(12) INTERNATIONAL APPLICATION PUBLISHED UNDER THE PATENT COOPERATION TREATY (PCT)

(19) World Intellectual Property Organization
International Bureau

(43) International Publication Date
22 May 2014 (22.05.2014)

(10) International Publication Number
WO 2014/076674 A2

(51) International Patent Classification:
A61B 5/00 (2006.01)

(21) International Application Number:
PCT/IB2013/060206

(22) International Filing Date:
18 November 2013 (18.11.2013)

(25) Filing Language: English

(26) Publication Language: English

(30) Priority Data:
61/727,342 16 November 2012 (16.11.2012) US
61/862,175 5 August 2013 (05.08.2013) US

(71) Applicant: **KONINKLIJKE PHILIPS N.V.** [NL/NL]; High Tech Campus 5, NL-5656 AE Eindhoven (NL).

(72) Inventors: **ERPELDING, Todd Nicholas**; c/o High Tech Campus 5, NL-5656 AE Eindhoven (NL). **KE, Haixin**; c/o High Tech Campus 5, NL-5656 AE Eindhoven (NL). **XIE, Hua**; c/o High Tech Campus 5, NL-5656 AE Eindhoven (NL). **WANG, Lihong**; c/o High Tech Campus 5, NL-5656 AE Eindhoven (NL).

(74) Agents: **STEFFEN, Thomas** et al.; High Tech Campus Building 5, NL-5656 AE Eindhoven (NL).

(81) Designated States *(unless otherwise indicated, for every kind of national protection available)*: AE, AG, AL, AM, AO, AT, AU, AZ, BA, BB, BG, BH, BN, BR, BW, BY, BZ, CA, CH, CL, CN, CO, CR, CU, CZ, DE, DK, DM, DO, DZ, EC, EE, EG, ES, FI, GB, GD, GE, GH, GM, GT, HN, HR, HU, ID, IL, IN, IR, IS, JP, KE, KG, KN, KP, KR, KZ, LA, LC, LK, LR, LS, LT, LU, LY, MA, MD, ME, MG, MK, MN, MW, MX, MY, MZ, NA, NG, NI, NO, NZ, OM, PA, PE, PG, PH, PL, PT, QA, RO, RS, RU, RW, SA, SC, SD, SE, SG, SK, SL, SM, ST, SV, SY, TH, TJ, TM, TN, TR, TT, TZ, UA, UG, US, UZ, VC, VN, ZA, ZM, ZW.

(84) Designated States *(unless otherwise indicated, for every kind of regional protection available)*: ARIPO (BW, GH, GM, KE, LR, LS, MW, MZ, NA, RW, SD, SL, SZ, TZ, UG, ZM, ZW), Eurasian (AM, AZ, BY, KG, KZ, RU, TJ, TM), European (AL, AT, BE, BG, CH, CY, CZ, DE, DK, EE, ES, FI, FR, GB, GR, HR, HU, IE, IS, IT, LT, LU, LV, MC, MK, MT, NL, NO, PL, PT, RO, RS, SE, SI, SK, SM, TR), OAPI (BF, BJ, CF, CG, CI, CM, GA, GN, GQ, GW, KM, ML, MR, NE, SN, TD, TG).

Published:
— *without international search report and to be republished upon receipt of that report (Rule 48.2(g))*

(54) Title: IN-PLACE CLUTTER REDUCTION FOR PHOTOACOUSTIC IMAGING

(57) Abstract: A medical image acquiring device (100) repeatedly photoacoustically images an intracorporeal spatial location (132) that is common image to image, varying at least one of ultrasound transducer acquisition location, and light excitation, image to image so that averaging of the images serves to decorrelate background clutter that exists in the images, of the spatial location, individually. The varying of one may be in the absence of the other, and may entail movement. The movement may be of an optical fiber bundle, a light guide or free space optics, as by rotation (140), translation or vibration. It may be of an optical mask or engineered diffuser, or it may entail toggling between active states of two spaced apart, separate optical fiber branches near the probe surface. If the movement pertains instead to acquisition location, it may imply translation of the active receive aperture (127), or rotation of the transducer as in the imaging plane in which case co-registering of the images may use synchronous B-mode images to detect skin orientation and a landmark.

In-Place Clutter Reduction for Photoacoustic Imaging

FIELD OF THE INVENTION

The present invention relates to photoacoustic imaging and, more particularly, to averaging photoacoustic images.

BACKGROUND OF THE INVENTION

Photoacoustic (PA) imaging is a non-ionizing and noninvasive hybrid imaging technique that can measure strong optical absorption contrasts with high ultrasonic spatial resolution at depths beyond the optical diffusion limit. Both image resolution and depth are highly scalable with the ultrasonic frequency. Since ultrasonic scattering is 2-3 orders of magnitude less than optical scattering on the basis of per unit path length, photoacoustic imaging can break through the fundamental limitation of existing pure optical imaging. Photoacoustic imaging relies on the photoacoustic effect to generate pressure waves that can be detected by ultrasound array transducers. Typically, a short laser pulse illuminates tissue leading to optical absorption, rapid heating and thermoelastic expansion to produce pressure waves. Since photoacoustic images can be reconstructed following ultrasound detection with conventional array transducers, photoacoustic imaging is highly compatible with ultrasound imaging. The same detection mechanism greatly simplifies spatial registration between photoacoustic and ultrasound images. Photoacoustic imaging is an emerging modality that could expand the scope of diagnostic ultrasonography into new clinical applications, such as image-guided sentinel lymph node biopsy, breast cancer diagnosis, and therapy monitoring.

The preferred implementation for photoacoustic imaging is the so-called reflection mode, which is most similar to hand-held ultrasonography. In reflection mode photoacoustic imaging, both the light source and the ultrasound transducer are placed in close proximity on the skin surface. A typical photoacoustic transducer is designed with an optical fiber bundle bifurcated to flank both sides of the ultrasound array transducer. Photoacoustic waves are received by the ultrasound transducer in the same geometry as pulse-echo ultrasound.

Photoacoustic imaging extends the imaging depths beyond the optical diffusion limit while maintaining ultrasonic spatial resolution. It enables clinical applications at depths and/or spatial resolutions that would be impossible with pure optical imaging techniques.

However, deep reflection-mode photoacoustic imaging suffers from background clutter signals that degrade image contrast. One source of background clutter is from optical absorption near the skin surface (blood vessels, melanin), launching acoustic waves that are backscattered to the transducer from acoustic inhomogeneities in tissue.

Jaeger et al. describe a method for clutter reduction based on tissue shear deformation and motion-compensated signal averaging ("Improved Contrast Deep Optoacoustic Imaging Using Displacement-Compensated Averaging: Breast Tumor Phantom Studies", M Jaeger et al., Phys. Med. Biol. 56 (2011) 5889-5901), hereinafter "the Jaeger study."

SUMMARY OF THE INVENTION

What is proposed herein below is directed to addressing one or more of the above-discussed concerns.

The Jaeger study presents a method that depends on tissue deformation and is therefore anatomy specific, unusable for example on the rib cage. Axial deformation is an alternative proposed in the Jaeger study, and preferable in that coupling gel can be used, but is still anatomy specific. Both forms of deformation yield varying results clinician to clinician and instance to instance.

What is proposed herein below relates to an automatable and relatively simple to implement technology that affords reproducibility of results and is less demanding on the skill and experience of the user.

In an aspect of what is proposed herein, a medical image acquiring device repeatedly photoacoustically images an intracorporeal spatial location that is common image to image. This is done by varying the ultrasound transducer acquisition location, light excitation, or both, image to image so that averaging of the images serves to decorrelate background clutter that exists in the images, of the spatial location, individually.

In complementary, related aspects, for medical image acquisition a computer readable medium or alternatively a transitory, propagating signal is part of what is proposed herein. A computer program embodied within a computer readable medium as described below, or, alternatively, embodied within a transitory, propagating signal, has instructions executable by a processor for performing a plurality of acts, from among which is the act of: repeatedly photoacoustically imaging an intracorporeal spatial location that is common image to image, varying at least one of ultrasound transducer acquisition location, and light excitation, image to image so that averaging of the images serves to decorrelate background clutter that exists in the images, of said spatial location, individually.

In another aspect of what is proposed herein below, a photoacoustic apparatus that has an array of multiple ultrasound transducer elements is configured for forming receiving apertures at different positions along the array and for, based on what is temporally a single pulse of light and based on imaging acquisition by ones of the apertures that is responsive to the pulse, performing spatial compounding to produce a photoacoustic frame from the single pulse.

Details of the innovative photoacoustic background clutter reduction technology are set forth further below, with the aid of the following drawings, which are not drawn to scale, and in which the reference numerals pertain to the same or similar structures throughout the several views.

BRIEF DESCRIPTION OF THE DRAWINGS

Fig. 1 is a set of schematic diagrams of respectively an exemplary photoacoustic medical image acquiring device, and a particular embodiment, in accordance with the present invention;

Fig. 2 is a schematic and conceptual diagram of another particular embodiment of an exemplary photoacoustic medical image acquiring device and an example of its use, in accordance with the present invention; and

Fig. 3 is a schematic diagram of a face view of an exemplary photoacoustic medical image acquiring device, with side views of different embodiments for the probe, in accordance with the present invention.

DETAILED DESCRIPTION OF EMBODIMENTS

Fig. 1 depicts, by way of illustrative and non-limitative example, a photoacoustic (PA) medical image acquiring device 100. The device 100 includes an imaging probe 102, such as an integrated photoacoustic-ultrasound imaging probe, and a processor 103. It further includes a pulsed light source (not shown), typically of pulse length less than 20 nanosecond (ns), such as a high-energy laser, for illuminating body tissue and exciting PA waves. Also included is an elongated light delivery system 104 extending from the light source to, and into, the imaging probe 102. The light delivery system 104 includes an optical fiber bundle 106 and optionally, for example, optical components that couple light into the fiber optic bundle. The probe 102 includes an ultrasound transducer 108, here implemented with a linear array or matrix array. The probe 102 also includes a patient-interface surface 110, as part of its housing 112, for contacting a body 114 of a medical subject such as a patient, human or animal. The patient-interface surface 110 has a light-emitting location 115. The two bifurcated 116 legs of the fiber bundle 106 flank both elevational 118 sides of the transducer 108, and terminate near the patient-interface surface 110 in two, respective, spaced-apart excitation sources 120. The fiber optic bundle 106 is made up of multiple sub-bundles arranged in parallel. The sub-bundles are each bifurcated 116, the parallel arrangement extending the legs laterally 122 on each elevational 118 side of the transducer 108.

Typically, a single laser shot is used to reconstruct one photoacoustic imaging frame 123. Light is strongly scattered in biological tissue, so one laser shot can illuminate a broad region that covers the imaging field of view. Modern ultrasound systems have multi-channel, parallel data acquisition. So the photoacoustic imaging frame rate is usually limited by the laser repetition rate. Pulsed lasers with repetition rates of 30-50 Hz or higher can offer frame rates with minimal motion between consecutive frames.

The transducer 108 has an array 124, here a linear array although it may be a matrix array. The transducer 108 has a number of active receiving, or “receive”, apertures 125, with each aperture typically comprising multiple transducer elements 126 of the array 124, such as eight. The active receiving apertures 125, corresponding to an image acquisition location of the transducer 108, span the transducer laterally 122. They may also overlap, by half the aperture length, for example. The apertures 125 receive in parallel the radiofrequency (RF) data returning as a result of the laser pulse.

The acquisition location is the location, with respect to the probe 102, at which the PA wave is sensed on the transducer 108.

Two different positions 127 are shown in Fig. 1 for respective apertures, which may correspond to an aperture translation along the array 124.

The desired or “true” PA signal is elicited by a light pulse 128 whose spatial spreading 129 begins upon leaving the excitation sources 120, denoted in Fig. 1 with the boxed number “1.” The single light pulse 128 is temporally a single light pulse despite its issuance from two spatially different excitation sources 120, since the issuance from the two sources is simultaneous. The light passes through skin 130 of the patient and is absorbed by a PA target at a location 132 within the body 114. The resulting thermoelastic expansion of the target, represented in Fig. 1 by the concentric dotted lines, generates an acoustic wave that results in the desired PA signal being received by the transducer 108. The differential strength and differential timing of the signal with respect to elements 126 of the set of one or more active receiving apertures 125, allows for the creation of a PA image 123 representative of the nature and location of PA targets within the field of view of the transducer 108.

Problematically, however, optical absorption near a skin surface 134 is caused by melanin and superficial blood vessels136. Specifically, the optical absorption leads to photoacoustic waves that propagate directly 138 to the receiving apertures 125 and propagate into deeper tissue of the body 114, as well. As the photoacoustic waves propagate into the tissue, acoustic scatterers generate echoes, which are backscattered to the transducer 108. Both the directly coupled PA waves and backscattered "PA echoes" generate background clutter signals that degrade photoacoustic image contrast. These signals are not random, but rather deterministic and are therefore not effectively reduced by simple signal averaging. The signals appear spatially over the individual PA image 123 as areas of false brightness or intensity.

The clutter signal amplitude can be reduced, but not eliminated, by choosing an optical wavelength with low surface absorption. However, it is difficult to overcome the strong optical fluence at the skin surface and high melanin levels from skin pigmentation.

Also, wavelength selection may not be good option for imaging many light-absorbing targets or PA contrast agents.

The technique in the Jaeger study has the above-noted limitations.

Among the innovative techniques proposed herein, the fiber optic bundle 106 is, at the excitation sources 120, rocked angularly back and forth laterally. Thus, the rocking is with respect to the probe 102. The rocking is performed along the imaging plane of the linear array 124 or, for three-dimensional imaging, along the central lateral imaging plane of the matrix array.

This is an example of varying the light excitation without varying the imaging acquisition location of the transducer 108, i.e., the transducer remains stationary. Multiple PA images 123, e.g., ten, are acquired over the course of each angular rotation as represented by the two-headed arrow 140. The probe 102 which may be applied manually is likewise kept stationary, i.e., in place, during rotation. The rocking can be accomplished mechanically by means of a motor (not shown), within the probe 102, that is controlled by the processor 103. Or the fiber optic bundle 106 can, at excitation sources 120 for example, be mechanically rotated by means of a manually rotatable actuator, such as a thumbwheel, free-hand manipulable from outside the probe housing 112. If two back and forth motions via the thumbwheel can be done within a second, even tissue motion can be adequately captured by averaging ten frames acquired over the back or forth motion. The true PA signal adds coherently from consecutive frames since the scattered light reaching the target while the fiber bundle 106 is rocked is virtually identical. However, the photoacoustic echo signals are dependent on the surface optical absorption and backscattered signals from acoustic scatterers, which vary over the course of the rotation back or forth. The averaging of the frames causes cancellation of the backscattered signals of different frames and therefore serves to decorrelate the background clutter.

Advantageously, there is no need, during the repeated imaging, for shear deformation of body tissue. Likewise, there is no need, during the repeated imaging, for axial deformation of body tissue which is anatomically limited in its application. Neither kind of deformation leads to reproducible results or to a practical protocol for clinicians relatively experienced in the use of ultrasound.

In a similar, converse approach, the acquisition location varying entails the active receiving aperture(s) 125 being moved with respect to the probe 102. In the current example, the light-emitting location 115 on the patient-interface surface 110 of the probe 102 is stationary with respect to the probe. The movement of the aperture(s) 125 can therefore be regarded also as with respect to the light-emitting location 115.

More specifically, the aperture movement is done in the current example by rocking the transducer 108 back and forth laterally, i.e., along the imaging plane of the linear array 124 or, for three-dimensional imaging, along the central lateral imaging plane of the matrix array. There are a number of transducer-rocking versions. In each example presented herein below, the transducer is separated from the skin surface by a distance. That gap is filled by coupling fluid, coupling gel or a compliant gel standoff.

In one implementation, the probe 102 is configured with a fluid-fillable enclosure (not shown) in which the transducer is fixed for rotational, i.e., rocking, movement, the fluid serving as a coupling medium. The fluid, if containing water or an aqueous solution, is degassed, an example of which is provided in U.S. Patent Publication No. 2005/0154309 to Etchells et al., the entire disclosure of which is incorporated herein by reference. The fluid may also include mineral oils or hydro-gels. In addition to providing clearance for the transducer rocking, the fluid cools the probe 102 and the patient’s skin in contact with the probe.

The rocking varies the image acquisition location, while not varying the light excitation, as illustrated by example in Fig. 2. In particular, the active receiving aperture 125, or multiple currently active receiving apertures, are rocked while light delivery is not varied. The common intracorporeal spatial location 132 is imaged responsively one-to-one to the laser shots fired during a rotational swing. The transducer 108 has an ultrasound-receiving surface 204, i.e., "transducer surface", denoted in Fig. 2 by the line of alternating short and long line segments. Here, for illustration, it is assumed there are 2N + 1 PA images 208 for spatial alignment. The center image in the swing from one rotational extreme to the other corresponds to N = 0, and is labeled I_0. In particular, B-mode acquisition is interleaved, i.e., alternated, with PA acquisition, so that, for example, each PA frame is immediately followed by a corresponding B-mode frame acquired with the same transducer orientation. In addition, an extracorporeal structure 212 is employed for determining any in-plane displacement of the transducer 108. The extracorporeal structure 212 can be elongated, locationally fixed with respect to the probe during acquisition of the B-mode images and equivalently with respect to the axis of rotation, and disposed parallel, or otherwise transverse, to the axis. The extracorporeal structure 212 should be distinctively echogenic, like metal. Since the rotational axis can be regarded as normal to the sheet of Fig. 2, the extracorporeal structure 212, such as a wire, appears in the imaging plane of a linear array 124, for example, as a point target. The point target within the B-mode image can be estimated by speckle tracking which is performed automatically and without the need for user intervention. The device 100 identifies, in the B-mode image, a fixed, i.e., motionless, landmark, such as a vein or bone, within the anatomy, which can be done from a few preliminary frame to frame pattern matching comparisons. From the spatial relationship between the point target and the landmark, the in-plane displacement, if any, is determinined. If the device 100 is used for imaging parts of the anatomy that are motionless, the landmark is not needed. In this case, the position of the point target with respect to the anatomy, as derived from the B-mode image, indicates the in-plane displacement if any. This can entail co-registering the differently oriented B-mode images by pattern-matching on the anatomy. In either the case of motionless or moving anatomy, from the B-mode imaging, a skin surface orientation relative to the transducer surface 204 can be determined automatically and without the need for user intervention. In particular, a rotational angle 216 between the skin surface 134 and the transducer surface 204 is measured. By combining the determined rotational angle 216 and the determined point target displacement, a rigid transformation model can be derived for application to each PA image 208 to register it with respect to the reference imaging geometry which is usually set at the central transducer position corresponding to the PA image I_0 in Fig. 2. Spatial compounding of the PA images, to reduce background clutter, entails applying the rigid image-specific transformations, and averaging the overlapping areas to form the final compound image. Because of the one-way propagation model applied in receive beamforming, the true PA signal will remain correlated and unchanged at the same location; whereas, the PA-echo and clutter noises will spread out uncorrelated among different frames -- hence their signal level will be reduced after averaging out. Compared to the deformation-compensated scheme in Jaeger study, the instant scheme has the advantage of easy image registration for aligning individual component images. In particular, local motion tracking for each pixel in the entire imaging plane is not needed. The wire 212 may alternatively reside in a compliant stand-off pad that is usable in lieu of coupling gel, the pad being firmly attached to the probe housing.

In a variation on the above embodiment, the extracorporeal structure 212 is separate from the probe 102, and is held stationary with respect to the skin surface 134 instead of with respect to the probe. This alleviates the need to find a motionless structure as a landmark in the case of moving anatomy. The wire, or other extracorporeal structure 212, may be fixed to, and suspended in, a water bath or the stand-off pad. Inadvertent motion while holding the probe is not transferred to the water bath or pad, since the latter remain stationary.

Instead of the lateral rocking of the transducer 108 or of the excitation sources 120, the angular rocking may be elevational.

Also, instead of rotationally rocking the transducer 108, the probe 102 can be configured to, in another way, vary the image acquisition location. The probe 102 may, for example, be configured for other movement of the transducer 108, such as translation, that provides a different viewing angle. An example is a translational rocking back and forth, laterally or elevationally.

In an alternative embodiment, that depends on electronically formed apertures, the frame 123 is acquired in response to a single issued light pulse 128. In particular, all of the elements 126 of the array 124 simultaneously sense the RF data generated in reaction to excitation by the light pulse 128. Different apertures 127, overlapping as described above, provide the viewpoint variation. In particular, a photoacoustic apparatus that has an array 124 of multiple ultrasound transducer elements 126 is configured for forming receiving apertures 127 at different positions along the array and for, based on what is temporally a single pulse of light 128 and based on imaging acquisition by ones, or all, of the apertures that is responsive to the single pulse, performing spatial compounding to produce a photoacoustic frame 123 from the single pulse. Specifically, images that are reconstructed from different apertures 127 are spatially compounded. Advantageously, frame acquisition and processing is rapid, and background clutter is reduced as in the previously described embodiments.

The success of spatial compounding depends on the degree of statistical independence among the sequential images that will be combined for averaging. To ensure statistically independent speckle patterns, the optimal aperture translation step, based on pulse-echo ultrasound imaging, should be equal to approximately half its length in the lateral direction. Thus, for an aperture length of eight transducer elements 126, a displacement between consecutively adjacent active receiving apertures 125 might be set at four elements. Nevertheless theoretical and experimental results have demonstrated that even partially correlated apertures can reduce clutter noise. The optimum aperture displacement for photoacoustic imaging can be derived from auto-correlation analysis. Since the translation of the active receiving aperture 125 is controlled by the ultrasound device 100, PA images 123 acquired at different aperture positions can be accurately registered and the overlapping area can be averaged together to form the final compound image. Spatial compounding can be expanded from a one-dimensional to a two-dimensional array 124 for more flexibility in aperture selection and translation. This embodiment enjoys the same advantage of easy image registration for aligning individual component images, in comparison to the technique in the Jaeger study.

As a variation of the above alternative embodiment, the acquisition location varying entails the active receiving aperture 125 being moved electronically, more generally with respect to the probe 102 and for example with respect to the light-emitting location 115 on the patient-interface surface 110 of the probe, in synchrony with repeated PA imaging. The movement with respect to the probe 102 serves as an alternative form of varying the acquisition location while not varying the light excitation. As in the embodiment above that moves the entire transducer 108, this embodiment repeatedly photoacoustically images the common intracorporeal spatial location 132 in the course of performing the PA excitation-acquisition sequence so that the target is viewed from different angles. In the current example however, the device 100 is configured for varying the acquisition location by electronically translating the aperture 125, the location varying being likewise performed in synchrony with the repeated imaging.

Referring again to the embodiment shown in Fig. 1, the device 100 can be modified in different ways to vary the light excitation.

On the one hand, more generally, an integrated photoacoustic-ultrasound imaging probe configured for varying light excitation without varying acquisition location incorporates light delivery into the probe housing or has light delivery attached to the probe. Signal averaging, as in the above-discussed embodiments, is performed over multiple frames to reduce background clutter.

On the other hand and for example, as seen in Fig. 3, an integrated photoacoustic-ultrasound imaging probe 300 may include in addition, or instead of, an optical fiber bundle 302 (the constituent sub-bundles 304 being shown in Fig. 3), free space optics 306 and/or a light guide (or "light pipe") 308. The optical beam pattern is moved slightly, for example via translation 310, rotation 312 and/or vibration 314, relative to an ultrasound array 316 of a transducer of the probe 300 during imaging. A motor (not shown) is incorporated within the probe to provide the movement. The movement of the optical beam is such that consecutive imaging frames provide substantial clutter reduction. A sufficient amount of movement can be derived empirically. Thus, the excitation varying entails moving excitation sources 318 which are disposed near an output end 320 of a light delivery system 322. In the case of free space optics 306, this may involve moving, e.g., further apart or closer together, PA image to PA image a pair of curved mirrors 324, 326 that receive light input from a bifurcated deflection mirror 328 illuminated by a laser 330. Movement image to image may alternate movement together with movement apart with each subsequent image, but need not do so and may, for example, transition by degrees of movement in one direction for a while. The light travels a path 331 indicated by the parallel-stem arrows.

In another embodiment, the device includes an optical mask 332, an engineered diffuser 334, or the both the mask and the diffuser. Here, too, the excitation varying entails spatially altering a beam 336 by physically moving 337, via a motor and PA image to PA image, the mask 332 and/or diffuser 334 into and out of the light path, entirely or to varying degrees. One or both of the beam altering components 332, 334 can be disposed either at a distal end 338 of the light delivery system 322; or a proximal end 342 of the light delivery system 322, for example at an input side of the fiber bundle or light guide after the connection of the laser 330 and before entry into the probe 300. Or any combination of the components 332, 334 can be disposed at either the distal or proximal end. Disposal at the proximal end 342 has the advantage that the component 332, 334 need not be made part of the probe 300 whose design is limited by more factors than those of the part of the device 100 on the other side of the cable.

An alternative embodiment switches 344 the optical illumination between the two fiber bundles placed on opposite sides of the ultrasound array 316. In this embodiment, two separate, spaced-apart fiber bundles rather than a single, bifurcated fiber bundle can be used. A fast scanning mirror galvanometer can be used to switch the input beam from one fiber bundle to the next fiber bundle on consecutive laser shots. In effect, an active state with respect to illumination is switched from one source of the excitation to another source of the excitation, the two sources being spaced apart. In the present embodiment, the switching toggles back and forth between two such sources. The final photoacoustic image displayed can be taken as the geometric mean of the two images acquired with the right and left fiber bundles separately. Equivalently, the arithmetic mean of logarithmic values of the two images may be taken.

An example of varying both the light excitation and the acquisition location, image to image, would be toggling back and forth between the two separate fiber bundle branches or sources image to image while also translating an active receiving aperture 125 image to image.

What is proposed herein above is applicable in a number of clinical applications, including cancer diagnosis, staging, and therapy monitoring.

A medical image acquiring device repeatedly photoacoustically images an intracorporeal spatial location that is common image to image, varying at least one of ultrasound transducer acquisition location, and light excitation, image to image so that averaging of the images serves to decorrelate background clutter that exists in the images, of the spatial location, individually. The varying of one may be in the absence of the other, and may entail movement. The movement may be of an optical fiber bundle, a light guide or free space optics, as by rotation, translation or vibration. It may be of an optical mask or engineered diffuser, or it may entail toggling between active states of two spaced apart, separate optical fiber branches near the probe surface. If the movement pertains instead to acquisition location, it may imply translation of the active receive aperture, or rotation of the transducer as in the imaging plane in which case co-registering of the images may use synchronous B-mode images to detect skin orientation and a landmark.

While the invention has been illustrated and described in detail in the drawings and foregoing description, such illustration and description are to be considered illustrative or exemplary and not restrictive; the invention is not limited to the disclosed embodiments.

For example, the averaging of frames may be weighted to favor those acquired at more central acquisition locations. Nor is it required that the spatial compounding image transformations for co-registration be rigid.

Other variations to the disclosed embodiments can be understood and effected by those skilled in the art in practicing the claimed invention, from a study of the drawings, the disclosure, and the appended claims. In the claims, the word "comprising" does not exclude other elements or steps, and the indefinite article "a" or "an" does not exclude a plurality. Any reference signs in the claims should not be construed as limiting the scope.

A computer program can be stored momentarily, temporarily or for a longer period of time on a suitable computer-readable medium, such as a floppy disk, a magnetic hard disk drive, a solid-state medium such as a solid state hard disk, flash memory, a USB thumb drive, read-only memory (ROM), an optical storage medium such as an optical disk, and a magneto-optical disk. Examples of optical disks include compact disks (CD) and digital versatile disks (DVD), for example CD-ROM, CD-RW, CD-R, DVD-ROM, DVD-RW, or DVD-R disks. Such a computer-readable medium is non-transitory only in the sense of not being a transitory, propagating signal, but includes other forms of computer-readable media such as register memory, processor cache, RAM and other volatile memory.

A single processor or other unit may fulfill the functions of several items recited in the claims. The mere fact that certain measures are recited in mutually different dependent claims does not indicate that a combination of these measures cannot be used to advantage.

CLAIMS:

1. A medical image acquiring device configured for repeatedly photoacoustically imaging an intracorporeal spatial location that is common image to image, varying at least one of ultrasound transducer acquisition location, and light excitation, image to image so that averaging of the images serves to decorrelate background clutter that exists in the images (208), of said spatial location, individually.

2. The device of claim 1, said varying comprising the varying of said acquisition location (125).

3. The device of claim 2, configured for, in said varying of said acquisition location, not performing said varying of said excitation.

4. The device of claim 2, further configured for having an active receiving aperture and for electronically translating said aperture (127), the location varying comprising said translating.

5. The device of claim 2, comprising:

an imaging probe (300), said probe including a patient-interface surface and a light-emitting location on said surface, and further including an ultrasound transducer configured with a receiving aperture,

said device being configured for said aperture being moved with respect to said light-emitting location, the acquisition location varying comprising the moving.

6. The device of claim 5, comprising a motor, said moving comprising moving said transducer (108) via said motor.

7. The device of claim 5, said transducer having a surface (204), said averaging comprising measuring an angle between the transducer surface and said patient-interface surface.

8. The device of claim 5, configured for said aperture being mechanically rotated, said moving comprising mechanically rotating said aperture.

9. The device of claim 8, configured for said rotating along (122) an imaging plane.

10. The device of claim 8, configured for said averaging, said averaging comprising tracking, automatically and without need for user intervention, in images, a particular extracorporeal structure (212) to determine displacement, if any, of said probe.

11. The device of claim 10, said structure being elongated, locationally fixed either with respect to said probe during acquisition of the tracked images or with respect to skin (130) of a subject of said imaging where said probe is applied to said skin, and disposed transverse to an axis of said rotating.

12. The device of claim 1, configured for said averaging, said averaging comprising determining, automatically, without need for user intervention, from imaging via said transducer, a skin surface orientation (216).

13. The device of claim 1, said varying comprising the varying (144) of said excitation.

14. The device of claim 13, configured for, in said varying of said excitation, not performing said varying of said acquisition location.

15. The device of claim 13, said device comprising sources (120) of said excitation that are spaced apart, said device being configured for switching an active state from one of said sources to another, image to image.

16. The device of claim 15, said switching comprising toggling back and forth between two of said sources.

17. The device of claim 13, configured for emitting a light beam (128) for said excitation, the excitation varying comprising spatially altering said beam.

18. The device of claim 17, said device comprising an optical mask (332), and engineered diffuser (334), or both said mask and said diffuser, said device configured for said altering correspondingly via physically moving said mask and/or said diffuser.

19. The device of claim 13, comprising, for said excitation, at least one of an optical fiber bundle (106), free space optics and a light guide, said device being configured for physically moving one or more of said bundle, optics and guide, the excitation varying comprising said moving.

20. The device of claim 18, comprising a motor, said moving comprising, via said motor, at least one of translating, rotating and vibrating (314).

21. The device of claim 1, comprising an elongated light delivery system that has a distal end (338) for physically being applied to a body that contains said spatial location, said system being configured for delivering said excitation, said system having a proximal end and at least one of an optical mask and an engineered diffuser, said device being configured such that one or more of said at least one is disposed at correspondingly at least one of said proximal end and said distal end.

22. The device of claim 21, configured for the varying of said excitation, said varying of said excitation comprising spatially altering, via said one or more, a beam emitted by said system (322).

23. A computer readable medium embodying a computer program for medical image acquisition, said program embodying a computer program having instructions executable by a processor for performing a plurality of acts, from among which is the act of:

repeatedly photoacoustically imaging an intracorporeal spatial location (132) that is common image to image, varying at least one of ultrasound transducer acquisition location, and light excitation, image to image so that averaging of the images serves to decorrelate background clutter that exists in the images, of said spatial location, individually.

24. A photoacoustic apparatus comprising:

an array (124) of multiple ultrasound transducer elements,

said apparatus configured for forming receiving apertures at different positions along said array and for, based on what is temporally a single pulse of light and based on imaging acquisition by ones of said apertures that is responsive to said pulse, performing spatial compounding to produce a photoacoustic frame from said single pulse.

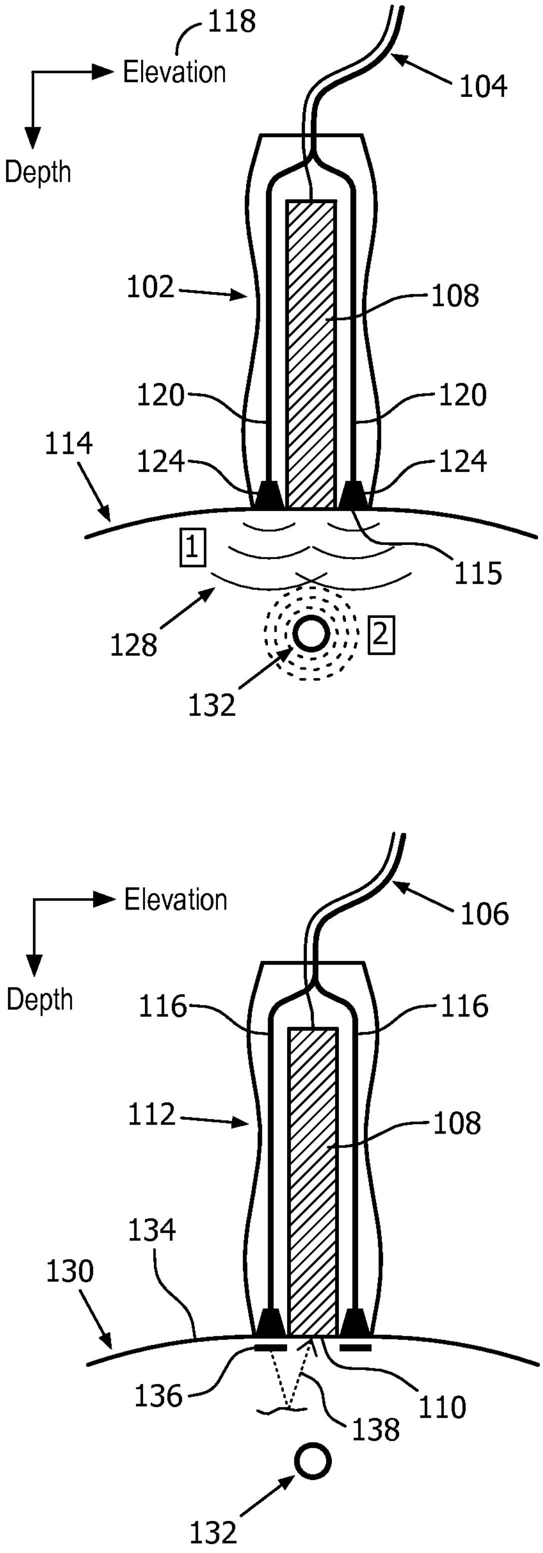
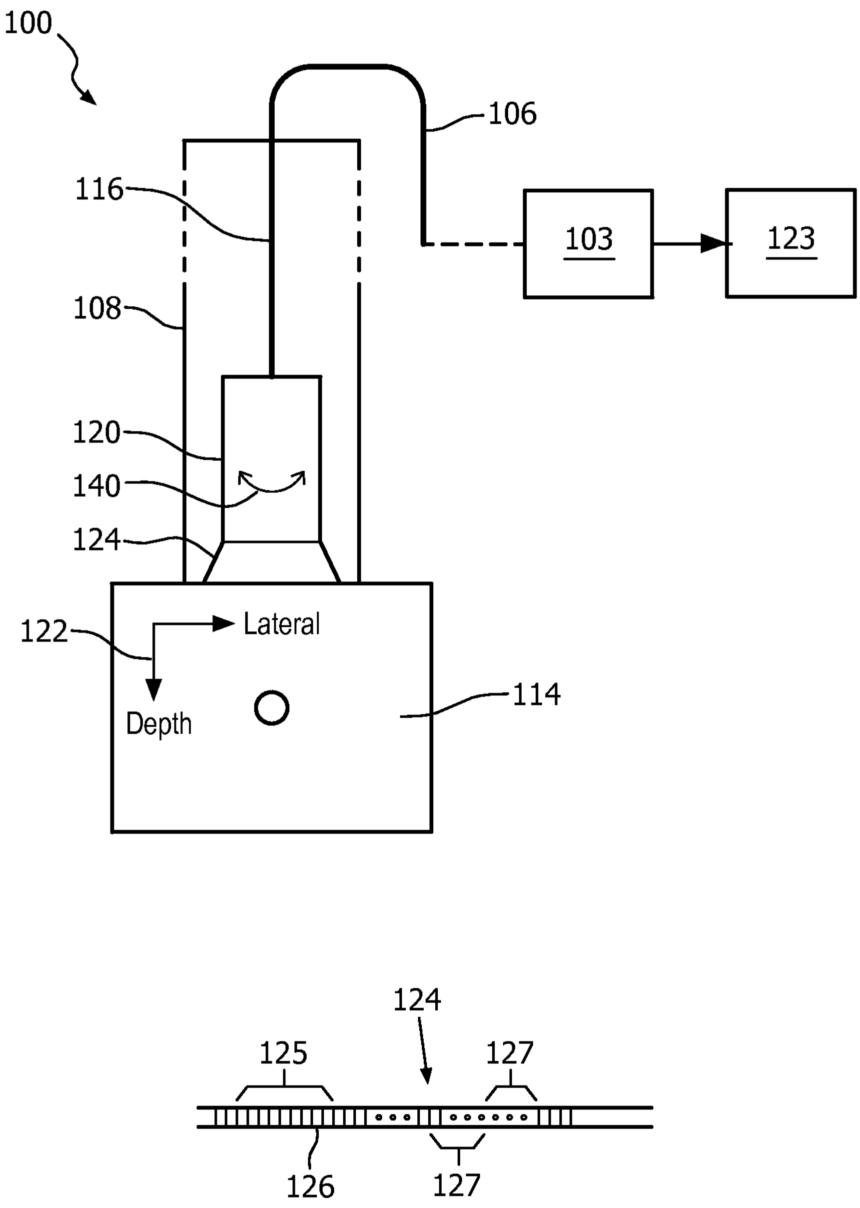

118
Elevation
Depth
104
102
108
120
120
114
124
124
1
115
128
2
132
Elevation
Depth
106
116
116
112
108
134
130
136
110
138
132

FIG. 1



(Continued)

FIG. 1

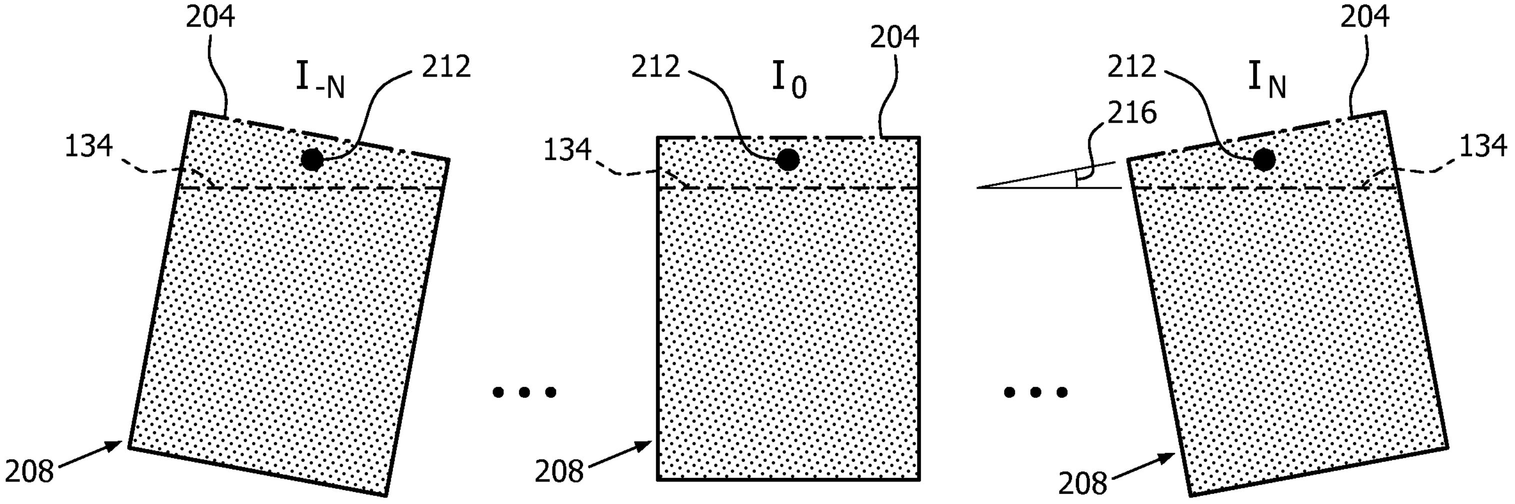


FIG. 2

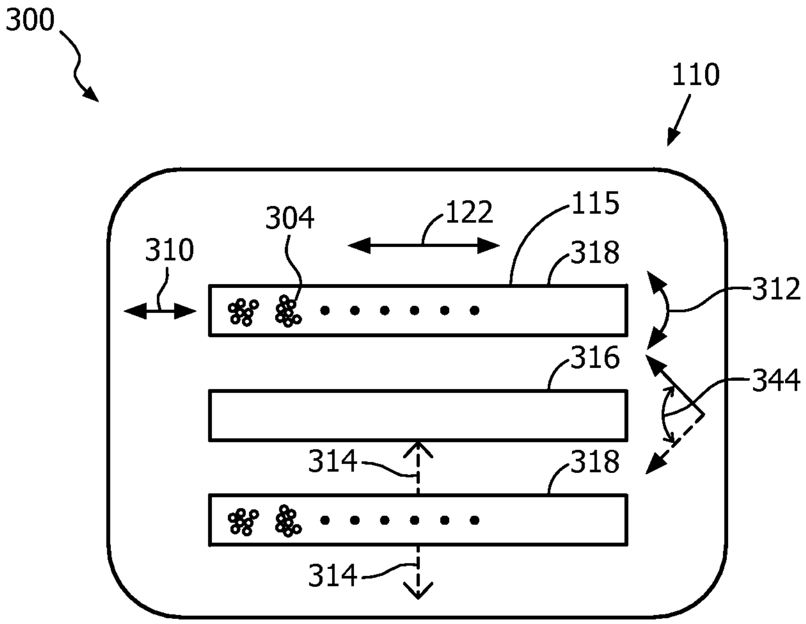
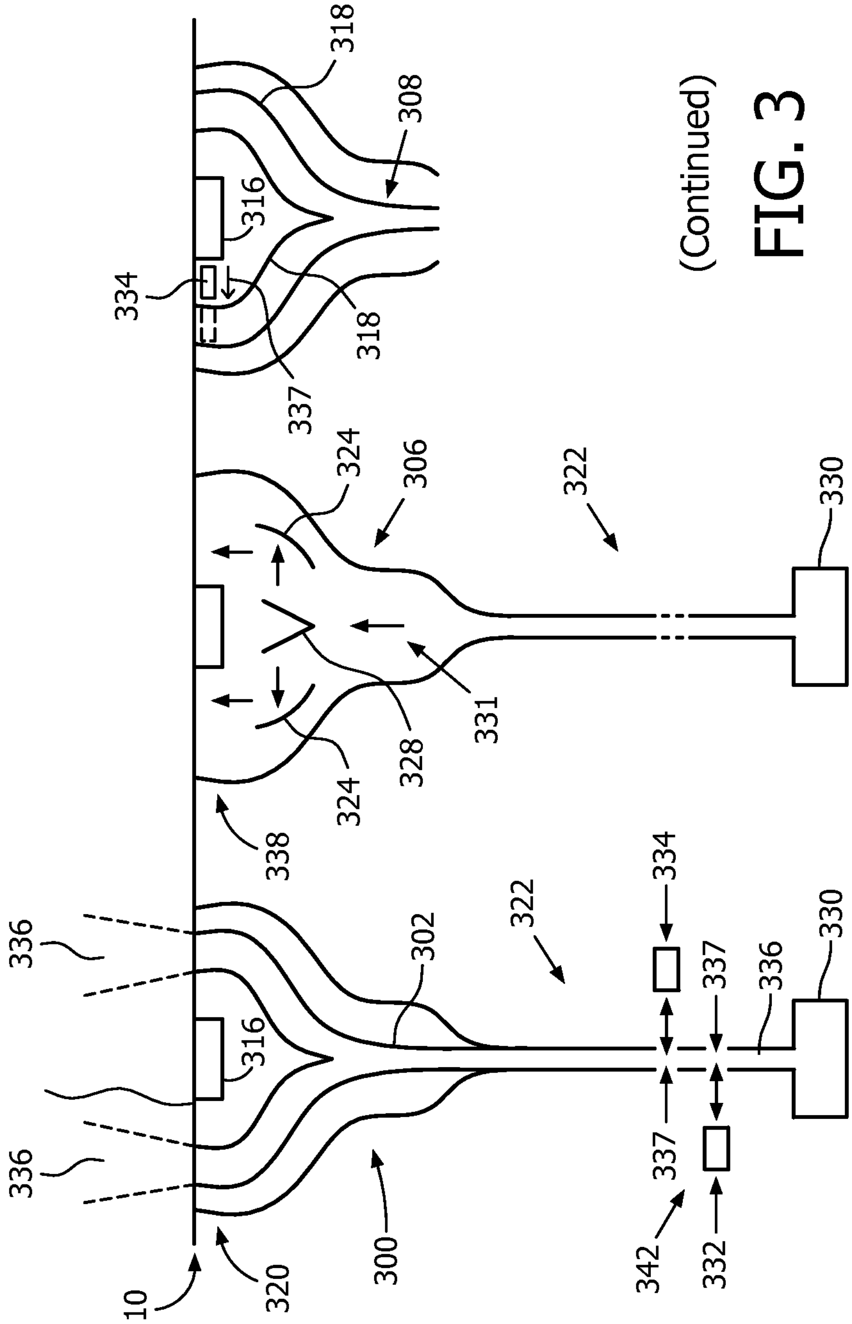


FIG. 3



(Continued)

FIG. 3