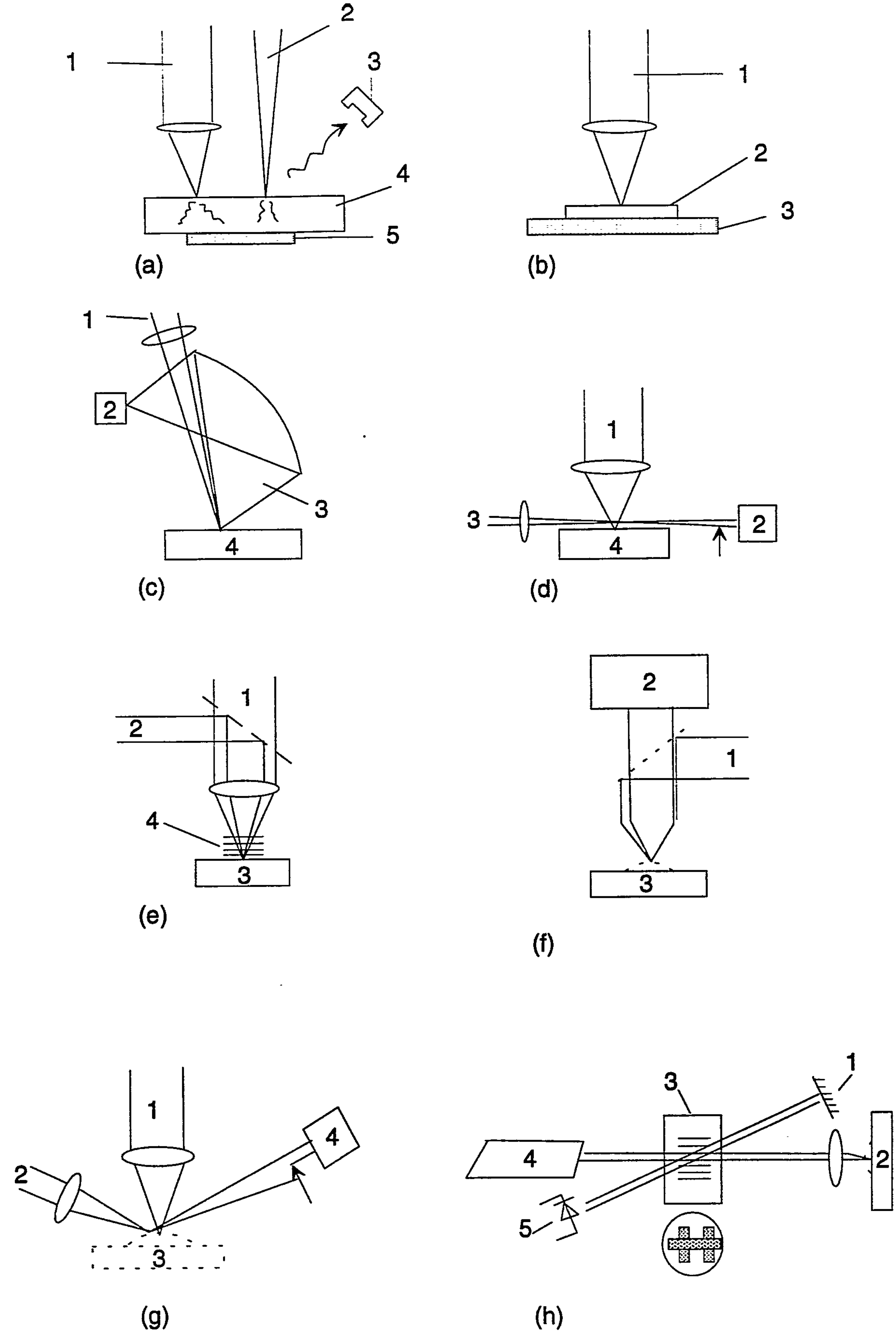


Diagram 2: Photo-thermal and photo-acoustic detection techniques

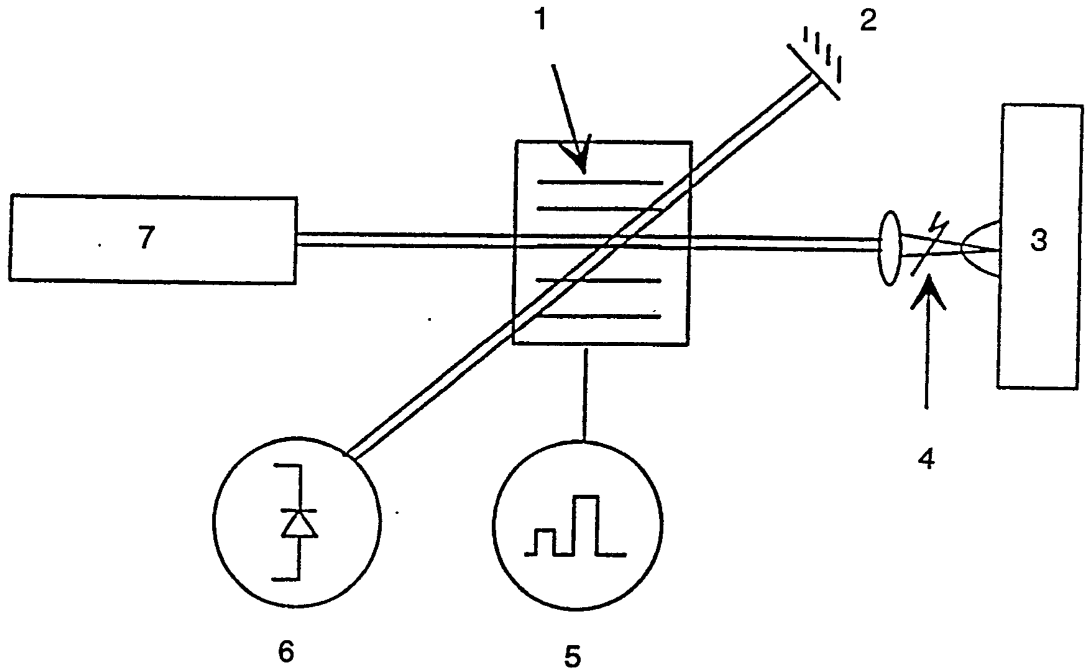


Diagram 3: Photo-displacement sensing using laser probe (modified Michelson interfermometer)

1. Adder
2. Time Memory
3. Transducer array
4. Imaging plane

Diagram 1: Quasi-real-time synthetic aperture imaging (cf ref (4))

(a) Photoacoustic

1. pump laser
2. electron beam
3, microphone
4. sample
5. PZT

(b) Pyroelectric Detection

1. pump laser
2. sample
3. pyroelectric detector

(c) Photothermal

1. pump
2. detector
3. thermal I.R.

(d) Mirage Effect

1. pump
2. detector
3. probe beam

(e) Bragg Scattering

1. pump
2. probe beam
3. sample
4. acoustic wave in the gas

(f) Photo-displacement Interferoxetrical detection

1. pump
2. interferometer
3. sample

(g) Photo-displacement detection by Beam Deflection

1. pump
2. probe beam
3. sample
4. detector

(h) Photo-Displacement with a single laser

1. mirror
2. sample
3. Bragg-cell
4. HeNe laser
5. photodetector

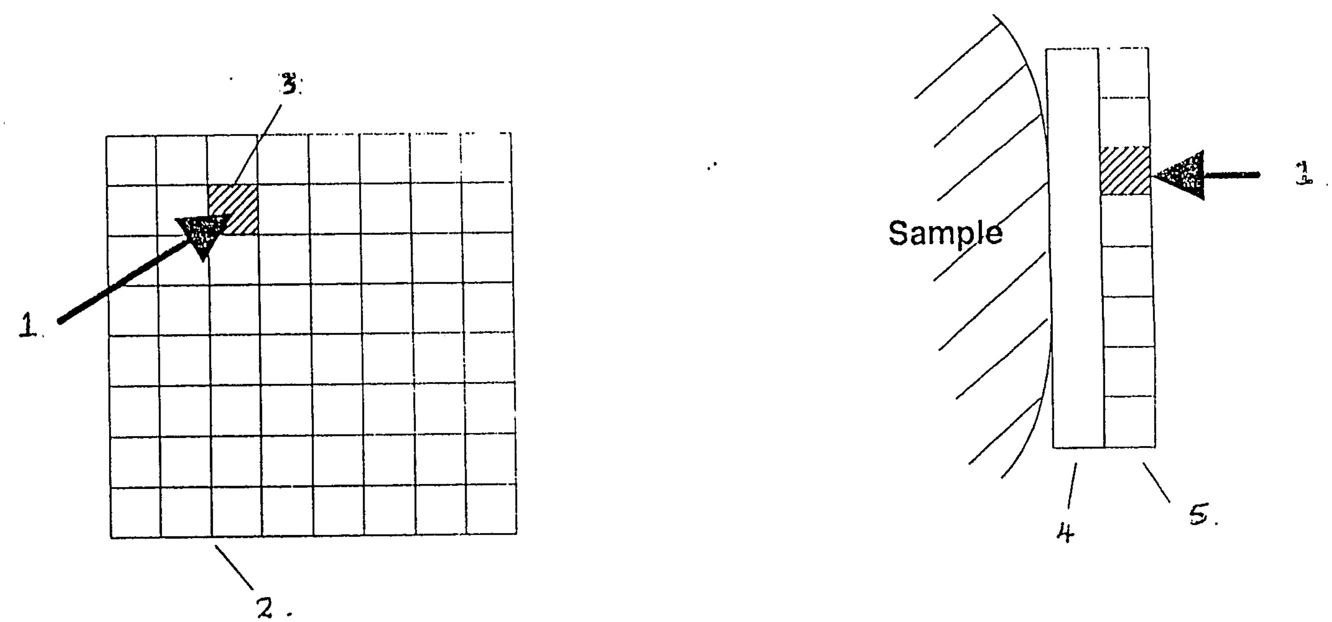
Diagram 2: Photo-thermal and photo-acoustic detection techniques

1. Bragg Cell
2. Mirror
3. Photoacoustic Transducer
4. Scanning Optics
5. Modulator
6. Detector
7. Laser

Diagram 3: Photo-displacement sensing using laser probe (modified Michelson interfermometer)

Diagram 4. Optically Controlled Ultrasound Array

Suggested embodiment:



OPTICALLY CONTROLLED ULTRASOUND ARRAY

Background to the invention

This invention relates to ultrasonic imaging or detection systems used in a number of applications including medical diagnostic imaging and NDT. These systems normally employ a transducer to change electrical energy into acoustic vibrations. These transducers are normally made of a piezoelectric material such as PZT, composite materials containing PZT or similar transduction material, PVDF, crystal quartz, LiNbO3, ZnO each used depending on the frequency characteristics required. Transducers can also be made of magnetostrictive material.

The technology underpinning acoustic transducers is non-trivial and demanding applications in NDT and medical imaging constantly drive transducer technology improvements. Many of the imaging transducers are of a single element type which is then mechanically scanned. Applications for acoustic imaging range from a few KHz (sub-sea use) to several GHz (microscopy). One of the leading applications remains in medical imaging where frequencies in the few MHz range are used. Here, solid-state arrays are replacing mechanical scanners; single element transducers and annular arrays are being replaced by 1.5 and 2D arrays with dynamic focusing. Although, linear arrays with fine transducer pitch are in common use, the logistics of transducer manufacture and the considerable switching and processing has made a large, fine pitch 2D array as yet un-realistic.

The generation of acoustic waves through the interaction of light on a suitable material has many applications in imaging and defect detection, particularly in NDT. The phenomenon of Photo-acoustics was probably first used in practice by Alexander Graham Bell in 1880 (" On the production and reproduction of sound by light", Am. J. Sci., 20, pp 305-324). In 1881, Both Lord J.W. Strutt, 'Rayleigh' ("The photophone", Nature 23, p 274) and W.H. Preece ("On the conversion of radiant energy to sonorous vibrations", proc. Roy. Soc. 31, pp 506-520) provided some explanation of the phenomenon.

The generation of ultrasound by laser pulses was first suggested by 'Dick' R.M. White ("J. Appl. Phys. 34, p.3559) in 1963. A useful review appears in (D.A. Hutchins, Phys. Ac., Vol. XVIII, W.P. Mason & R.N. Thurston, eds., Academic Press, San Diego, 1988, pp.21-123).

In general, the generation of elastic waves using a laser, is accompanied by:

- plasma formation (metals)
- dielectric breakdown
- electrostriction and local piezoelectricity
- vaporisation, ablation or thermoplastic behaviour
- thermoelastic effects
    - photothermal
    - photoacoustic

The first four effects are related to high intensities. With lower intensities, one is in the thermoelastic effect, which is suitable for NDT and imaging. The interaction of light with matter is often accompanied by the coupling of energy into an acoustic mode, in turn generating heat due to the thermal conductivity, viscosity and relaxation mechanisms in the material.

The periodic heating of the surface by a modulated light source gives rise to a temperature cycling. This can be detected thermally (IR radiometry, pyroelectric detection), optically (diffraction effects in the gaseous coupling medium in the vicinity of the heated spot) or acoustically (microphone on solid, gaseous photo-acoustic cell). In general, the heating is accompanied by a thermo-elastic effect and a thermo-acoustic effect (heated gas). Photothermal and photoacoustic effects are thus both there and it is up to the detection scheme to use one or the other as an imaging mechanism.

The thermal diffusion length, the depth of effective penetration and interaction, is proportional to the root of the thermal conductivity and inversely proportional to the modulation frequency and the sound velocity. The thermal effect is governed by a diffusion process and it can be detected optically in the close vicinity (within one diffusion length) of the source, both in the imaging medium (skin) and in the coupling medium (air). Further afield, the acoustic effect can be detected using a suitable transducer (e.g. a microphone); in the air, the effect of the acoustic field is attenuated and also suffers a large impedance mismatch between the solid and the gas.

PRIOR ART

Ultrasonic array transducers are in wide use. At frequencies applicable to medical imaging, these are normally made of PZT, PZT/epoxy matrices or PVDF. In general, PZT transducers with only a few dB insertion loss (20log of the ratio between power in and power out) are possible, but bandwiths obtained are limited (5 to 20% typically). Using composites or PVDF, it is possible to obtain much wider bandwidths, even those approaching 100%, but inevitably, at the loss of efficiency (insertion losses of several tens of dB. PZT is a fired, compressed ceramic powder material, which is sturdy, but needs protection from moisture and extreme heat or large fields. Its acoustic impedance is c. 30MRayls, nearly two orders of magnitude higher than that of water. In order to improve the transmission of sound from the PZT element into water (or tissue, which has similar acoustic impedance) matching layers (1/4 wavelength thick) of soft materials such as silicon rubbers are used.

The technology of the transducer manufacture is thus non-trivial and demanding. For large arrays, the logistics of switching adds yet another level of extreme difficulty. Each transducer must be switched according to the phase or timing requirement of the array. The responses must be comparable from transducer to transducer. Each element must be able to cope with both large transmission signals and small return signals, at times 100 dB smaller than the transmitted signal. This requirement for large dynamic range imposes demanding specifications on the switching and amplification electronics; added to this are the speed requirements. First, there must be sufficiently fast recovery after the large transmission signals. Then, there must be sufficient on/off (extinction) ratios in the switches. Finally, there must be sufficient isolation from element to element.

The ultrasonic array transducer and switching assembly are then connected to the transmitter/receiver electronics through a cable, often with as many wires as there

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are elements, sometimes with some of these having been coded and compressed so that the cable is not impossibly fat.

SUMMARY OF THE INVENTION

This invention provides an optical means of activating and addressing the ultrasonic transducer. In general, there are three ways in which one can achieve optical addressing and activation.

There are in essence three major aspects to this invention:

- Optically addressing and activating an ultrasonic transducer element by optically switching the electrical port of a conventional acoustic transducer element
- Optically activating a photo-sensitive cell which will generate the electrical signal necessary to energise a piezo-electric transducer
- Optically generating an acoustic signal by means of a photo-thermal/photo-acoustic transduction process through the absorption of the light energy. In this aspect of the invention, the acoustic wave can be generated either entirely by the light input, or by means of preconditioning the medium to need a limited light input to change its properties (eg matter phase change).

In all aspects of this invention, the optical addressing and activation will afford reduction or elimination of transducer wiring and solid-state multiplexing (switching); remote addressing; and fast scanning.

According to one aspect of the invention, acoustic waves are generated on the surface of a material to be tested (skin, metals, ceramics etc.) by an optical means of addressing pixel elements, which can be physically isolated or defined by the area of illumination. The pixel elements are designed to be light absorbing and generating an acoustic wave either directly or indirectly. In the direct method, the light absorbing pixel medium is designed for high photo-acoustic transduction efficiencies. In the indirect method, the light absorbing medium is designed for high efficiency photo-electric interaction.

DETAILED DESCRIPTION OF THE INVENTION

According to one aspect of this invention, the pixelated transducer is made of a series of adjacent cells filled with a light absorbing dye. In another aspect of the invention, the pixelated transducer is made of a simple single layer or a complex multi-layer sandwich of soft and solid material designed for optimum transduction efficiency. In yet another aspect of the invention, the slab is not physically pixelated, but pixel elements are in effect defined by the size of the laser spot scanning over the surface. The scanning laser light spot may be a single or a series of simultaneous spots allowing serial or partially parallel phasing

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of the elemental sources.

The scanning of harmonically or amplitude modulated light over the pixelated slab causes a logarithmic increase and an exponential decrease in surface temperature, in harmony to the modulation of the exciting laser light. Depending on the thermal properties of the scanned surface, there may or may not be a nett increase in surface temperature, a dc effect superimposed on the ac signal.

One aspect of this invention relies on the photo-acoustic generation of thermo-elastic waves as the basis for a new type of ultrasonic transducer array with optical phasing. According to our invention, the ultrasonic array transducer consists of a slab of pixelated photoacoustic elements. An amplitude or harmonically modulated light source scans the surface of the pixelated slab, causing the generation of an ultrasonic wave within the light absorbing medium contained within the slab. This medium may be a light absorbing dye or a metal with good photo-acoustic properties.

The process of photo-acoustic transduction generally has very small quantum efficiency. It is nevertheless possible to generate sufficient acoustic signals as the generation and dissipation of sufficiently high optical powers as excitation is not a major difficulty.

The proposed system consists of:

- A high power semiconductor laser source (Generics is already capable of producing 25 W into an optical fibre);
- An intensity modulation system (mechanical or Bragg-cell based);
- A novel multi-element photo-acoustic transducer array as proposed by Generics;
- A detection mechanism (photo-displacement detection directly on transducer elements or separate single-element piezoelectric detector).
- A digital imaging algorithm and the associated hardware (e.g. the Stanford DAISY system)

The proposed transducer is intended to provide acoustic energy at ultrasound frequencies and is thus a possible alternative to medical ultrasound arrays in some applications. The principle of operation of the transducer is photo-acoustic because the ultrasonic energy is thermally generated.

Photo-acoustic imaging is an established technique in NDT and much work has been carried out describing resolution and contrast mechanisms in microscopic imaging applications [6-10]. The art will be familiar to experts.

Thermal waves can be generated by any time-varying heat source. Light absorbed locally and characteristically by the illuminated portion of a sample is converted partly into thermal energy due to absorption. In harmony with the incident light intensity modulation frequency, a periodic heat flow results into the surrounding region. This periodic heat flow (diffusion) is non-radiative within the sample because it is so highly damped that it reduces to 1/e of its peak within one wavelength (the thermal diffusion length). The periodic heat flow also results in an acoustic wave which is not critically damped and thus propagates like a normal wave, interacting with and attenuated by sample properties over a much larger distance. In general, three processes contribute to the contrast:

- Local variation in the light energy absorbtion characteristics of the surface (reflection, absorption, de-excitation)

- Scattering of photo-acoustically generated thermal waves

- Scattering of photo-acoustically generated ultrasonic waves

High resolution photo-acoustic microscopy has depended mainly on the first of these effects, thus providing surface contrast. Interior imaging microscopy (ie photo-thermal microscopy) has depended largely on the second. The proposed medical ultrasound array will be based mainly on the third.

Typically, modulation frequencies in the 100 kHz to 10 MHz or more result in thermal wavelengths in metals of 10 - 1 μm (some five times less in thermal insulators). However, the corresponding acoustic wavelengths are in the order of 15 mm - 150 μm. In photo-acoustic and photo-thermal imaging using low chopping frequencies, the acoustic wavelength is too coarse to be of interest and as a result, the acoustic wave is used as a carrier for finer thermal information. The Generics concept however makes use of the propagating acoustic wave in its interaction with the sample. In the case of medical imaging, it is intended typically to work at 2 to 5 MHz for useful resolutions.

The Generics concept thus boils down to using the photo-acoustic effect as a source of scannable acoustic energy in a new type of medical ultrasonic array.

FOCUSING AND SCANNING

Although it is appreciated that an ideal solution to fast imaging applications such as blood flow and certain specific cardiac cases is very important, we consider this to be a specialised area of flow analysis as opposed to imaging. It is, in our opinion, premature to address this particularly demanding specific case. During the discussion, we postulated 3D imaging and thus mainly spoke of a 2D array. It is indeed the expectation that the Generics concept will address 3D imaging applications; the approach will be to design first a linear array, followed by a phased array [1,2,11,12,13]. We thus describe the operation principle of a linear array as a basis of operation for a 2D array (3D imaging).

In a linear array, a single or a group of elements are fired to illuminate the sample in a direction normal to the face of the array. The scanning is performed by translating across the face of the array the firing sequence. In a phased array, a larger group of more closely packed transducers are fired simultaneously with appropriate phase (or timing) in order to produce a steerable focus.

There are a number of possible modes in which the proposed transducer may be scanned and a focus produced. We have postulated a system, which, on the

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limit, relies on sequential triggering of a single transducer at a time. This will be the worse (slowest) case. We thus choose this case as an example to illustrate how the focusing and scanning may be achieved. Furthermore, the description is centred on a linear transducer capable of producing a B-scan image. Other more complex modalities are possible, but their working principle will be essentially based on the simpler steps to be described here.

In an ideal transducer both efficiency and bandwidth are high. The photo-acoustic transduction process is a lossy process and it is expected that efficiencies will be low and bandwidths fairly high. Efficiency will allow detectable return signals with sufficient dynamic range. Bandwidth will allow short pule operation and thus both high inherent depth resolutions and faster possible scanning/processing rates.

We propose, as a possible limiting case, a system using synthetic focusing of acoustic images. Such a system has been proposed and successfully demonstrated and implemented elsewhere, on a switched, hardwired, conventional acoustic transducer array [3,4]. The adaptation of the system to the proposed photo-acoustic system is best explained on diagram 1.

In transmit mode, the Bragg cell modulates the intensity of the laser beam at the required frequency, say 2 MHz. The modulated beam is then made to scan over the surface of the array (initially a linear array) so as to excite, with a pulse, a single element at a time (it may be possible to illuminate several elements simultaneously for phased array mode). Assuming a 30% bandwidth, the shortest possible pulse will be 1500 ns. Assuming an attenuation of 2 dB/cm in tissue, an imaging depth in B-scan mode of 5 cm will result in two-way attenuation of 20 dB at normal incidence. The wavelength in the tissue will be in the order of 750 μm and in the transducer possibly in the order of 1.5 mm. A transducer pixel of this proportion will have a Fresnel distance of approximately 200 μm at which little self-focusing occurs. The 3 cycle pulse emanating from the transducer will then travel into the depth, with any echo from 5 cm depth returning between 66 microseconds (at normal incidence) and c. 100 microseconds (for a 32 element array of say 8 cm length). If an image area of say 256x256 is required, the return signal can be quantised in .4 microsecond intervals, corresponding to a fraction of the 3 cycle pulse, well over the Nyquist limit. Allowing up to 100 microseconds delay per element, sequential triggering will translate to a frame rate of c. 3 ms or 300 Hz. With a 20 MHz clock rate, 10 samples per wavelength would be possible and up to 300 frames can be produced every second.

In receive mode, for every point of interest in the xz B-scan plane, a look-up delay table will give the appropriate pre-calculated delays per element channel. For each channel, a feature in time could have emanated from any point on a circle centred at the element and within the imaging area. By adding the time corrected signals from all the elements, as in a backprojection, the intersection between these circles, corresponding to the position of the feature is emphasised by the sum of the intensities. The received signal is upshifted by twice the modulation frequency as it passes through the Bragg cell for a second time before detection.

In a 2D array of say 32x32, one simple way of producing 3D images would be

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to repeat the above operation for all elements, resulting in complete coverage of a volume approximately 8x8x5 cm in approximately 100 ms.

It is thus conceivable that real-time imaging will be possible even in 3D mode. At these rates, a jitter of 10 m/s from a feature in the tissue will cause an error of one two hundredth of a wavelength in the depth direction and around one wavelength from adjacent elements in the lateral direction.

We conclude that a quasi-real-time imaging system based on sequential aperture synthesis is likely to be possible. There is thus no need for the array to be either simultaneously illuminated nor for it to be illuminated necessarily within the phase delay between the closest and the fartest elements to points of interest in the imaging area.

DETECTION- PRINCIPLE

Whilst a number of detection schemes may be proposed (single piece piezoelectric detector, charge coupled delay devices etc.), it is likely that the signal levels will be too low for most of these. Signal levels are briefly discussed in the next section.

We consider first, detection based on a single piezoelectric element of wavelength proportions placed in the centre of the photo-acoustic array. It is shown in the next section that acoustic power levels c. 1mW can be expected for laser power input of 25 W, after a number of simplifying assumptions. In such a case the signal will be easily detectable and the detection bandwidth will not be a limiting factor. The implications of this on the scanning logistics must be considered. In transmit mode, the scanning is as described previously. In receive, however, the x,z position of the raster scan will correspond to a single distance, from the fired transducer to the x,z point and back to the receive transducer. This distance is unique for every fired transducer. Looking at the reflections due back from each fired transducer, the stored value for each delay could map to a number of possible positions. The aliasing thus produced is gradually corrected as the response due from other transducers are also added, emphasising true reflection features in common amongst the various transducers. This problem will be reduced if a phased array is employed, as the point of focus will be predefined during transmission.

The implications of the above during transmission scanning is that each line should be scanned fast enough for the longest delayed and shortest delayed returns to interact. For a 38 degree aperture half-angle (F 1.6) array to focus at one corner 5 cm deep, the shortest time is 66 microsecond and the longest 126 microsecond allowing 60 microsecond for scanning the elements sequentially, 17 kHz line rate (1.87 microsecond per element on a 32 element array). To focus at 1 cm depth along the axis of the transducer, this time difference is reduced to 14 microseconds. Thus the scanning speed and direction will define the focus uniquely in transmit. The digitised returns can be limited to a specific time window, expected to be the delay between the defined focus and the receive transducer. Thus during reconstruction, each x,z point in the raster will be associated to a specific memory location where the digitised return signal has been stored.

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Naturally, it would be attractive to uphold the wireless nature of the transducer in receive mode also and to this end, we consider an alternative detection scheme using optical detection.

A number of photo-thermal and photo-acoustic schemes have been proposed in the literature and shown schematically in diagram 2, amongst which we propose a very highly sensitive surface displacement detection scheme known as photo-displacement microscopy [5], shown separately in diagram 3. This scheme is fairly well established and is otherwise known as the laser probe technique and has been demonstrated to produce sensitivities to 10^{-4} angstrom displacements per Hz bandwidth.

To this end, a modified Michelson interferometer is used which incorporates an acoustically controlled Bragg cell. The Bragg cell will act as an active grating/splitter performing a number of tasks, amongst which are:

- Beam scanning;
- Intensity modulation of the incident beam;
- Isolation of incident and received beams in the frequency domain;
- Heterodyning of the signals for detection;
- The use of a single laser for surface heating and for displacement detection.

The photo-displacement signal is shifted to the second harmonic of the Bragg modulation frequency and easily detected in a bandwidth appropriate to the signal to noise levels achieved. It is this final detection bandwidth which is believed will ultimately define the frame rate.

Two possible modes of detection were described. It is concluded that although the detection scheme will be non-trivial, it will nevertheless be possible either to detect the signal using a piezoelectric element in the centre of the photo-acoustic array or to employ a photo-displacement sensing technique based on a modified Michelson interferometer capable of 10^{-4} Angstrom/Hz depth resolution.

SIGNAL LEVELS

A full analysis of overall signal to noise performance of the system will be required. We present here some preliminary considerations.

We assume a 25 W laser power over an area of 400 dimensions (This is a power density presently achieved by Generics designed semiconductor laser cluster).

Using one-dimensional theory, the peak temperature rise can be shown to be proportional to the peak laser power density and inversely proportional to the chopping frequency, density, specific heat capacity and the thermal diffusion length. For the 25 W laser power into fibre dimensions, this translates to a temperature rise of c. 13 degrees in water (.2 in a metal) per cycle. This corresponds to a deflection in the order of 10^{-7} m on the surface of the heated spot. Based on further assumptions yet to be verified, we calculate strains of 10^{-4}, translating to an acoustic power output in the order of one Watt. Upon return from say 5 cm within tissue, this power will be reduced by some 30 dB, resulting in an acoustic power of 1 mW, c. 0 dBm, which is easily detectable acoustically from a single transducer.

These calculations and the assumptions taken are for illustrative purposes only and it is not suggested that a full 25 W laser will be used, nor that signal levels detected will be purely based on the above assumptions.

Based on the assumptions taken, it is concluded that adequate signal levels may be produced.

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Diagram 1. Quasi-real-time sequential synthetic aperture imaging (cf ref. [4])

Diagram 2. Photo-thermal and photo-acoustic detection techniques

Diagram 3. Photo-displacement sensing using laser probe (modified Michelson interferometer)

Diagram 4. Optically controlled ultrasound array

Optically Controlled Ultrasound Array

Claims

1 An ultrasonic or acoustic transducer element or array which is addressed and activated by optical means.

2 An ultrasonic or acoustic transducer element or array which is addressed and/or activated remotely.

3 A wireless ultrasonic or acoustic element or array.

4 A method of remotely generating or controlling acoustic or ultrasonic signals for the purpose of medical diagnosis or treatment or industrial non-destructive testing or acoustic treatment such as welding.

5 A means of generating acoustic or ultrasonic signals remotely by using a slab of pixelated or continuous material that is addressed by a scanning optical beam or an optical array.

6 The element or array of claim 1 where the electrical port of a conventional acoustic transducer element is optically switched.

7 The element or array of claim 1 where the optical addressing energises a piezo-electric transducer element or array through use of photo-sensitive cells.

8 The element or array of claim 1 where the acoustic signal is generated through a photo-thermal or photo-acoustic transduction process.

9 An acoustic or ultrasonic transducer element or array, where the scanning of the array is achieved through optically addressing the surface of the array directly, the surface being pixelated or continuous.

Patents Act 1977

Examiner's report to the Comptroller under Section 17 (The Search Report)

Application number

GB 9205577.1

Relevant Technical fields

(i) UK Cl (Edition L) G1G (GED, GEEA, GET, GES); H4J (JCX)

(ii) Int Cl (Edition 5) G01N (29/24); G10K (15/04); H04R (17/00, 23/00)

Search Examiner

D L SUMMERHAYES

Databases (see over)

(i) UK Patent Office

(ii) ONLINE DATABASE: WPI

Date of Search

8 JULY 1993

Documents considered relevant following a search in respect of claims 1-9

Category (see over)	Identity of document and relevant passages	Relevant to claim(s)
X	GB 1583057 (IBM)	1-4
X	US 4512197 (GUTFELD)	1-5, 9
X	US 4469977 (QUINN)	1-4
X	US 4269067 (TYNAN)	1-4
X	IBM Technical Disclosure Bulletin, Vol 21, Number 8, January 1979, R J von Gutfeld and S S Wang, "Laser Generated Acoustic Fan Beam Ultrasound Tomograph System" pages 3441-3442	1-4

SF2(p)

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Categories of documents

X: Document indicating lack of novelty or of inventive step.

Y: Document indicating lack of inventive step if combined with one or more other documents of the same category.

A: Document indicating technological background and/or state of the art.

P: Document published on or after the declared priority date but before the filing date of the present application.

E: Patent document published on or after, but with priority date earlier than, the filing date of the present application.

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