



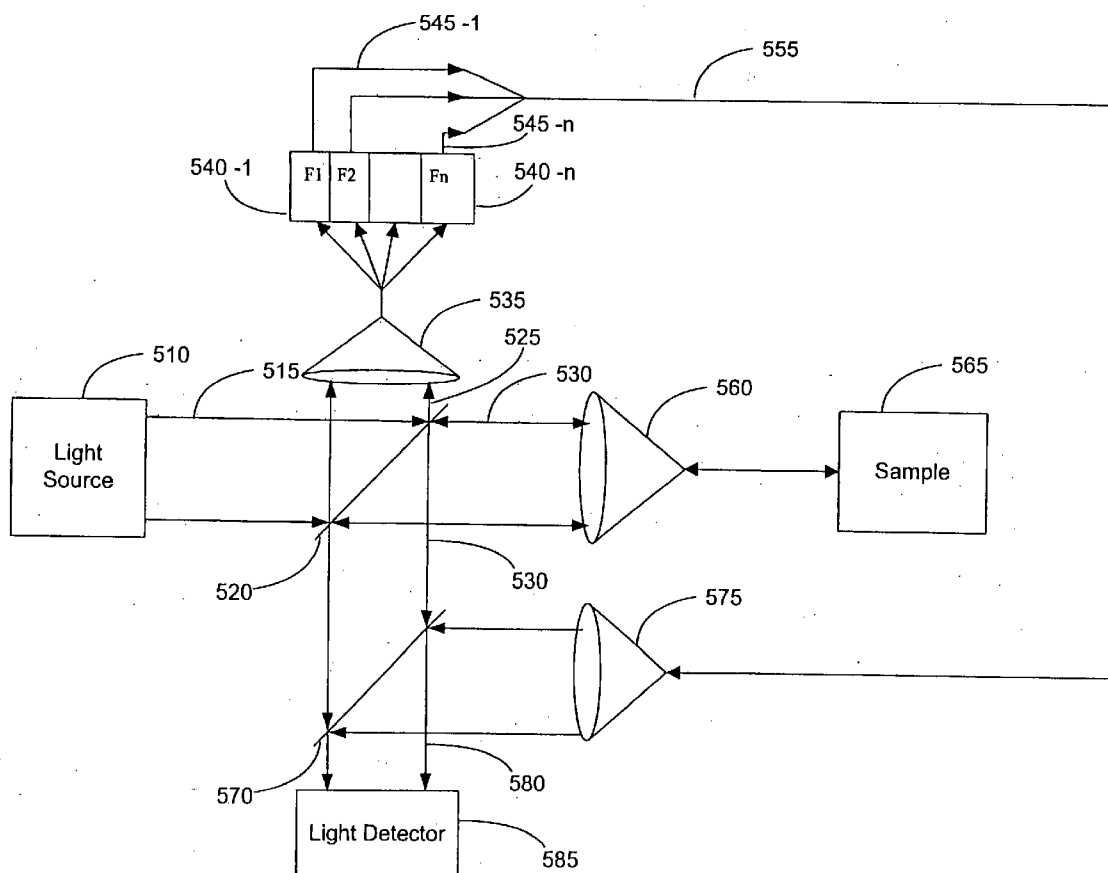
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(19) **United States**(12) **Patent Application Publication**
Barbato(10) **Pub. No.: US 2006/0106320 A1**(43) **Pub. Date: May 18, 2006**(54) **ELECTRONICALLY SCANNED OPTICAL
COHERENCE TOMOGRAPHY WITH
FREQUENCY MODULATED SIGNALS**09/906,903, filed on Jul. 16, 2001, now Pat. No.
6,728,571.**Publication Classification**(75) Inventor: **Louis J. Barbato**, Franklin, MA (US)(51) **Int. Cl.****A61B 6/00** (2006.01)(52) **U.S. Cl.** **600/478**

Correspondence Address:

ORRICK, HERRINGTON & SUTCLIFFE, LLP
IP PROSECUTION DEPARTMENT
4 PARK PLAZA
SUITE 1600
IRVINE, CA 92614-2558 (US)(57) **ABSTRACT**(73) Assignee: **SCIMED LIFE SYSTEMS, INC.**(21) Appl. No.: **11/284,684**(22) Filed: **Nov. 21, 2005****Related U.S. Application Data**(62) Division of application No. 10/690,948, filed on Oct.
21, 2003, which is a division of application No.

An improved Optical Coherence Domain Reflectometry (OCDR) system is provided. One embodiment of this OCDR system outputs a detector signal carrying image depth information on multiple modulation frequencies, where each modulation frequency corresponds to a different image depth. The image depth information from the detector signal may be resolved by tuning to the desired modulation frequency. Another system for imaging body tissue uses multiple frequency modulators such that the light beam does not travel from an optical fiber to free space.



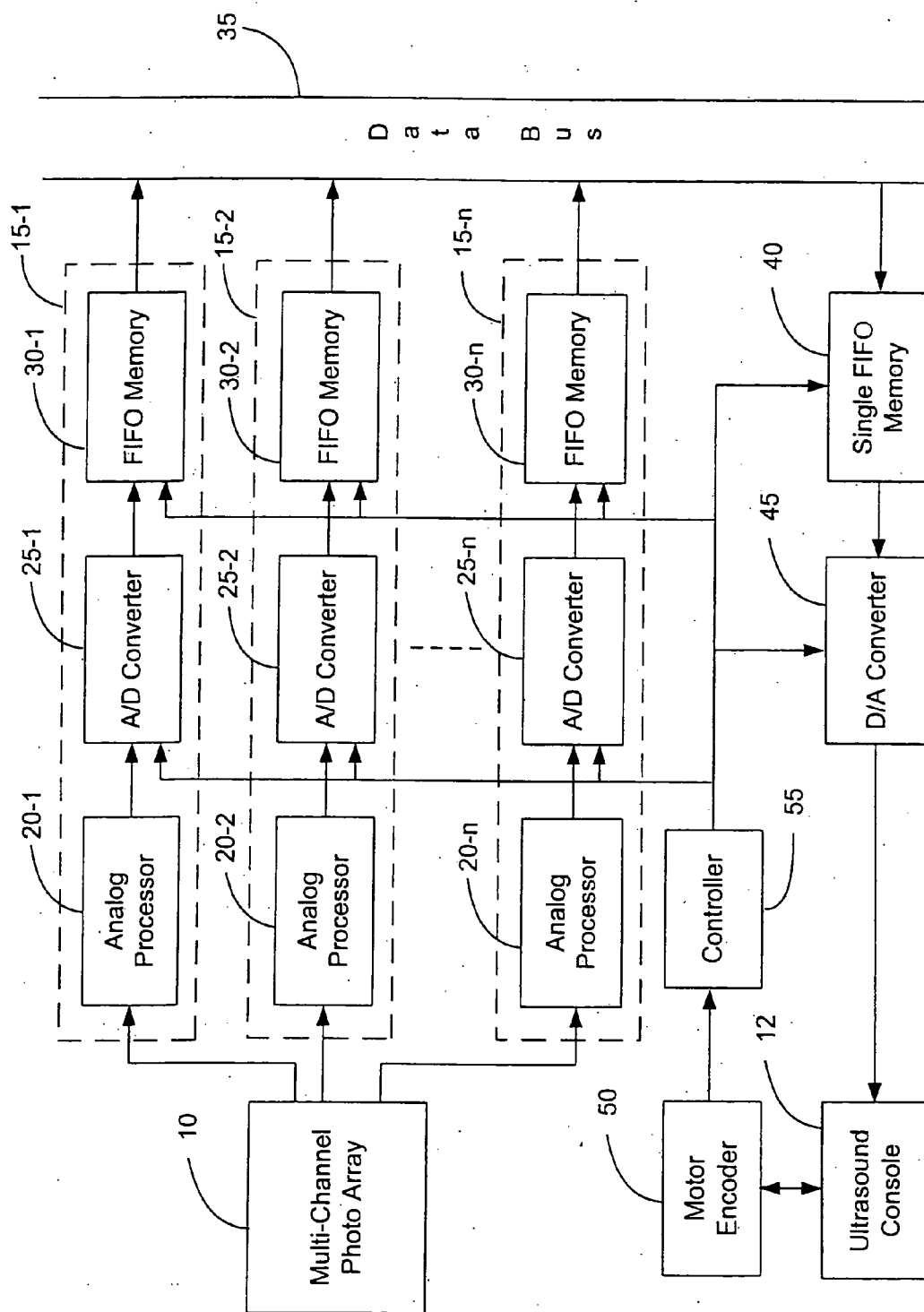


FIG. 1

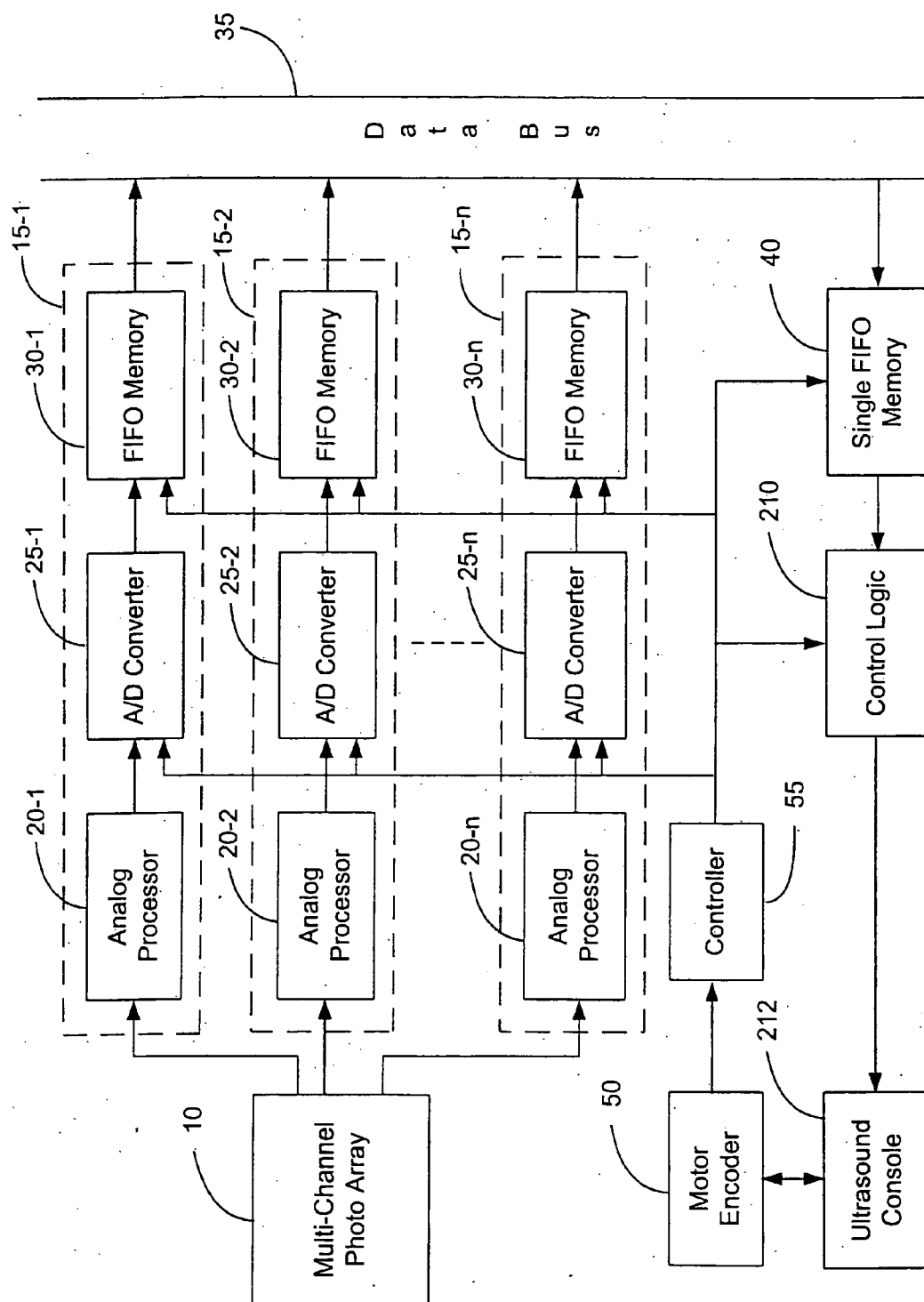


FIG. 2

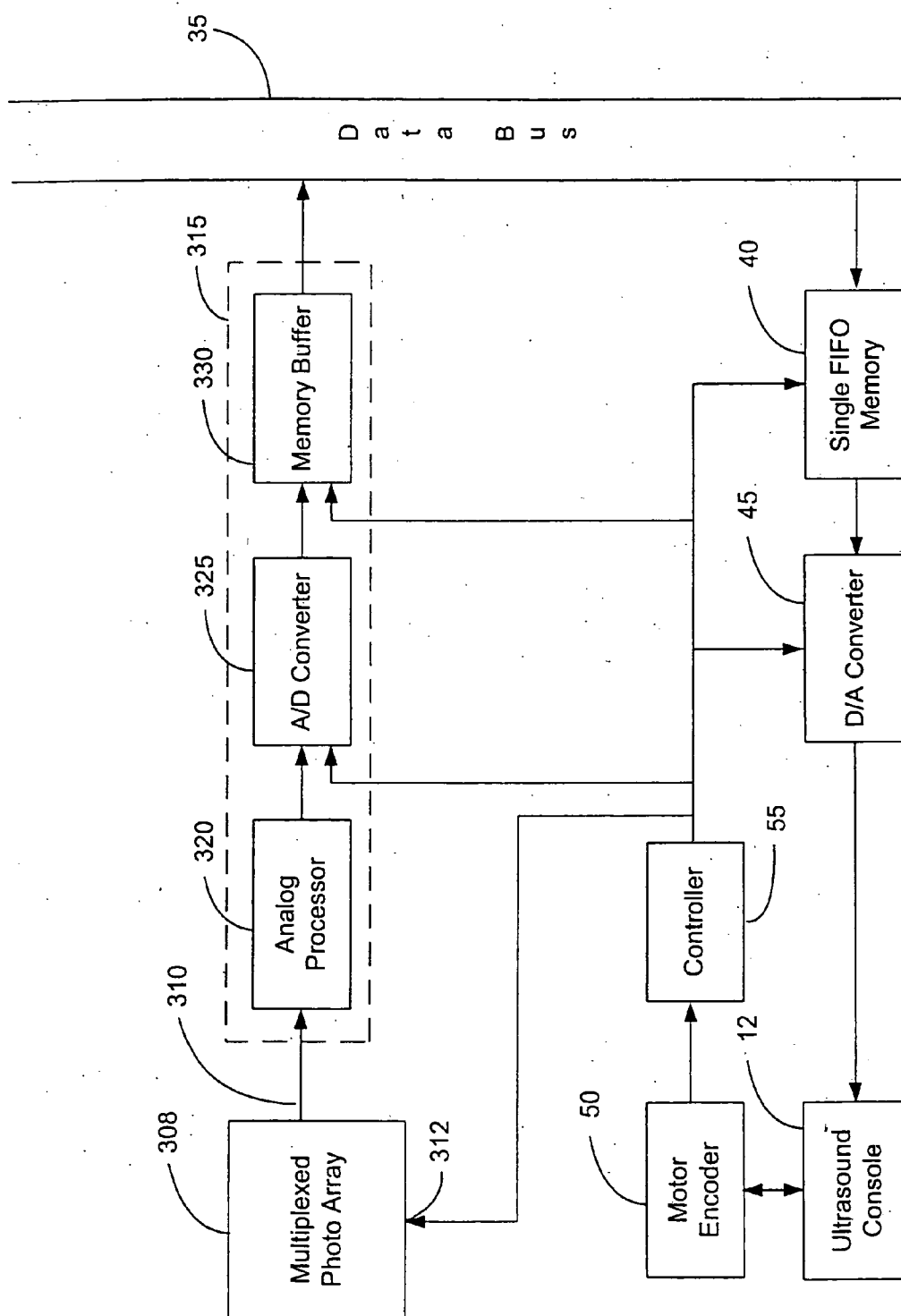


FIG. 3

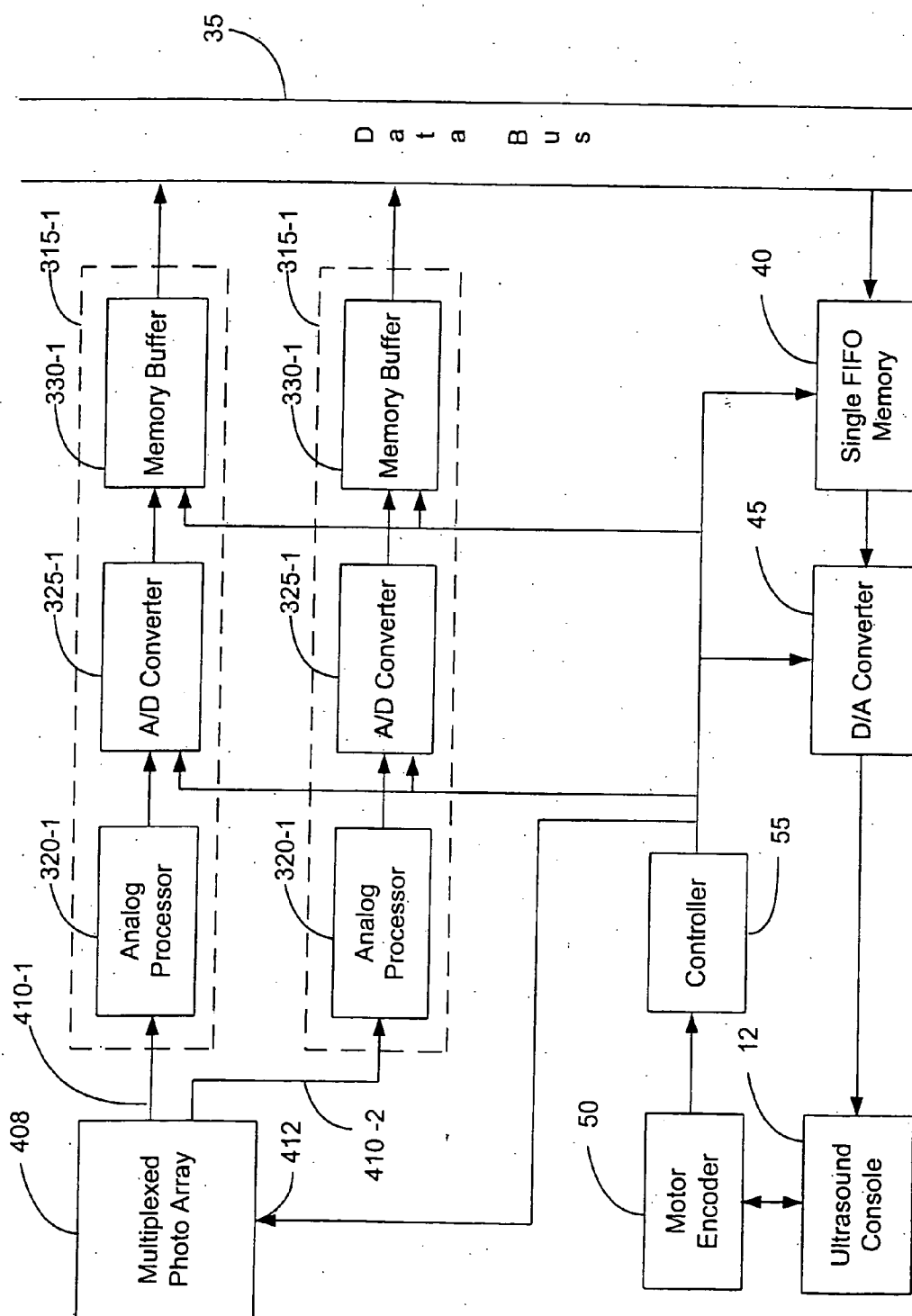


FIG. 4

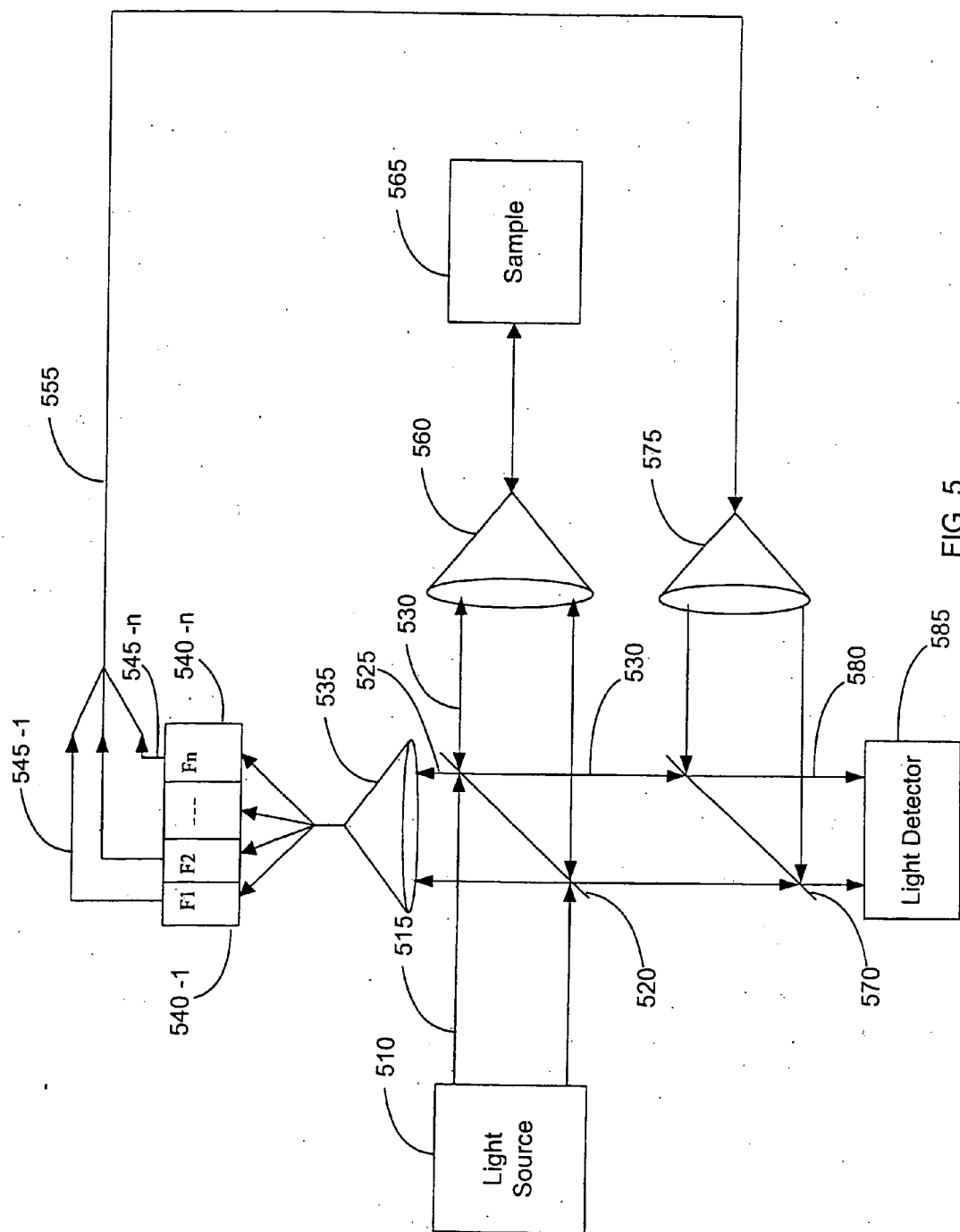


FIG. 5

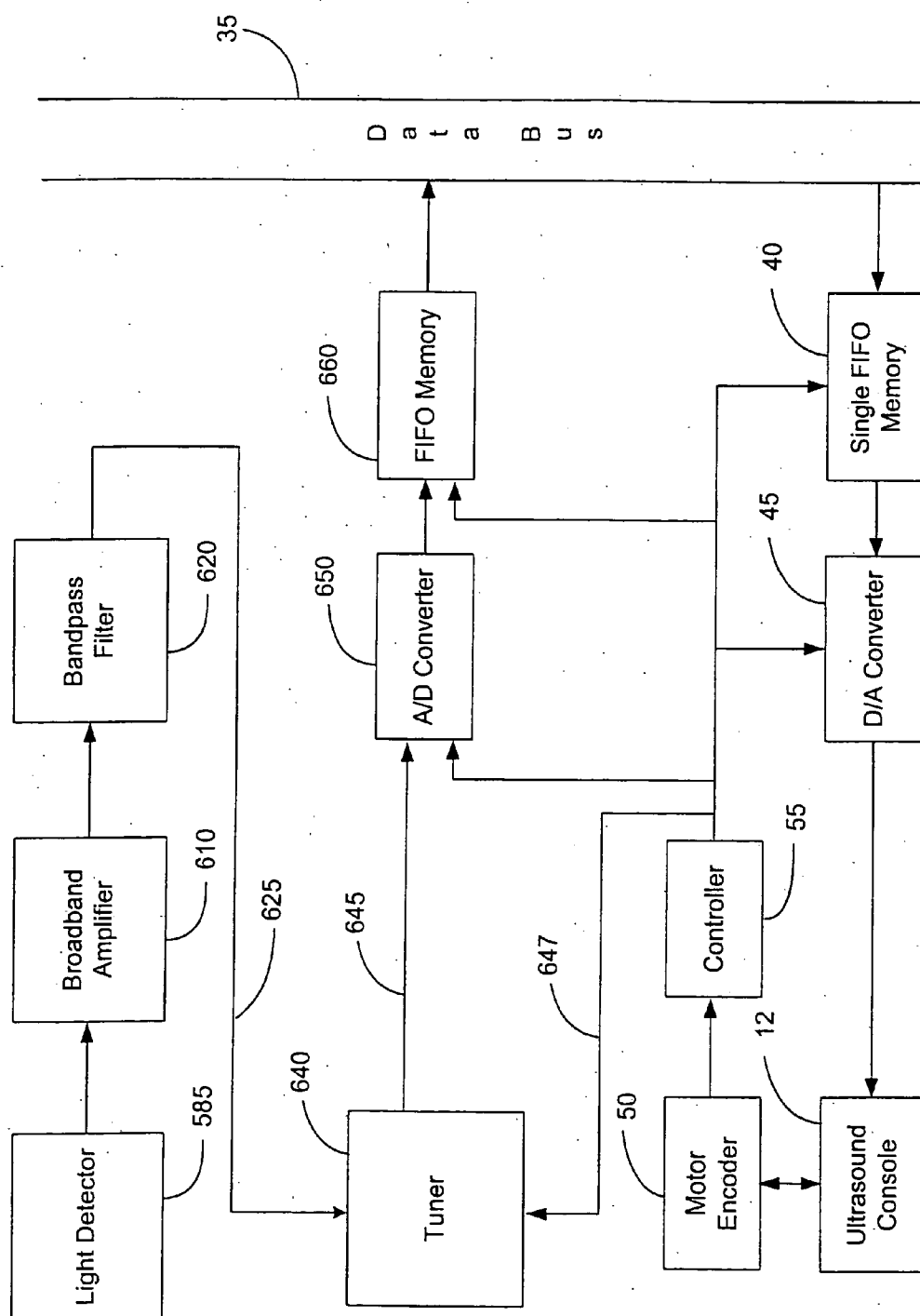
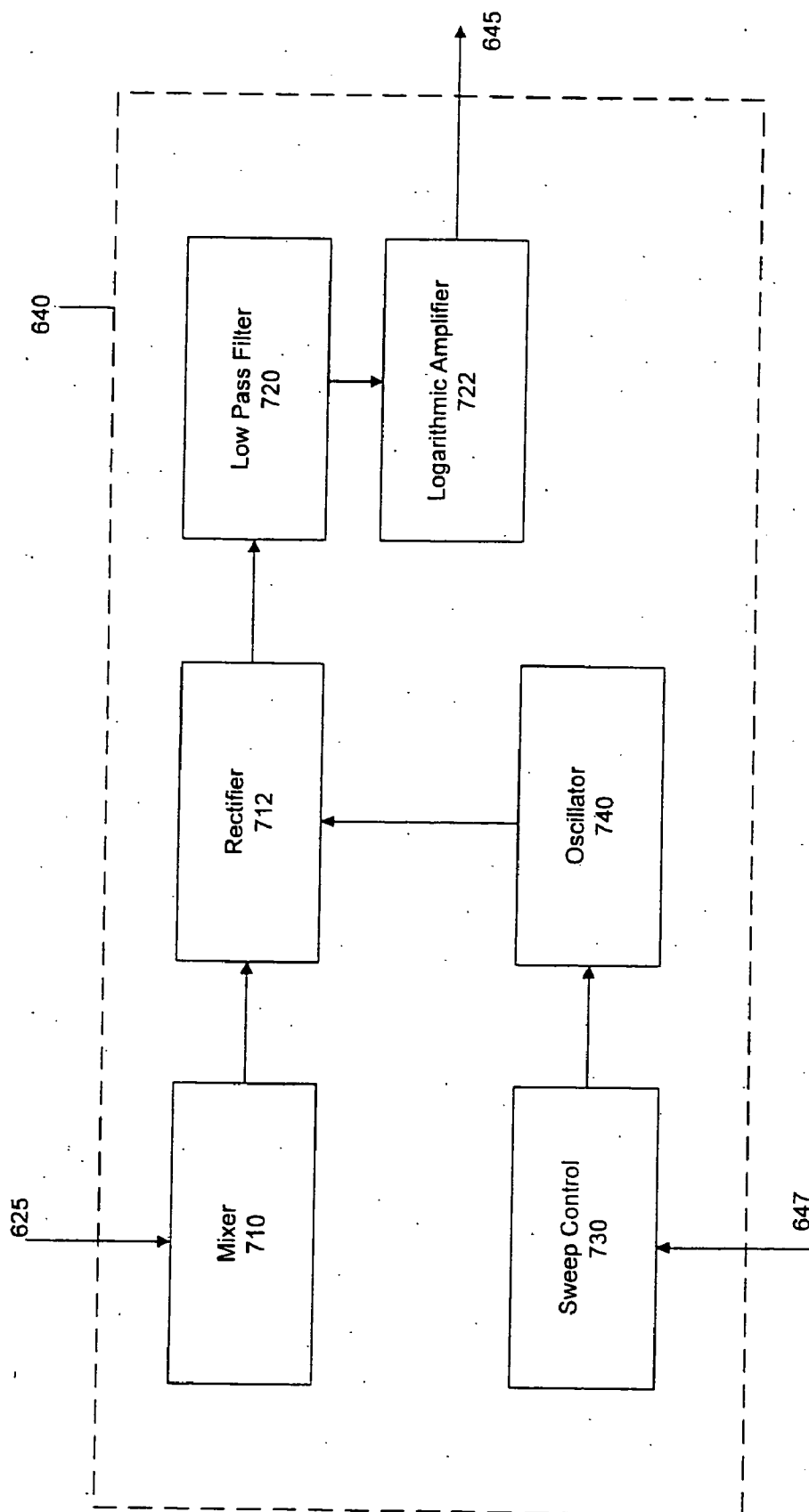


FIG. 6

FIG. 7



ELECTRONICALLY SCANNED OPTICAL COHERENCE TOMOGRAPHY WITH FREQUENCY MODULATED SIGNALS

CROSS REFERENCE TO RELATED APPLICATIONS

[0001] This application is a divisional of co-pending application Ser. No. 10/690,948, filed Oct. 21, 2003, which is a divisional of U.S. Pat. No. 6,728,571, filed Jul. 16, 2001, and all patents and patent applications of which are expressly incorporated herein by reference for all purposes.

FIELD OF THE INVENTION

[0002] The invention relates generally to a method and system for making optical measurements, and in particular, to a method and system for making optical measurements for a catheter-based imaging system such as Optical Coherence Domain Reflectometry (OCDR).

BACKGROUND OF THE INVENTION

[0003] Ultrasound medical imaging is often used to produce images of blood vessels and surrounding tissue. To image a blood vessel and surrounding tissue, an Intravascular Ultrasound (IVUS) catheter is typically used. The IVUS catheter comprises an elongated member and an ultrasound transducer located at a distal end of the elongated member. The elongated member is inserted into the blood vessel, and the ultrasound transducer is positioned at a desired location in the blood vessel. An ultrasound transducer is designed to transmit a specific resonant frequency, e.g., when it is excited by a pulse. The excite pulse signal causes the ultrasound transducer to emit ultrasound waves in the blood vessel. A portion of the emitted ultrasound waves is reflected back to the ultrasound transducer at tissue boundaries in the blood vessel and the surrounding tissue. The reflected ultrasound waves induce an echo signal at the ultrasound transducer. The echo signal is transmitted from the ultrasound transducer to an ultrasound console, which typically includes an ultrasound image processor and a display. The ultrasound console uses the received echo signal to image the blood vessel and the surrounding tissue.

[0004] In order to produce a radial cross-sectional image of a blood vessel and surrounding tissue, the ultrasound transducer is typically rotated along the axis of the elongated member. As the ultrasound transducer is rotated, the ultrasound transducer emits ultrasound waves in different radial directions. The resulting echo signals from the different radial directions are processed by the ultrasound console to produce a radial cross-sectional image of the blood vessel and the surrounding tissue. Alternatively, the ultrasonic transducer may be mounted in an assembly together with a reflective member (mirror), where the transducer emits ultrasonic energy in a substantially axial direction and the mirror is oriented to deflect the emitted ultrasonic energy in a radial direction.

[0005] The echo signal is a serial amplitude modulated signal, in which the amplitude of the signal varies with time. A typical echo signal has a time length of 8 μ s, which corresponds to an image depth of approximately 6 millimeters from the ultrasound transducer. The echo signal carries both image brightness information and image depth information, where depth may be taken with respect to the

ultrasound transducer. The image brightness information is provided by the amplitude of the echo signal. The image depth information is provided by the time position within the echo signal. For example, an earlier time position in the echo signal corresponds to a lower image depth than a later time position in the echo signal. This is because an ultrasound wave that is reflected back to the ultrasound transducer from a shallower depth reaches the ultrasound transducer before an ultrasound wave that is reflected back to the ultrasound transducer from a deeper depth. As a result, the ultrasound wave that is reflected back to the ultrasound transducer from the shallower depth has a shorter propagation delay time, which translates into an earlier time position in the echo signal.

[0006] Another imaging technique used to produce images of blood vessels and surrounding tissue is Optical Coherence Domain Reflectometry (OCDR). To image a blood vessel using OCDR, a fiber-optic catheter is inserted into the blood vessel. A proximal end of the fiber-optic catheter is coupled to an OCDR system. In the OCDR system, a laser generates a source beam. A beam splitter splits the source beam into a reference beam and a sample beam. The reference beam is diffracted by a diffraction grating into a diffraction beam. The sample beam is transmitted through the fiber-optic catheter and emitted in the blood vessel at a distal end of the catheter. Typically, the distal end of the catheter includes a prism for directing the sample beam into the blood vessel. A portion of the sample beam is reflected back to the distal end of the catheter by the blood vessel and the surrounding tissue. The reflected sample beam is transmitted to the OCDR system through the fiber-optic catheter. In the OCDR system, the reflected beam is mixed with the diffraction beam to produce a coherence-domain interference pattern, which is detected by an array of photo detectors.

[0007] The resulting interference pattern provides both image brightness information and image depth information, where depth may be taken with respect to the distal end of the catheter. The image brightness information is provided by the light intensity of the interference pattern. The image depth information is provided by the spatial position within the interference pattern. This is because the portion of the sample beam that is reflected back to the catheter from a certain depth in the body constructively interferes with the diffraction beam at a certain spatial position. Typically, the photo detectors of the photo array are arranged so that each photo detector detects the light intensity of the interference pattern at a certain spatial position within the interference pattern. Thus, the output of each photo detector provides image brightness information for a certain image depth. The photo array outputs parallel channels, where each parallel channel corresponds to the output of one of its photo detectors. The parallel channels of the photo array are inputted to an OCDR image processor to produce an image of the blood vessel and the surrounding tissue.

[0008] An advantage of the above-described OCDR system is that the array of photo detectors is able to capture image brightness information at multiple image depths in one instance. This enables the OCDR system to produce images at true video rates, e.g., 30 frames per second.

[0009] Optical coherence tomography (OCT) is an optical imaging technique, which achieves non-invasive, cross-sectional imaging of microscopic biological structures. OCT

is analogous to ultrasound imaging, only measuring the intensity of backscattered infrared light rather than sound. It can be implemented using high-speed fiber optics, which makes OCT compatible for interfacing with fiber optic endoscopes used for catheter-based imaging within the vascular system.

[0010] Mechanical OCT systems use a mirror mounted to a piezoelectric material to achieve mechanical scanning of depth information. However, the speed of these systems are limited by mechanical factors such as mass, the electromechanical properties of the piezoelectric material and the need to scan at a constant velocity. These limitations translate into scanning rates that are considerably slower than true video rate, such that real time viewing cannot be achieved.

[0011] In grating generated OCT systems, depth information is spatially translated across a beam of light using a diffraction grating that functions as a series of stepped mirrors. Each "mirror" interacts spatially with light reflected from a sample producing information from multiple points of depth. The parallel depth information is then captured electronically with an array of parallel photoelements. Unlike mechanical OCT systems, grating generated OCT systems can achieve video rates if fast electronic processing is used. However, grating generated OCT systems require a large number of parallel electronic channels, which are very large, complex and power consuming.

[0012] Therefore, there exists a need for an OCT system that utilizes a single electronic channel yet still achieves video rate scanning speeds.

SUMMARY OF THE INVENTION

[0013] The invention relates to an OCDR system that outputs a detector signal which carries image depth information on multiple modulation frequencies, where each modulation frequency preferably corresponds to a different image depth.

[0014] A first, separate aspect of the invention is an OCDR system which outputs a detector signal which carries image depth information on multiple modulation frequencies, where each modulation frequency corresponds to a different image depth.

[0015] A second, separate aspect of the invention is an OCDR system which outputs a detector signal which carries image depth information on multiple modulation frequencies such that the image depth information from the detector signal may be resolved by tuning to the desired modulation frequency.

[0016] A third, separate aspect of the invention is an OCDR system which has a combined signal having a plurality of modulated frequency signals, where each modulated frequency signal corresponds to a different aspect of the image.

[0017] A fourth, separate aspect of the invention is an OCDR system which has a combined signal having a plurality of modulated frequency signals, where each modulated frequency signal corresponds to a different image depth.

[0018] A fifth, separate aspect of the invention is an OCDR system which allows a combined beam to focus onto a single light detector.

[0019] A sixth, separate aspect of the invention is an OCDR system which enables the use of avalanche mode photodiodes.

[0020] A seventh, separate aspect of the invention is an OCDR system which has an interferometer with multiple modulators having no transition from fiber optic to free space.

[0021] An eighth, separate aspect of the invention is a method for imaging body tissue comprising the steps of frequency modulating image depth information onto a signal and outputting the signal

[0022] A ninth, separate aspect of the invention is a method for imaging body tissue comprising the steps of frequency modulating image depth information onto a signal, where each modulation frequency corresponds to a different image depth, and outputting the signal.

[0023] A tenth, separate aspect of the invention is a method for imaging body tissue comprising the step of providing a combined signal having a plurality of modulated frequency signals, where each modulated frequency signal corresponds to a different aspect of the image.

[0024] An eleventh, separate aspect of the invention is a method for imaging body tissue comprising the step of providing a combined signal having a plurality of modulated frequency signals, where each modulated frequency signal corresponds to a different image depth.

[0025] A twelfth, separate aspect of the invention is a method for imaging body tissue which allows a combined beam to focus onto a single light detector.

[0026] A thirteenth, separate aspect of the invention is a method for imaging body tissue which enables the use of avalanche mode photodiodes.

[0027] A fourteenth, separate aspect of the invention is a method for imaging body tissue which uses multiple modulators having no transition between fiber optic and free space.

[0028] The invention may include any one of these separate aspects individually, or any combination of these separate aspects.

[0029] Other systems, methods, features and advantages of the invention will be or will become apparent to one with skill in the art upon examination of the following figures and detailed description. It is intended that all such additional systems, methods, features and advantages be included within this description, be within the scope of the invention, and be protected by the accompanying claims.

BRIEF DESCRIPTION OF THE DRAWINGS

[0030] The components in the figures are not necessarily to scale, emphasis instead being placed upon illustrating the principles of the invention. Moreover, in the figures, like reference numerals designate corresponding parts throughout the different views.

[0031] **FIG. 1** is a block diagram illustrating an example embodiment of an electronics interface.

[0032] **FIG. 2** is a block diagram illustrating another example embodiment of an electronics interface.

[0033] FIG. 3 is a block diagram illustrating an electronics interface coupled between a single-channel multiplexed photo array and an ultrasound console.

[0034] FIG. 4 is a block diagram illustrating an electronics interface coupled between a two-channel multiplexed photo array and an ultrasound console.

[0035] FIG. 5 is a block diagram illustrating an example embodiment of an improved OCDR system.

[0036] FIG. 6 is a block diagram illustrating an electronics interface coupled between the OCDR system of FIG. 5 and an ultrasound console.

[0037] FIG. 7 is a block diagram illustrating an exemplary mixer.

DETAILED DESCRIPTION OF THE PREFERRED EMBODIMENTS

[0038] FIG. 1 shows an example embodiment of an electronics interface between an OCDR system and an ultrasound console. The electronics interface system is coupled between a multi-channel photo array 10 of an OCDR system and an ultrasound console 12. The photo array 10 comprises a plurality of photo detectors (not shown), e.g., 256 photo detectors. Each photo detector of the photo array 10 outputs a signal carrying image brightness information for a certain image depth. The photo detectors may be photodiodes, Charge Coupled Devices (CCDs), or the like. The photo array outputs a plurality of parallel channels, where each one of the parallel channels corresponds to the output of one of its photo detectors.

[0039] The electronics interface includes a plurality of channel processors 15-1 to 15-n, where each channel processor 15-1 to 15-n is coupled to one of the parallel channel outputs of the photo array 10. Each channel processor 15-n includes an analog processor 20-n, an A/D converter 25-n, and a First-In-First-Out (FIFO) memory buffer 30-n. For the sake of simplicity, the reference numeral 15-n refers to any of the channel processors 15-1 to 15-n; 20-n refers to any of analog processors 20-1 to 20-n; 30-n refers to any of FIFO memory buffers 30-1 to 30-n. The electronics interface also includes a data bus 35 coupled to the output of each one of the FIFO memory buffers 30-n of the channel processors 15-n. The electronics interface further includes a single FIFO memory buffer 40 coupled to the data bus 35, a digital-to-analog (D/A) converter 45 coupled to the output of the single FIFO memory buffer 40, and a controller 55. The output of the D/A converter 45 is coupled to the input of the ultrasound console 12. The controller 55 is also coupled to the analog-to-digital (A/D) converters 25-1 to 25-n, the FIFO memory buffers 30-1 to 30-n, the single FIFO memory buffer 40, and the D/A converter 45. The ultrasound console 12 and the controller 55 are each coupled to an ultrasound motor encoder 50 of an ultrasound Patient Interface Unit (PIU). The ultrasound motor 50 outputs encoder pulses which may be spaced, for example, 130 μ s apart, to the ultrasound console 12 and the controller 55. The controller 55 uses the received encoder pulses to synchronize the operation of the electronics interface with the ultrasound console 12.

[0040] During operation, each photo detector of the photo array 10 outputs a signal, such as a current, to one of the parallel channel outputs of the photo array 10. The signal of

each one of the parallel channel outputs may carry image brightness information for a certain image depth, where depth may be taken with respect to a catheter. Each channel processor 15-n of the electronics interface processes one of the parallel channel outputs of the photo array 10. The analog processor 20-n of each channel processor 15-n performs analog processing on the respective parallel channel output signal. The analog processing may include current-to-voltage conversion, signal amplification, bandpass filtering, logarithmic amplification and/or other functions. Logarithmic amplification may be used to translate the signal in the channel processor 15-n to a log scale in order to increase the dynamic range of the signal, for example, to a dynamic range of 100 dB. The output of each analog processor 20-n is coupled to the respective A/D converter 25-n.

[0041] When the controller 55 receives a first encoder pulse from the motor encoder 50, the controller 55 instructs each A/D converter 25-n to digitize the analog output from the respective analog processor 20-n for a predetermined data acquisition time, such as 100 μ s. During this time, each A/D converter 25-n writes its digital data into the respective FIFO memory buffer 30-n. At the end of the data acquisition time, the controller 55 instructs the FIFO memory buffers 30-1 to 30-n of the channel processors 15-1 to 15-n to sequentially write their digital data into the single FIFO memory buffer 40 via the data bus 35. Preferably, the controller 55 instructs the FIFO memory buffers 30-1 to 30-n of the channel processors 15-1 to 15-n corresponding to lower image depths to write their digital data into the single FIFO memory buffer 40 before the FIFO memory buffers 30-1 to 30-n of the channel processors 15-1 to 15-n corresponding to deeper image depths. That way, the digital data from the channel processors 15-1 to 15-n are written into the single FIFO memory buffer 40 in order of increasing image depth.

[0042] When the controller 55 receives a second (subsequent) encoder pulse from the motor encoder 50, the controller 55 instructs the single FIFO memory buffer 40 to output its stored digital data, which was taken during the previous encoder pulse, to the D/A converter 45. Preferably, the single FIFO memory buffer 40 outputs the digital data to the D/A converter 45 in the form of a serial digital data sequence in which the digital data in the digital data sequence are arranged in order of increasing image depth. The D/A converter 45 converts the digital data sequence from the single FIFO memory buffer 40 into a serial analog signal, which is outputted to the input of the ultrasound console 12.

[0043] Also, during the second encoder pulse, the controller 55 instructs each one of the A/D converters 25-n of the channel processors 15-n to acquire a new set of digital data, which is outputted to the ultrasound console 12 in serial analog form during a third encoder pulse. For each encoder pulse, the electronics interface outputs a serial analog signal to the ultrasound console 12 containing image data that was acquired during a previous encoder pulse. Thus, the electronics interface lags behind the ultrasound console by one encoder pulse, typically 130 μ s.

[0044] The serial analog signal outputted by the D/A converter 45 may carry both image brightness information and image depth information. The image brightness information is provided by the amplitude of the serial analog signal. The image depth information is provided by the time position within the serial analog signal. An earlier time

position in the serial analog signal corresponds to a lower image depth than a later time position in the serial analog signal. This is because the serial analog signal was converted from a serial digital data sequence whose digital data were arranged in order of increasing image depth.

[0045] The D/A conversion rate of the D/A converter 40 may be adjusted so that the serial analog signal has approximately the same time length as a typical echo signal, such as 8 μ s. In addition, a mixer (not shown), which may be a Double Side-Band Suppressed Carrier (DSBSC) mixer or other kinds of mixers, may be used to adjust, i.e., translate, the frequency of the serial analog signal before it is inputted to the ultrasound console 12. For example, the mixer may be used to adjust the frequency of the serial analog signal to the frequency of a typical echo signal that the ultrasound console 12 is configured to receive.

[0046] Therefore, the electronics interface processes the parallel channel outputs of the photo array 10 of the OCD system into a serial analog signal. The serial analog signal is similar to the echo signal of an ultrasound transducer in that it carries image brightness information and image depth information in a similar manner to a typical echo signal. This enables the ultrasound console 12 to process the serial analog signal outputted by the electronics interface into an image. In addition, the time length and/or frequency of the serial analog signal may be adjusted to better match the time length and/or frequency of a typical echo signal that the ultrasound console 12 is configured to receive.

[0047] FIG. 2 shows an electronics interface according to another example embodiment of the electronics interface. The electronics interface according to this embodiment is coupled between the multi-channel photo array 10 of an OCD system and an ultrasound console 212 having a digital input. The digital input enables the ultrasound console 212 to receive echo signals that have been digitized by a D/A converter. The digital input of the ultrasound console 212 may use, for example, a Versa Module Eurocard (VME) bus and/or a Peripheral Component Interconnect (PCI) bus to receive digital data.

[0048] The electronics interface according to this embodiment includes a control logic 210 in place of the D/A converter 45. The D/A converter 45 is not needed because the ultrasound console 212 has a digital input. The control logic 210 is coupled to the output of the single FIFO memory buffer 40 and the digital input of the ultrasound console 212.

[0049] The control logic 210 controls the transfer of the digital data sequence from the single FIFO memory buffer 40 to the digital input of the ultrasound console 212. Preferably, the control logic 210 transfers one digital data sequence for each encoder pulse of the motor encoder 50. For the case in which the digital input of the ultrasound console 212 uses a PCI bus, the control logic 210 may also perform handshaking functions to coordinate the transfer of data from the single FIFO memory buffer 40 to the digital input of the ultrasound console 212.

[0050] The ultrasound console 212 according to this embodiment may also include a software-based module for interpreting the digital data sequence received from the electronics interface. For example, the digital data in the digital data sequence may be arranged in order of increasing

image depth so that the different data positions in the digital data sequence correspond to different image depths. In this example, the software-based module may translate the different data positions in a received digital data sequence into their corresponding image depths. The digital data and their corresponding image depths may then be inputted to the ultrasound image processor of the ultrasound console 212 to produce an image.

[0051] FIG. 3 illustrates yet another embodiment of the electronics interface. The electronics interface according to this embodiment is coupled between a multiplexed photo array 308 of an OCD system and an ultrasound console 12. The multiplexed photo array 308 comprises a plurality of photo detectors (not shown), such as 256 photo detectors. Each photo detector of the multiplexed photo array 308 outputs a signal carrying image brightness information for a certain depth. The multiplexed photo array 308 according to this embodiment further includes a channel output 310 and a control input 312. The output of each one of the photo detectors of the multiplexed photo array 308 can be dynamically coupled to the channel output 310 via a switching network (not shown). The switching network is configured to couple the output of one of the photo detectors to the channel output 310 at a certain time and is controlled by the control input 312.

[0052] The electronics interface includes a channel processor 315 coupled to the channel output 310 of the multiplexed photo array 308. The channel processor 315 includes an analog processor 320 and an A/D converter 325. The electronics interface also includes a memory buffer 330, such as a RAM memory or any other kind of memory, coupled to the output of the A/D converter 325. The output of the memory buffer 330 is coupled to the data bus 35 of the electronics interface. The controller 55 of the electronics interface is coupled to the control input 312 of the multiplexed photo array 308. The controller 55 is also coupled to the A/D converter 325, memory buffer 330, D/A converter 45, the single FIFO memory 40, and the motor encoder 50.

[0053] When the controller 55 receives a first encoder pulse from the motor encoder 50, the controller 55 instructs the switching network of the multiplexed photo array 308 to sequentially output the signal of each one of the photo detectors to the channel output 310. The analog processor 320 of the channel processor 315 performs analog processing on the channel output 310 in a manner similar to the analog processor 20-n in FIG. 1. The controller 55 instructs the A/D converter 325 to digitize the output of the analog processor 320. Preferably, the controller 55 coordinates the timing of the switching network of the multiplexed photo array 308 and the A/D converter 325 such that the A/D converter 325 acquires at least one digital datum from the signal of each one of the photo detectors. The A/D converter 325 writes its digital data into the memory buffer 330. Preferably, the memory buffer 330 stores the received digital data into assigned memory addresses according to their corresponding image depths. The memory buffer 330 may determine the corresponding image depths of the digital data, for example, by having the switching network of the multiplexed photo array 308 sequentially output the signals of the photo detectors in order of increasing image depth, or alternatively, in order of decreasing depth. In other words, the switching network outputs the signals of the photo

detectors corresponding to shallower image depths before it outputs the signals of the photo detectors corresponding to deeper image depths.

[0054] The memory buffer 330 sequentially writes its digital data to the single FIFO memory buffer 40 via the data bus 35, preferably, in order of increasing image depth. When the controller 55 receives a second (subsequent) encoder pulse from the motor encoder 50, the digital data acquired during the first encoder pulse is outputted to the ultrasound console 12 in serial analog form in a manner similar to the electronics interface of FIG. 1. Alternatively, for the case in which the ultrasound console 12 has a digital input, the digital data may be outputted without the use of a D/A converter 45 to the ultrasound console 12 as a digital data sequence in a manner similar to the electronics interface of FIG. 2.

[0055] An advantage of the electronics interface according to this embodiment is that it reduces hardware costs by only requiring one channel processor 325 to process the output of the multiplexed photo array 308. This is because the multiplexed photo array 308 sequentially outputs the signals of its photo detectors on a single channel 310 instead of outputting the signals of its photo detectors on separate parallel channels.

[0056] FIG. 4 illustrates still another embodiment of the electronics interface which is coupled to a two-channel multiplexed photo array 408. The multiplexed array 408 includes a first channel output 410-1, and a second channel output 410-2. Each one of a first half of the photo detectors of the multiplexed photo array 408 can be dynamically coupled to the first channel output 410-1 via a first switching network (not shown). Each one of a second half of the photo detectors can be dynamically coupled to the second channel 410-2 via a second switching network (not shown).

[0057] The electronics interface of FIG. 4 includes a first and second channel processor 315-1, 315-2 coupled to the first and second channel output 410-1, 410-2, respectively. Each channel processor 315-1, 315-2 includes an analog processor 320-1, 320-2, an A/D converter 325-1, 325-2 and a memory buffer 330-1, 330-2. The output of the memory buffer 330-1, 330-2 of each one of the channel processors 315-1, 315-2 is coupled to the data bus 35. Alternatively, the first and second channel processors 315-1, 315-2 may share a common memory buffer. In this alternative case, the output of the A/D converter 325-1, 325-2 of each one of the channel processors 315-1, 315-2 would be coupled to the common memory buffer.

[0058] When the controller 55 receives a first encoder pulse from the motor encoder 50, the controller 55 instructs the first switching network of the multiplexed photo array 408 to sequentially output the signal of each one of the first half of the photo detectors to the first channel output 410-1. Similarly, the controller 55 instructs the second switching network to sequentially output the signal of each one of the second half of the photo detectors to the second channel output 410-2. The analog processor 320-1, 320-2 of each one of the channel processors 315-1, 315-2 performs analog processing on the respective channel output 410-1, 410-2. The controller 55 instructs each A/D converter 325-1, 325-2 to digitize the output of the respective analog processor 320-1, 320-2. Preferably, the controller 55 coordinates the timing of the first switching network and the A/D converter

325-1 of the first channel processor 315-1 such that the A/D converter 325-1 acquires at least one digital datum from the signal of each one of the first half of the photo detectors. Similarly, the controller 55 coordinates the timing of the second switching network and the A/D converter 325-2 of the second channel processor 315-2 such that the A/D converter 325-2 acquires at least one digital datum from the signals of each one of the second half of the photo detectors. Each one of the A/D converters 325-1, 325-2 writes its digital data into the respective memory buffer 330-1, 330-2. Preferably, each memory buffer 330-1, 330-2 stores its received digital data into assigned memory spaces according to their corresponding image depth.

[0059] The memory buffers 330-1, 330-2 sequentially write their digital data to the single FIFO memory buffer 40 via the data bus 35, preferably, in order of increasing image depth. When the controller 55 receives a second (subsequent) encoder pulse from the motor encoder 50, the digital data acquired during the first encoder pulse is outputted to the ultrasound console 12 in serial analog form in a manner similar to the electronics interface of FIG. 1. Alternatively, for the case in which the ultrasound console has a digital input, the digital data may be outputted to the ultrasound console 12 as a digital data sequence in a manner similar to the electronics interface of FIG. 2.

[0060] An advantage of the electronics interface according to this embodiment is that it may achieve higher data read out rates compared with the electronics interface of FIG. 3. This is because the electronics interface according to this embodiment uses two channel processors to simultaneously process two separate outputs of the multiplexed photo array. Thus, while adding to the hardware cost of the electronics interface, additional channel processors may increase the data read out rate of the electronics interface. Those skilled in the art will appreciate that the multiplexed photo array and the electronics interface may include any number of channels.

[0061] FIG. 5 shows an example embodiment of an improved OCDR system. The OCDR system includes a light source 510, such as a laser, a beam splitter 520, a beam combiner 570, and plurality of light modulators 540-1 to 540-n. The OCDR system also includes an optical fiber bundle comprising a plurality of optical fibers 545-n, such as 256 optical fibers, where each one of the optical fibers 545-n is coupled to one of the light modulators 540-n. For the sake of simplicity, the reference numeral 540-n refers to any of the light modulators and 545-n refers to any of the optical fibers. Preferably, the light traveling through the optical fibers preceding the light modulators 540-n do not travel in free space or air prior to their being received by the light modulators 540-n. In contrast to grating-based OCDR systems which rely on free space to carry information spatially in the light beam, this embodiment preferably carries information in different frequencies of light, thereby making free space unnecessary. The elimination of a transition from optical fiber to free space improves system efficiency because there is a certain amount of loss at each interface between fiber optic and air. Also, the light traveling from the modulators 540-n to the optical fibers 545-n preferably do not travel in free space. The optical fibers 545-n have different path lengths, which may differ from each other by multiples of, for example, 20 μm . The OCDR system also comprises a first optical coupler 535 positioned between the

beam splitter **520** and light modulators **540-n**, and a second optical coupler **560** positioned between the beam splitter **520** and a sample **565** to be imaged, e.g., a region of a body. The OCDR system further comprises a third optical coupler **575** positioned between a reference beam optical path **555** and the beam combiner **570**, and a light detector **585**. The light detector **585** may be, for example, an avalanche photodiode.

[0062] During operation, the light source **510** generates a source beam **515**. The beam splitter **520** splits the source beam **515** into a reference beam **525** and a sample beam **530**. The split may be a 50-50 split or a 10-90 split, in which 90% of the power of the source beam **515** goes into the sample beam **530** and 10% of the power goes into the reference beam **525**. The amount of power given to each one of the sample and reference **530**, **525** will depend on the application. The second optical coupler **560** directs the sample beam **530** to the sample **565** to be imaged. For example, the sample **565** may be a blood vessel and the second coupler **560** may couple the sample beam **530** to a proximal end of an optical-fiber catheter (not shown), which then transmits the sample beam **530** to the blood vessel. A portion of the sample beam **530** is reflected back from the sample **565** to the second coupler **560**. The second coupler **560** directs the reflected sample beam **530** back to the beam splitter **520**. The reflected sample beam **530** is then reflected off of the beam splitter **520** to the beam combiner **570**.

[0063] The first coupler **535** splits the reference beam **525** into a plurality of beams and couples each one of the plurality of beams to one of the light modulators **540-1** to **540-n**. Each light modulator **540-n** modulates the respective beam at a unique modulation frequency, for example, within an operating frequency range of 10 MHz. In this example embodiment, each light modulator **540-n** may be a Lithium Niobate crystal, where a sinusoidal voltage is applied to the crystal to modulate the frequency of the respective beam. Each light modulator **540-n** outputs its frequency-modulated beam to one of the optical fibers **545-n**. The frequency-modulated beams propagate through their respective optical fiber **545-n** and are merged into a frequency-modulated reference beam **555**, for example, using a lens (not shown).

[0064] The resulting frequency-modulated reference beam **555** contains each modulation frequency of the light modulators **540-n**. In addition, each modulation frequency of the frequency-modulated reference beam **555** experiences a slightly different propagation delay path length. This is because the frequency-modulated beams from the light modulators **540-n** propagate through optical fibers **545-n** of different path lengths before merging into the frequency-modulated reference beam **555**. The frequency-modulated reference beam **555** is directed along a reference beam optical path **555** to the third optical coupler **575**. The third optical coupler **575** directs the frequency-modulated reference beam **555** to the beam combiner **570**, which combines the frequency modulated-reference **555** and the reflected sample beam **530** into a combined beam **580**. The resulting combined beam **580** is focused onto the light detector **585**, for example, by an optical lens (not shown).

[0065] The combined beam **580** preferably carries both image brightness information and image depth information of the sample **565**. The image brightness information is provided by the light intensity of combined beam **580**. The image depth information is provided by the different modulation

frequencies of the combined beam **580**. This is because the different modulation frequencies of the frequency-modulated reference beam **555** experience different propagation delay path lengths due to the different path lengths of the optical fibers **545-n**. As a result, the different modulation frequencies of the frequency-modulated reference beam **555** constructively interfere with portions of the reflected sample beam **530** reflected from different depths in the sample **565**. Thus, the depth information contained in the combined beam **580** is a function of modulation frequency.

[0066] The light detector **585** detects the combined beam **580** and outputs a detector signal (not shown) that carries image brightness information at multiple image depths, where each image depth corresponds to a different modulation frequency. Therefore, the image depth information contained in the detector signal may be resolved by tuning a conventional radio tuner to the desired modulation frequency. This can be achieved using a sweep oscillator and standard detection hardware such as heterodyne detection.

[0067] An advantage of the OCDR system according to this embodiment is that it focuses the light energy of the combined beam **580** onto one light detector, instead of wasting a portion of the light energy due to spacing between photo detectors in a photo array. In addition, the OCDR system enables the use of avalanche mode photodiodes, which are far more sensitive than standard photo-elements.

[0068] By using multiple frequency modulators, only one electronic channel needs to be constructed for processing the imaging information. The use of a single electronic channel dramatically reduces the size and complexity of the electronics interface. In addition, the present invention contemplates high-speed electronic processing such that video rates, and therefore, real time viewing, can be realized.

[0069] FIG. 6 shows an electronics interface coupled between the light detector **585** of the OCDR system of FIG. 5 and an ultrasound console **12**. The electronics interface includes a broadband amplifier **610** coupled to the light detector **585** of the OCDR system, a broadband filter **620** coupled to the output of the broadband amplifier **610**, and a tuner **640** coupled to the output of the broadband filter **620**. For example, the tuner **640** may be a radio tuner and the broadband filter **620** may be a bandpass filter. The tuner **640** is coupled also to lines **645** and **647**. The electronics interface further includes an A/D converter **650** coupled to the tuner **640** via line **645**, and a FIFO memory **660** coupled to the output of the A/D converter **650**. The output of the FIFO memory **660** is coupled to a data bus **35**. A single FIFO memory **660** is coupled to the data bus **35**. The single FIFO memory **40** is coupled to a D/A converter **45**. The D/A converter **45** is coupled to the ultrasound console **12**. The ultrasound console **12** is coupled to a motor encoder **50**. The controller **55** in this example embodiment is coupled to the motor encoder **50**, tuner **640**, A/D converter **650**, FIFO memory **660**, D/A converter **45** and the single FIFO memory **40**. The controller **55** controls the tuned frequency of the tuner **640** via line **647**.

[0070] During operation, the broadband amplifier **610** amplifies the detector signal from the light detector **585** and passes the amplified signal to the broadband filter **620**. The broadband filter **620** passes the portion of the amplified detector signal that is within the modulation frequency range

of the OCDR system to the tuner **640**. The tuner **640** is configured to tune to one of the modulation frequencies of the detector signal at a time.

[0071] When the controller **55** receives a first encoder pulse from the motor encoder **50**, the controller **55** instructs the tuner **640** to sweep its tuning frequency across the entire modulation frequency range of the OCDR system. This causes the tuner **640** to sequentially tune to each one of the modulation frequencies of the detector signal. The A/D converter **650** digitizes the tuned detector signal from the tuner **640**. Preferably, the controller **55** coordinates the timing of the tuner **640** and the A/D converter **650** such that the A/D converter **650** acquires at least one digital datum for each tuned modulation frequency of the detector signal. The A/D converter **650** writes its digital data to the FIFO memory **660**. Preferably, the A/D converter **650** writes its digital data to the FIFO memory **660** in order of increasing image depth. This may be done, for example, by having the lower modulation frequencies of the detector signal correspond to shallower image depths and the higher modulation frequencies of the detector signal correspond to the deeper image depths, and by sweeping the tuning frequency of the tuner **640** from a low frequency to a high frequency.

[0072] The FIFO memory **660** sequentially writes its digital data to the single FIFO memory buffer **40** via the data bus **35**, preferably, in order of increasing image depth. When the controller **55** receives a second (subsequent) encoder pulse from the motor encoder **50**, the digital data acquired during the first encoder pulse is outputted to the ultrasound console **12** in serial analog form in a manner similar to the electronics interface of **FIG. 1**. Alternatively, for the case in which the ultrasound console **12** has a digital input, the digital data may be outputted to the ultrasound console **12** as a digital data sequence in a manner similar to the electronics interface of **FIG. 2**.

[0073] **FIG. 7** illustrates a block diagram of an exemplary tuner **640** that may be used in the electronics interface of **FIG. 6**. The tuner **640** includes a sweep control **730** coupled to the controller **55** by line **647**, and a local oscillator **740** coupled to the output of the sweep control **730**. The sweep control **730** varies the frequency of the local oscillator **740** based on the control signals from the controller **55** over line **647**. The tuner **640** further includes a mixer **710** coupled to the output of the oscillator **740** and to the output **625** of the broadband filter **620**.

[0074] The output of the mixer **710** is coupled to a rectifier **712**, which is coupled to a low pass filter **720**. The mixer **710** down converts, i.e., translates, the modulation frequencies of the detector signal to lower frequencies by mixing, i.e., multiplying, the output **625** of the broadband filter **620** with the output of the oscillator **740**. The low pass filter **720** is configured to pass one of the down-converted modulation frequencies of the detector signal at a time. The frequency of the local oscillator **740** controls which one of the down-converted modulation frequencies of the detector signal is passed through the low pass filter **720**. Thus, the tuner **640** can be tuned to different modulation frequencies of the detector signal by varying the frequency of the oscillator **740** using the sweep control **730**. The low pass filter **720** is coupled to a logarithmic amplifier **722**.

[0075] Also, it is contemplated that any embodiment described in this patent specification can be modified such

that, for example, the number of any of the components can be increased or decreased as desired. Further, it is contemplated that features shown in one embodiment may be added to those of another embodiment, or features shown in one embodiment may be deleted, as desired. Any one or more of the features depicted in any of the figures, or described in the accompanying text, may be interchanged with that of another figure to form still other embodiments.

[0076] While various embodiments of the application have been described, it will be apparent to those of ordinary skill in the art that many embodiments and implementations are possible that are within the scope of the invention. Therefore, the invention is not to be restricted or limited except in accordance with the following claims and their equivalents.

What is claimed is:

1. An optical system for imaging body tissue comprising:

a beam generator which receives a light beam and generates a reference light beam and a sample light beam, the sample light beam being directed at body tissue to be imaged;

a first coupler which receives the reference light beam and generates a plurality of reference light beams;

a plurality of first optical fibers, each first optical fiber being coupled to receive one of the plurality of reference light beams;

a plurality of modulators, each modulator being coupled to one of the plurality of first optical fibers for receiving one of the plurality of reference light beams from one of the plurality of first optical fibers, each modulator modulating the reference light beam at a unique modulation frequency;

a plurality of second optical fibers, each one of the second optical fibers being coupled to one of the plurality of modulators for receiving the modulated reference light beam from the modulator;

a second coupler which receives a reflected sample light beam which is reflected off the body tissue; and

a beam combiner coupled to the second coupler, the beam combiner receiving the reflected sample light beam from the second coupler, receiving the modulated reference light beam and generating a combined light beam.

2. The optical system for imaging body tissue of claim 1 wherein the beam generator is a beam splitter.

3. The optical system for imaging body tissue of claim 2 wherein the beam splitter splits the received light beam into the reference light beam and the sample light beam where the reference light beam and the sample light beam have a desired proportion of power.

4. The optical system for imaging body tissue of claim 1 wherein each one of the plurality of modulators includes a Lithium Niobate crystal, where a sinusoidal voltage is applied to the crystal to modulate the frequency of the reference light beam.

5. The optical system for imaging body tissue of claim 1 further comprising a beam merger coupled to the plurality of second optical fibers, the beam merger receiving the modulated reference light beams from the plurality of second

optical fibers and merging the modulated reference light beams into a merged reference light beam.

6. The optical system for imaging body tissue of claim 5 wherein the beam merger is a lens.

7. The optical system for imaging body tissue of claim 5 further comprising a third coupler coupled to receive the merged reference light beam from the beam merger.

8. The optical system for imaging body tissue of claim 7 wherein the third coupler directs the merged reference light beam to the beam combiner.

9. The optical system for imaging body tissue of claim 1 further comprising a light detector which is coupled to the beam combiner to receive the combined light beam.

10. The optical system for imaging body tissue of claim 9 further comprising a lens coupled to the beam combiner, the lens receiving the combined light beam and directing the combined light beam to the light detector.

11. The optical system for imaging body tissue of claim 1 wherein the combined light beam carries a first information and a second information about the body tissue.

12. The optical system for imaging body tissue of claim 11 wherein the first information is image depth information of the body tissue.

13. The optical system for imaging body tissue of claim 12 wherein the second information is image brightness information of the body tissue.

14. The optical system for imaging body tissue of claim 12 wherein the image depth information is provided by the different modulation frequencies of the modulated reference light beams.

15. The optical system for imaging body tissue of claim 13 wherein the image brightness information is provided by the light intensity of the combined light beam.

16. The optical system for imaging body tissue of claim 1 wherein the combined light beam is the result of interference between the reflected sample light beam and the modulated reference light beam.

17. The optical system for imaging body tissue of claim 16 wherein each one of the plurality of second optical fibers has a different path length.

18. The optical system for imaging body tissue of claim 1 wherein the plurality of modulators are coupled to the plurality of first optical fibers such that the plurality of reference light beams do not travel in free space.

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