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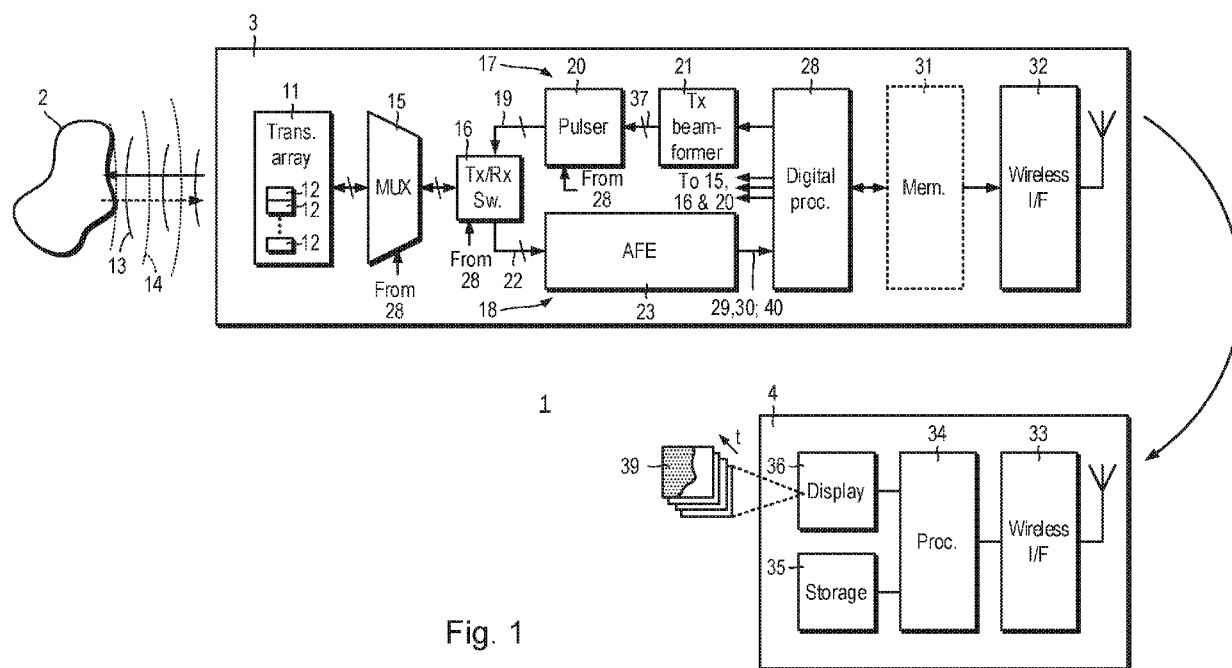


Fig. 1

(57) **Abstract:** An ultrasonic imaging device is disclosed. The device comprises an ultrasonic transducer array (11) comprising a plurality of transducer elements (12), a multiplexer (15) comprising a plurality of inputs and at least one output, each input coupled to a respective transducer element, an analogue front-end (23) comprising at least one channel, each channel coupled to a respective output of the multiplexer and each channel including a respective analogue demodulator (44, 45) arranged to extract in-phase and quadrature components of a signal from the transducer and to provide demodulated in-phase and quadrature signal components (50, 51), at least two analogue-to-digital converters (60, 61) configured to receive signals comprising or obtained from the in-phase and quadrature signal components and to provide digitised demodulated in-phase and quadrature signal components and a wireless interface (32) configured to transmit signals comprising or obtained from the digitised demodulated in-phase and quadrature signal components.

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Ultrasonic imaging device

Field of the Invention

The present invention relates to an ultrasonic imaging device and to an ultrasonic
5 imaging system which includes an ultrasonic imaging device and a processing system.

Background

Point of care ultrasonography has been the focus of extensive research over the past few
decades and various single-chip, miniaturized and wearable ultrasonic imaging systems
10 have been proposed.

For example, US 2014/288428 A1 describes single-chip ultrasonic imaging in which
on-chip signal processing is used to reduce data bandwidth. US 2012/101386 A1
describes wrapping an array of acoustic transducers wrapped around the circumference
15 of a capsule for endoscopy and sending generated echo image signals to receiver devices
attached or worn on the body.

WO 2015/138643 A1 discloses a wearable ultrasound system comprising an ultrasound
probe, a proximal wearable component electrically interconnected with the ultrasound
20 probe adapted to be wearable on the hand, wrist, or arm of a user, and including at
least one user interface mechanism. US 2012/0065479 A1 describes a wearable patch
which comprises an ultrasound sensor array.

As ultrasonic imaging systems become smaller and adapted for remote use, for instance
25 in the form of a capsule to be swallowed or a patch to be worn by a patient, they will be
faced with additional challenges such as limits on power consumption and transmission
data rates.

Summary

According to a first aspect of the present invention there is provided an ultrasonic imaging device comprising an ultrasonic transducer array comprising a plurality of transducer elements, a multiplexer comprising a plurality of inputs and at least one
5 output, each input coupled to a respective transducer element, an analogue front-end comprising at least one channel, each channel coupled to a respective output of the multiplexer and each channel including a respective analogue demodulator arranged to extract in-phase and quadrature components of a signal from the transducer and to provide demodulated in-phase and quadrature signal components, at least two
10 analogue-to-digital converters configured to receive signals comprising or obtained from the in-phase and quadrature signal components and to provide digitised demodulated in-phase and quadrature signal components and a wireless interface configured to transmit signals comprising or obtained from the digitised demodulated in-phase and quadrature signal components.

15 This can be helpful particularly in applications employing frame rates less than 30 Hz, for example, in a range of 2 to 4 Hz, and in wireless applications with limited data bandwidths.

20 According to a second aspect of the present invention there is provided an ultrasonic imaging device comprising an ultrasonic transducer array comprising a plurality of transducer elements, a multiplexer comprising a plurality of inputs and at least one output, each input coupled to a respective transducer element, an analogue front-end comprising at least one channel, each channel coupled to a respective output of the
25 multiplexer and each channel arranged to provide an RF signal in dependence on the signal from the transducer, at least one analogue-to-digital converter configured to receive a signal comprising or obtained from the RF signal and to provide a digitised RF signal, a digital processor arranged to extract in-phase and quadrature components from the digitised RF signal and to provide digitised demodulated in-phase and
30 quadrature signal components and a wireless interface configured to transmit signals comprising or obtained from the digitised demodulated in-phase and quadrature signal components. Each channel may comprise a respective amplifier.

This can be helpful particularly in applications employing frame rates less than 30 Hz,
35 for example, in a range of 2 to 4 Hz, and in wireless applications with limited data bandwidths.

The analogue front-end may include the at least two analogue-to-digital converters.

5 Each channel of the analogue front-end may include first and second low-pass filters arranged to filter the in-phase and quadrature signal components respectively before the in-phase and quadrature signal components are digitised by the analogue-to-digital converters.

10 The analogue-to-digital converters may sample at a uniform rate. Thus, the low-pass filters may provide, to the wireless interface, filtered digitised demodulated in-phase and quadrature signal components (herein also referred to as “quasi digitised demodulated in-phase and quadrature signals”) which are based on the digitised demodulated in-phase and quadrature signal components using low-pass filters.

15 The analogue-to-digital converters may sample at a non-uniform rate. Thus, the analogue-to-digital converters may provide filtered digitised demodulated in-phase and quadrature signal components to the wireless interface without the need for low-pass filters.

20 The analogue-to-digital converter may directly sample RF data or amplified RF data at a uniform rate. Thus, the analogue-to-digital converter(s) may provide digitised RF data to a digital demodulator.

25 The first and second low-pass filters may have respective bandwidths which are less than the Nyquist cut-off frequency. The analogue-to-digital converters may be configured to supply the digitised demodulated in-phase and quadrature signal components to the wireless interface and wherein wireless interface is configured to transmit the digitised demodulated in-phase and quadrature signal components.

30 The ultrasonic imaging device may further comprise a processor arranged, for each channel, to receive the digitised demodulated in-phase and quadrature signal components and to perform beamforming based on the digitised demodulated in-phase and quadrature signal components.

The processor may be configured to receive RF signals and perform digital demodulation, supplying the image processor with digitised in-phase and quadrature signal components.

5 The processor may be configured to apply a time delay to digitised RF or demodulated in-phase and quadrature signal components. The processor may be arranged to carry out interpolation and to provide interpolated in-phase and quadrature signal values. The time delay may be applied during interpolation, i.e. as part of interpolation. The time delay may be applied after interpolation. The processor may be arranged to carry
10 out phase rotation on the interpolated in-phase and quadrature signal values to produce first and second RF signal values. The processor may be arranged to carry out summation of the first and second RF signals to a given memory location corresponding to a given pixel. The processor may be arranged to repeat interpolation, phase rotation and summation for a series of digitised demodulated in-phase and
15 quadrature signal components.

The ultrasonic imaging device may further comprise memory (such as Flash memory or SRAM) for storing a partial or complete two-dimensional image or a three-dimensional image comprising scan-line samples or pixel values received from the processor, where
20 a scan-line comprises a set of beamformed imaging points focused in a particular angular direction.

The processor may be configured to compute time delays using either Cartesian coordinates or polar coordinates for rectangular or sector images respectively.

25

The processor may be an ASIC, FPGA or other type of monolithic integrated circuit. The monolithic integrated circuit may include the memory (i.e. be on-chip memory). The memory may comprise a separate integrated circuit (i.e. be off-chip memory).

30 The wireless interface may be configured to transmit scan-line samples or a two-dimensional image or three-dimensional image.

The ultrasonic imaging device may further comprise an excitation circuit comprising a pulser, a transmit beamformer and a processor configured to control the beamformer.

35 The processor of the excitation circuit may be configured to perform synthetic transmit

aperture beamforming, focused phased array beamforming or focused synthetic phased array beamforming.

5 The same processor may be used for synthetic transmit aperture beamforming and synthetic receive aperture beamforming.

The processor(s) may comprise an ASIC or an FPGA. A monolithic integrated circuit may provide the analogue front end may and the processor(s). In other words, the imaging device may be implemented in a single chip.

10

The analogue front-end may comprise one channel.

15 The ultrasonic imaging device may comprise a housing which contains the ultrasonic transducer array, the multiplexer, the analogue front-end, the at least two analogue-to-digital converters and the wireless interface and wherein the housing has a volume less than 5 cm³ and preferably less than 2 cm³.

20 The ultrasonic imaging device may be adapted to be a capsule for swallowing by a human subject or a non-human animal subject or for passing through a passage of a non-animal subject.

The ultrasonic imaging device may be adapted to be a patch for applying to a surface of a subject.

25 The ultrasonic imaging device may be adapted to be a hand-held wand for scanning over a surface of a subject.

30 According to a third aspect of the present invention there is provided a device comprising a network interface and a processor coupled to the network interface (for example, a wired or wireless network interface), wherein the processor is configured to receive parametric or filtered quasi in-phase and quadrature signals, to reconstruct in-phase and quadrature signals and to perform synthetic aperture beamforming.

35 According to a fourth aspect of the present invention there is provided an ultrasonic imaging system comprising an ultrasonic imaging device according to the first aspect or second aspect of the present invention and a processing device comprising a network

interface (for example, a wired or wireless network interface), a processor, storage and a display, wherein the ultrasonic imaging device and the processing device are in wireless communication.

Brief Description of the Drawings

Certain embodiments of the present invention will now be described, by way of example, with reference to the accompanying drawings, in which:

Figure 1 is a schematic block diagram of an ultrasonic imaging system;

5 Figure 2 schematically illustrates synthetic aperture beamforming;

Figure 3 is a schematic block diagram of an imaging device having a first receive signal processing arrangement employing synthetic aperture beamforming on demodulated I/Q signals;

Figure 4 is a finite state machine for a synthetic aperture beamforming process;

10 Figure 5 shows first, second and third simulated two-dimensional mode ultrasound images;

Figure 6 is a schematic block diagram of an imaging device having a second receive signal processing arrangement employing compressive synthetic aperture beamforming on demodulated I/Q signals and a back-end processing device;

15 Figure 7 shows fourth, fifth and sixth two-dimensional mode ultrasound images;

Figure 8 is a schematic view of an ultrasonic imaging device in the form of a capsule;

Figure 9 is a schematic view of an ultrasonic imaging device in the form of a wearable patch;

20 Figure 10 is a schematic view of an ultrasonic imaging device in the form of a hand-held wand;

Figure 11 schematically illustrates synthetic phased subarray beamforming;

Figure 12 shows first, second and third measured B-mode ultrasound images; and

25 Figure 13 is a schematic block diagram of an imaging device having a third receive signal processing arrangement which is a variant of the first receive signal processing arrangement.

Detailed Description of Certain Embodiments

In the following description, like parts are denoted by like reference numerals.

30 Ultrasonic imaging system 1

Referring to Figure 1, a system 1 for ultrasonically imaging a sample 2 in real time is shown. The system 1 includes an ultrasonic imaging device 3 and a processing system 4. The imaging device 3 is capable of capturing ultrasound signals of the sample 2, processing the ultrasound signals in real-time, in particular demodulating RF signals in the analogue domain and optionally carrying out synthetic aperture beamforming to form two-dimensional images, and transmitting processed signals to the processing

35

system 4. The processing system 4 may carry out further processing of the signals, such as signal reconstruction and baseband beamforming, and display images.

5 The ultrasonic imaging device 3 includes an ultrasonic transducer array 11 comprising an array of N transducer elements 12 for generating ultrasound waves 13 and detecting reflected ultrasound waves 14, where N is greater than 1 and may be, for example, N equal to 32, 64 or 128. The imaging device 3 includes a multiplexer/demultiplexer 15 and transmit/receive switches 16 which couple the transducer array 11 to excitation and detection circuitry 17, 18.

10

The excitation circuitry 17 generates excitation pulses 19 for the transducer array 11 and includes a pulser 20 and a transmit beamformer 21. The detection circuitry 18 processes signals 22 received from the transducer array 11 and includes an analogue front-end 23 for demodulating the received signals 22 in the analogue domain into in-
15 phase and quadrature signals (herein also referred to as “demodulated I/Q signals” or “I/Q signals”) and generating digitised I/Q signals 29, 30, or for taking the received signals 22 and generating digitised RF signals 40. As will be explained in more detail hereinafter, demodulating the received signals 22 in the analogue domain can help to reduce digital processing overhead which can lead to reduced power consumption and
20 reduced bandwidth needed for transmission to the processing device 4.

The ultrasonic imaging device 3 includes a digital processor 28. The digital processor 28, among other things, controls transmission beamforming. The digital processor 28 can also perform other functions, such as (e.g. log) compression. The digital processor
25 28 may also carry out quadrature demodulation of the digitised RF signals 40.

Optionally, the digital processor 28 may carry out synthetic aperture beamforming of digitalised I/Q signals received from analogue front-end 22 to produce two-dimensional images. However, as will be also be explained hereinafter, the ultrasonic
30 imaging device 3 need not carry out synthetic aperture beamforming. Instead, the ultrasonic imaging device 3 simply demodulate signals, employ a low-pass filter having a bandwidth which is reduced below the Nyquist cut-off frequency and transmit low-rate samples to the processing system 5 which carries out signal reconstruction and baseband beamforming.

Optionally, the digital processor 28 may carry out coherent or incoherent overlapping of scan-line samples, which are transmitted wirelessly to be processed by the system 4 in order to form an ultrasound image.

5 The ultrasonic imaging device 3 may include memory 31, for example in the form of static random-access memory (SRAM), for storing processed signals 29, 30, for example in the form of scan line samples or 2D image pixel values, before transmission to the processing device 4. The ultrasonic imaging device 3 includes memory 31 when image processing is carried out in the ultrasonic imaging device 3. However, if image
10 processing is carried out in the processing device 4, then the memory 31 may be omitted.

The ultrasonic imaging device 3 includes a wireless interface 32 for transmitting wireless signals 32 to the processing system 4. The wireless interface 32 may take the
15 form of a MICS band transceiver having a suitably high bit rate (for example, greater than 800 kps), an ISM-band transceiver, a WiFi transceiver, Bluetooth (RTM) transceiver, ZigBee (RTM) transceiver or other form of suitable wireless network transceiver.

20 The ultrasonic imaging device 3 includes a battery (not shown) and may include an energy-harvesting device (not shown). As will be explained in more detail later, the ultrasonic imaging device 3 is capable of operating remotely from the processing system 4.

25 Referring still to Figure 1, the processing system 4 includes a wireless interface 33, a digital processor 34, non-volatile memory or storage 35 (for example, in Flash memory or solid-state disk drive) and a display 36 for displaying two-dimensional images 39. The processing system 4 takes the form of a computer system having wireless connectivity and may be a portable or handheld computing device having wireless
30 connectivity, such as a lap-top computer, tablet computer or even smartphone.

The processing system 4 may be configured to convert scan-line samples to pixel values using a form of interpolation. The processing system 4 may be configured to implement coherent or incoherent overlapping of scan-lines, or beamforming using RF or in-phase
35 and quadrature samples from the ultrasound imaging device 3.

Transmission beamforming

Referring still to Figure 1, transmit beamforming employs synthetic aperture beamforming whereby an aperture is synthetically formed by multiplexing a group of transducer elements 12 over the transducer array 11. A single transducer element 12
5 can be excited at each step in the beamforming, although multiple transducer elements 12 may be excited (with or without aperture apodization) to improve signal-to-noise ratio. A phase delay, τ_n , may be applied to each pulse 19 exciting a respective transducer element 12 so as to form a focused beam, plane wave or a parabolic defocusing lens. In the case of a parabolic defocusing lens, the delay, τ_n , for the n^{th}
10 transducer element 12 is set to $(1/v)(x_n^2/2z_d)$ where x_n is the distance to the n^{th} element 12 from the sub-aperture centre, z_d is the distance of a defocal point from the sub-aperture and v is the velocity of sound which, for body tissue, is $1,540 \text{ ms}^{-1}$.

The digital processor 28 controls the digital beamformer 21 which produces delayed
15 excitation pulses 37. These pulses are amplified by the pulser 20 to an appropriate voltage level, for example, between 15 and 50 V or between 15 and 100 V, depending on the type of transducer. For example, piezoelectric micro-machined ultrasound transducers may be excited using CMOS-level voltages, i.e. 3.3 V to 5 V. The excitation signal is a unipolar or bipolar pulse with a duration of half the carrier period, one or
20 multiple periods of the carrier signal. Other transducer types, such as bulk piezo-ceramic, capacitive micro-machined ultrasound transduce and the like, may require higher voltage excitation signals, i.e. $> 5 \text{ V}$.

External high voltage switches (not shown) in the transmit/receive switches 16 may be
25 used protect the analogue front-end 23 during transmission. These switches are controlled using the digital processor 28. During transmission, the switches are open, thereby isolating high voltage pulses from the analogue front-end 23. During reception, the switches are closed thereby allowing reflected signals through to the analogue front-end 23.

30

Synthetic aperture beamforming

As mentioned hereinbefore, the ultrasonic imaging device 3 can perform synthetic aperture beamforming on digitised demodulated I/Q signals. Thus, the ultrasonic imaging device 3 may employ both receive and transmit aperture beamforming with
35 I/Q demodulation. This can be used to reduce hardware complexity and power consumption.

Details regarding synthetic receive aperture beamforming protocol can be found in K. L. Gammelmark and J. A. Jensen: "Multielement synthetic transmit aperture imaging using temporal encoding", IEEE Transactions on Medical Imaging, volume 22, pages
5 552–563 (2003) which is incorporated herein by reference. Details regarding synthetic transmit aperture beamforming can be found in H. Azhari, Basics of Biomedical Ultrasound for Engineers, 1st ed. Wiley-IEEE Press, 2010 which is incorporated herein by reference.

10 In synthetic receive aperture beamforming (SRA), only one or a group of elements 12 are used in receive, resulting in a small active receive aperture. Different elements 12 are multiplexed to synthesise a larger aperture. In regarding synthetic transmit aperture (STA), only a single element 12 is used for transmission. This creates a cylindrical wave 13 front which covers the whole region of interest. The echo 14 is
15 received by all elements 12 and processing is done in parallel to form a low-resolution image. A second transmission yields a second image and so forth. After N_t transducer elements have transmitted, low-resolution images are summed and a high-resolution image is created.

20 The ultrasonic imaging device 3 can combine synthetic receive aperture and synthetic transmit aperture by serialising formation of both transmit and receive apertures so as to maximise the signal-to-noise ratio and reduce hardware complexity.

Referring to Figure 2, an element 12 or a group of elements 12 forming a parabolic
25 defocusing lens is excited to ensonify a region of interest (step SA). Transmission is carried out $n=N/M$ times for all receive elements 12 (step SB) and M receive channels are required to process the reflection signal (where $N > M$ and $M \geq 1$). For example, Figure 2 shows the beamforming process for $M = 1$ receive channel, which is the simplest case requiring the least hardware complexity. The result is a series of n low-
30 resolution images 37₁, 37₂...37_n which are combined to form a higher-resolution image 38 (step SC). Spatial compounding is then used to increase the signal-to-noise ratio. This process is repeated for N_t different transmit positions, such that the final image is an average of the higher resolution images. This approach can help to reduce hardware complexity and cost and means that only one channel is required, if necessary.

Referring also to Figure 11, a group of elements 12 may be excited to form a focused ultrasound beam 70 in multiple angle or directions. Transmission is carried out $n=N/M$ times for each angle and M receive channels are used to process the reflection signal. Each transmission results in a set of K scan-lines focused in the direction of the transmitted focused beam. Each scan-line sample is calculated by applying full dynamic beamforming, *i.e.*, by applying a delay, interpolation, phase rotation and summation, as described in more detail hereinafter.

Figure 11 illustrates a single transmission direction. Each subsequent transmission angle will yield a group of K scan lines which may be coherently or incoherently overlapped with the previous group in the imaging device 3 or the processor 4. The processor 4 carries out conversion of scan-line samples to pixel values (*i.e.*, scan line conversion).

In conventional systems, beamforming is carried out in the RF domain. By demodulating data first, however, the bandwidth and sampling rate may be decreased, leading to a substantial saving in power, as will now be described in more detail.

Overview

A transducer can be considered to produce a bandpass signal $R(t)$ which may be expressed as:

$$R(t) = A(t)\cos(\omega_c t + \varphi) \quad (1)$$

where $A(t)$ is the amplitude envelope, ω_c the carrier frequency in radians per second, φ the phase and t is time. Expansion of $R(t)$ yields:

$$R(t) = A_I(t)\cos(\omega_c t + \varphi) + A_Q(t)\sin(\omega_c t + \varphi) \quad (2)$$

where $A_I(t) = A(t)\cos\omega_c t$ and $A_Q(t) = A(t)\sin\omega_c t$ are the in-phase and quadrature components respectively. These components may be obtained by mixing the signal with a reference signal in the analogue domain and filtering the result. Since $A_I(t)$ and $A_Q(t)$ are baseband signals, they may be sampled at a lower rate. This reduces the computational burden on the beamforming processor. After sampling, the next step is to appropriately phase-rotate the I/Q data for focusing.

For a given pixel location $\vec{r}_p$ at depth index k , the required time delay instance $t_p(i,j)$ to take the signal value for summation is calculated by dividing the distance by the speed of sound, c , in the medium, namely:

5

$$t_p(i,j) = \frac{|\vec{r}_p - \vec{r}_e(i)| + |\vec{r}_p - \vec{r}_r(j)|}{c} \quad (3)$$

where $\vec{r}_p$ is the imaging point, $\vec{r}_e(i)$ is the location of the i^{th} transmitting element and $\vec{r}_r(j)$ is the location of the j^{th} receiving element. A corresponding discretised delay index $I_p(i,j)$ may then be calculated for Cartesian or polar image coordinates. If this
 10 delay is applied directly to the I/Q data, then critical frequency-dependent phase errors distort the final image. Furthermore, uniform samples do not fall on the exact sampling points required for focusing at all pixels. Therefore, an interpolation factor of K is applied. In particular, if N_s sample points are obtained, then there are many as $K \times N_s$ index locations between 1 and
 15 $I_p(i,j)_{\max}$. The index value is read from a lookup table that is calculated *a priori*, based on the locations of each pixel $\vec{r}_p$, and transmitting element, i , or receiving element, j .

For each index location $I_p(i,j)$, the I or Q data are then interpolated on-the-fly using linear, quadratic or other suitable form of interpolation.

20

The next step is to phase rotate by re-modulating or upconverting the IQ sample points back to RF by mixing the interpolated result with new discrete reference signals:

$$I_{ref}[n] = \cos[\omega_c n] \quad (4)$$

$$Q_{ref}[n] = \sin[\omega_c n] \quad (5)$$

25 where ω_c is the carrier frequency in rad/s and n is the discretised time index. Again, $I_{ref}[n]$ and $Q_{ref}[n]$ are calculated *a priori*. The interpolated I and Q values are multiplied by the reference signals at $n = I_p$ and then summed to yield the RF amplitude:

$$R[n] = A_I[n] \cos[\omega_c n] + A_Q[n] \sin[\omega_c n] \quad (6)$$

$$= A[n] \cos \phi \cdot \cos[\omega_c n] + A[n] \sin \phi \cdot \sin[\omega_c n] \quad (7)$$

$$= A[n] \cos[\omega_c n + \phi] \quad (8)$$

This value is then added to the scan-line or pixel location, and the process is repeated for all i, j and n values, resulting in a low-resolution image. These low-resolution scan-lines or images are summed or averaged to obtain a higher resolution image, which
 5 may then be transmitted via a wireless transmission link to an external post-processor, i.e. processor 34 (Figure 1). The final focussed signal $y_f(\vec{r}_p)$ is:

$$y_f(\vec{r}_p) = \sum_{j=1}^N \sum_{i=1}^{N_t} a(I_p(i, j)) R(I_p(i, j)) \quad (9)$$

where $a(I_p(i, j))$ is the apodization (weighting) function, $R(I_p(i, j))$ is the phase-shifter
 10 I/Q sum at $I_p(i, j)$, N is the number of transducer elements and N_t is the number of transmissions.

The process lends itself to an iterative, pipelined approach that may easily be implemented in a hardware description language (HDL). Calculations for parallel
 15 groups of pixels may be pipelined during the reflection period, and the only memory required is for the image frame (which is updated dynamically), a single delay matrix, an array of sine/cosine values and dynamic apodization constants. In the case of single-element synthetic aperture beamforming, the frame rate is a function of the number of transmissions, i_{max} , the lateral pixel resolution/size of the receive aperture,
 20 j_{max} , the axial pixel resolution, k_{max} , the number of pixels calculated in parallel per pipeline interval, N_p , and the system clock frequency, f_{clk} :

$$FR = \frac{f_{clk}}{2 \cdot i_{max} \cdot j_{max} \cdot k_{max} / N_p} \quad (10)$$

A factor of two in the denominator is introduced to account for pipelining send and
 25 receive operations in hardware over two clock cycles.

The frame rate/image quality can be increased at the expense of frequency/area/power consumption. For instance, more transmit positions i_{max} implies better spatial
 30 compounding and signal-to-noise ratio and, therefore, better image quality. Likewise, a larger k_{max} implies a better axial pixel resolution. However, increasing i_{max} or k_{max} leads to a lower frame rate if the clock frequency f_{clk} remains constant because

transmit/receive operations are multiplexed, for example, through one or a few channels.

Dynamic apodization is also applied to keep the F-number ($f\#$) constant as a function of imaging depth. The F-number is defined as the ratio of the imaging depth, z , to the aperture size, α . The synthetic aperture is dynamically grown as a function of the imaging depth in order to keep $f\#$ constant. The number of lines, l , to consider in a window for focusing to a depth z are calculated using the following expression:

$$l(z) = \frac{z_k}{f\# \cdot \Delta x} \quad (11)$$

10

where z_k is the distance between the aperture and sample and Δx is the inter-element spacing. This equation is used to derive a set of *a priori* constants that are stored in memory to allow for real-time dynamic apodization.

15 Synthetic aperture beamforming arrangement (analogue-domain demodulation)

Referring to Figure 3, a first arrangement of imaging device 3 is shown. For clarity, the excitation circuitry 17 is not shown in Figure 3. In the first arrangement, demodulation is carried out in the analogue domain.

20 The receive circuitry 18 includes an analogue front end 23 comprising M analogue front-end channels $23_1, \dots, 23_M$ and the digital processor 28 comprising M digital processing channels $28_1, \dots, 28_M$, where $N > M$ and $M \geq 1$, and N is the number of transducer elements.

25 Analogue front-end

The, or each, analogue front-end channel 23_i (where $i = 1, \dots, M$ and $M \geq 1$) includes a preamplifier 41, a signal splitter 43, first and second active or passive mixers 44, 45, an oscillator (not shown) and a 90° phase shifter (not shown) which generate first and second carrier signals 48, 49 for the first and second mixers 44, 45 respectively, first and second programmable gain amplifiers 52, 53, first and second low-pass filters 56, 57 and first and second analogue-to-digital converters 60, 61. The carrier signals 48, 49 may take the form of square, sine or other shape of wave which are orthogonal (i.e. phase-shifted by 90°) with respect to each other.

30

The preamplifier 41 may take the form of a fully differential preamplifier which functions as a low-noise amplifier. The preamplifier 41 performs time-gain compensation to account for exponential tissue attenuation. The gain is increased exponentially over time by controlling the gain of a variable gain amplifier (VGA). The
5 VGA sweeps the gain from, for example, 20 dB to 35 dB over the reflection period, thereby shifting the noise floor to an appropriate level.

After the signal 22 has been amplified, the amplified signal 42 is down-converted using the mixers 52, 53 into I and Q components 50, 51. The signals are processed along
10 separate, matched channels. The signal 50, 51 is then amplified again using the programmable gain amplifiers 52, 53. The required gain may be selected by switching between resistor combinations (not shown) on the amplifier's feedback loop (not shown). Lastly, image rejection is carried out by means of the low-pass filters 56, 57 which may each take the form of sixth-order Butterworth low-pass filter.

15 After analogue-to-digital conversion, the discretised I/Q signals 29, 30 are processed by the digital beamformer 28.

The analogue front end 23 may operate with a transducer centre frequency of 2.5 MHz, I/Q bandwidth of 1.25 MHz (Nyquist sampling frequency of 2.5 MHz), gain of 46 ± 6
20 dB, an input referred dynamic range at 1 kHz (THD < 1%) of 61 dB, and input referred noise floor of 7.5 μ V and a CMRR of 82 dB. These parameters can, however, be varied.

These parameters are variable depending on the resolution requirements of the
25 application. For low frequency imaging applications such as devices 81 and 31 in Figure 9, the centre frequency may be in the range of 2-20 MHz. The Nyquist sampling frequency depends on the bandwidth of the transducer and hence the I/Q signal bandwidth. The required receiver gain is a function of the power of the transmission pulse exciting to the transducer, which affects the pressure of the ultrasound wave. The
30 receiver gain is typically in the range of 10-100 dB. The noise floor and the total harmonic distortion of the receiver affects the dynamic range of the device. With time-gain control, the dynamic range may be, for example between 50-100 dB.

Beam-forming digital processor

Conventional synthetic aperture algorithms tend to be computationally intensive and typically require a large memory capacity. Thus, a serialised or pipelined approach is used herein to process data dynamically, in real-time.

- 5 The digital processor 28 carries out beamforming including dynamically focusing and apodizing data in order to form a 2D image. The digital processor 28 may take the form of an application-specific integrated circuit (ASIC) or field-programmable gate array (FPGA).
- 10 Referring still to Figure 3, the, or each, digital processing channel 28_i (where $i = 1, \dots, M$ and $M \geq 1$) includes an interpolation module 62, a phase rotate module 63 and a summing module 64. The interpolation module 62 includes, or has access to, memory 65, for example in the form of read-only memory, which holds a look-up table 66.
- 15 The modules 62, 63, 64 may be implemented in hardware.

Referring also to Figure 4, after initialisation, first transmission is carried out and I/Q signals 29, 30 are sampled and read into memory 31. At the end of a reflection period, calculations begin and then continue during a following reflection period. Calculations

20 on parallel groups of pixels are pipelined over multiple clocks cycles. The number of parallel operations that may be carried out depends on the logic capacity of the device and this, in turn, determines the maximum frame rate and image size. Delay values, $t_p(i, j)$, are read from the look-up table 66 stored in read-only memory 65, and used in the interpolation process hereinbefore described above. Dynamic apodization is also

25 applied by first reading values of $I(z)$ calculated *a priori* and stored in read-only memory 65. These values are used to dynamically apodize the receive aperture in real-time. Finally, after the pixel value is calculated, it is added to a global 2D image array stored in random access memory 31. At the end of one iterative cycle, a high-resolution frame 39 is formed and the process enters a write state in which the image is

30 transmitted to a backend processor (not shown), for example, for log compression.

If the beamformer is implemented in an ASIC, it can be synthesized in Cadence (RTM) using, for example, an AMS 0.18 μm CMOS process. A clock frequency of 25 MHz can be used, $N_t = 8$ (i.e. eight transducers) and pixel resolution 32×350 . For these

35 parameters, the estimated maximum frame rate is 4 Hz, the estimated beamformer size

is $1.55 \text{ mm} \times 1.55 \text{ mm}$ and the estimated power consumption is 15 mW. These parameters and characteristics can, however, be varied.

Simulation results

5 Raw ultrasound data are captured using a Verasonics (RTM) Vantage 256 system using a Philips (RTM) P4-1 phased array (central frequency at 2.9 MHz) with 96 active elements. The imaged medium takes the form a phantom containing 8×3 cross-sectional wires.

10 The synthetic aperture beamforming process is simulated using in MATLAB (RTM) by mixing the RF signals which are sampled at 10 MHz with 2.5 MHz sine and cosine references and filtering the result with a 4th order Butterworth low-pass filter having a cut-off frequency of 1.25 MHz. The digital beamforming process hereinbefore described is implemented in Verilog (RTM) and simulated in an HDL simulation
15 environment in the form of ModelSim (RTM).

Referring to Figure 5, first, second and third two-dimensional (2D)-mode (or “B-mode”) ultrasound images are shown.

20 The first 2D-mode image is obtained using quadrature beamforming carried out with 8 transmit elements. The second 2D-mode image is obtained using quadrature beamforming carried out with 48 transmit elements. The third 2D-mode image is a comparative example which is obtained using beamforming carried out in the RF-domain.

25 The signal-to-noise ratio of the third 2D-mode ultrasound image (i.e. obtained using beamforming carried out in the RF-domain) is practically identical to that of the quadrature image, for the same number of transmissions and F-number. Decreasing the number of transmissions (i.e. decreasing N_t) leads to a reduction in the signal-to-
30 noise ratio due to larger side lobes (not shown) and increased speckle noise. Furthermore, the signal-to-noise and lateral/axial resolution decrease as a function of depth. Nevertheless, satisfactory images can be obtained using only 8 transmit elements.

35 *Measured results*

Measured results were obtained using the architecture shown in Figure 3.

Raw ultrasound data were captured on a Verasonics Vantage 256 system using a P4-1 phased array (central frequency at 2.5 MHz). The synthetic aperture method was used, and RF signals were sampled at 10 MHz. The imaged medium contained a hyperechoic phantom and had an attenuation rate of 0.7 dB/cm/MHz. The RF signals were multiplexed through an analogue front-end (implemented in AMS 0.35µm CMOS) using an arbitrary waveform generator (Picoscope 5442B). Beamforming was carried out on field programmable gate array (FPGA) and the image was transmitted to a computer system for post-processing. Finally, envelope detection was carried out in MATLAB and the image was logarithmically compressed.

Figure 12 shows first, second and third measured B-mode ultrasound images obtained using RF-domain synthetic aperture beamforming (48 transmit elements), in-phase and quadrature beamforming (48 and 15 transmit elements respectively).

Synthetic aperture beamforming arrangement (digital-domain demodulation)
Referring to Figure 13, a second arrangement of imaging device 3 is shown.

The second arrangement is similar to the first arrangement, except that demodulation is not carried out in the analogue domain, but in the digital domain.

Analogue front-end

The, or each, analogue front-end channel 23_i (where $i = 1, \dots, M$ and $M \geq 1$) includes a preamplifier 41, a programmable gain amplifier 111 which generates an amplified RF signal 112, a low-pass filter 113 and an analogue-to-digital converter 60 which outputs a digitised RF signal 40.

Beam-forming digital processor

The digital processor 28 carries out beamforming including dynamically focusing and apodizing data in order to form a 2D image. The digital processor 28 may take the form of an application-specific integrated circuit (ASIC) or field-programmable gate array (FPGA).

The digital processor 28 includes a demodulator stage 114 which receives the discretised RF signal 40 and generates discretised I/Q signals 29, 30. The digital processor 28 performs beamforming substantially as hereinbefore described.

Compressive synthetic aperture imaging within an FRI framework

Overview

A sampling paradigm for certain classes of parametric signals can be used. Parametric
 5 signals with k parameters may be sampled and reconstructed using only $2k$ parameters.
 These signals have a finite rate of innovation (FRI). A sampling scheme can be applied
 to periodic and finite streams of FRI signals, such as Dirac impulses, non-uniform
 splines, and piecewise polynomials. An appropriate sample kernel, such as sinc,
 Gaussian, sum of sincs, etc., can be applied to extract a set of Fourier coefficients which
 10 are then used to obtain an annihilating filter. The locations and amplitudes of the
 pulses are finally determined. A brief review can be found in M. Vetterli, P. Marziliano,
 and T. Blu: "Sampling signals with finite rate of innovation," IEEE Transactions on
 Signal Processing, volume 50, pages 1417–1428 (2002).

15 Consider an FRI signal $x(t)$, for example, an ultrasound A-mode signal, comprising a
 finite stream of pulses with pulse shape $p(t)$, amplitudes $\{c_k\}_{k=0}^{K-1}$ and time locations
 $\{t_k\}_{k=0}^{K-1}$:

$$x(t) = \sum_{k=0}^{K-1} c_k p(t - t_k) \quad (12)$$

20 The sample values are obtained by filtering the signal with a sampling kernel. A sinc
 kernel is defined as $h_B(t) = B \text{sinc}(Bt)$, with bandwidth $B = 1/T$. The convolution
 product is:

$$y_n = \langle h_B(t - nT), x(t) \rangle \quad n = 0, \dots, N-1 \quad (13)$$

25 This is equivalent to:

$$y_n = \sum_{k=0}^{K-1} c_k B \text{sinc}\left(\frac{t_k}{T} - n\right) \quad (14)$$

$$= (-1)^n \sum_{k=0}^{K-1} \frac{c_k B \text{sinc}\left(\frac{\pi t_k}{T}\right)}{\pi \left(\frac{t_k}{T} - n\right)} \quad (15)$$

$$(16)$$

$$\Leftrightarrow (-1)^n y_n = \frac{1}{\pi} \sum_{k=0}^{K-1} c_k B \frac{\sin\left(\frac{\pi t_k}{T}\right)}{\left(\frac{t_k}{T} - n\right)}$$

Since the signal has K degrees of freedom, $N \geq 2K$ samples are required to sufficiently recover the signal. The reconstruction method requires two systems of linear equations, namely one for the locations of the Gaussian pulses involving a matrix V and one for the weights of the pulses involving a matrix A . Define a Lagrange polynomial $L_k =$
 5 $(P(u)/(u - t_k/T))$ of degree $K - 1$, where $P(u) = \prod_{k=0}^{K-1} (u - t_k/T)$. Multiplying both sides of equation 16 by $P(n)$ yields an expression in terms of the interpolating polynomials:

$$\underbrace{(-1)^{n+1} P(n) y_n}_{Y_n} = \sum_{k=0}^{K-1} c_k \underbrace{B \sin\left(\frac{\pi t_k}{T}\right) \frac{L_k(n)}{\pi}}_{[A]_{nk}} \quad (17)$$

$$\Leftrightarrow \mathbf{Y} = \mathbf{A} \cdot \mathbf{c} \quad (18)$$

10

To find the K locations t_k , i.e. the time delays of the pulses, an annihilating equation is derived to find the roots of $P(u)$. Since the right-hand side of equation 17 is a polynomial of degree $K - 1$ in the variable n , if K finite differences are applied, then the left-hand side will become zero, i.e. $\Delta^K((-1)^n P(n) y_n) = 0$, $n = K, \dots, N - 1$. Letting

15 $P(u) = \sum_k p_k u^k$ leads to an annihilating filter equation equal to:

$$\sum_{k=0}^K p_k \underbrace{\Delta^K((-1)^n n^k y_n)}_{[V]_{nk}} = 0 \quad (19)$$

$$\Leftrightarrow \mathbf{V} \cdot \mathbf{p} = 0 \quad (20)$$

where $\mathbf{V}$ is an $(N - K) \times (K + 1)$ matrix. The system admits a solution when $\text{Rank}(\mathbf{V}) \leq$
 20 K and $N \geq 2K$. Thus, equation 19 may be used to find the $K + 1$ unknowns p_k , which leads to K locations t_k as these are the roots of $P(u)$. Once the locations have been determined, the weights of the Gaussian pulses c_k may be found by solving the system in equation 18 for $n = 0, \dots, K - 1$. The system has no solution if $\text{Rank}(\mathbf{A}) = K$, where $A = \in \mathbb{R}^{K \times K}$ is defined by equation 17. A more detailed discussion of the annihilating
 25 filter method is provided in M. Vetterli *et al. ibid.*. Theoretically, the result does not

depend on the sampling period T . However, $\mathbf{V}$ may be poorly conditioned if T is not chosen appropriately. As simulation results show hereinafter, oversampling yields an increase in the signal-to-noise of the reconstructed result.

5 The sinc kernel herein before described has infinite time support and is non-causal. In the frequency domain, it is represented by an ideal low-pass filter with an infinite roll-off. Practically, the sinc kernel may be approximated in hardware by means a high-order analogue low-pass filter. Simulations demonstrate the performance of multiple filter types and orders.

10

Compressive synthetic aperture beamforming arrangement

Referring to Figure 5, a second arrangement of imaging device 3 is shown. For clarity, the excitation circuitry 17 is not shown in Figure 6.

15 The receive circuitry 18 includes an analogue front-end 23 which includes a preamplifier 41, a signal splitter 43, first and second passive mixers 44, 45, an oscillator (not shown) and a 90° phase shifter (not shown) which generate first and second carrier signals 48, 49 for the first and second mixers 44, 45 respectively, first and second programmable gain amplifiers 52, 53, first and second low-pass filters 56, 57 and first and second
20 analogue-to-digital converters 60, 61. The carrier signals 48, 49 may take the form of square, sine or other shape of wave which are orthogonal (i.e. phase-shifted by 90°) with respect to each other.

The analogue front-end 23 amplifies and demodulates the RF signal 22 into I and Q
25 components. This is achieved by mixing the RF waveform 22 with reference signals 48, 49 centred at the carrier frequency. The assumption is made that both the I and Q signals satisfy the FRI criterion, namely they both have finite rates of innovation. The signals are then filtered and bandlimited below the original I/Q bandwidth. This is carried out in the analogue domain in order to reduce the sampling frequency and thus
30 the data bandwidth. This leads to a significant power saving, as the power budget is predominated by the power consumption of the ADC and wireless transceiver. By compressing the signal in the analogue domain, the computational burden is shifted to the digital back end, which carries out reconstruction of the I/Q signals and finally baseband beamforming.

35

Simulation results

Raw ultrasound data are captured using a Verasonics (RTM) Vantage 256 system using a Philips (RTM) P4-1 phased array (central frequency at 2.9 MHz) with 96 active elements. The imaged medium takes the form a phantom containing 8×3 cross-sectional wires.

5

The compressive synthetic aperture beamforming processes is used to produce a full two-dimensional mode image from the RF dataset hereinbefore described. First, the RF dataset was filtered using a fourth-order low-pass filter kernel with K values of 10, 20 and 40. Given $F = 4$ and $\tau = 140.8 \mu\text{s}$, the corresponding low-rate sampling frequencies are 575 kHz, 1.14 MHz and 2.28 MHz respectively. The I and Q components are reconstructed using the process hereinbefore described. Finally, synthetic aperture beamforming is carried out using the quadrature method process hereinbefore described.

10

15 Referring to Figure 7, fourth, fifth and sixth two-dimensional (2D)-mode (or “B-mode”) ultrasound images are shown.

20

Increasing L improves the reconstruction accuracy and, thus, lateral resolution and image quality. As hereinbefore described, increasing L eventually pushes the low-rate sampling above that of the Nyquist quadrature sampling frequency. Ideally L should be as small as possible to minimise the sampling rate, and thus the power consumption of the transmission link.

Devices

25

Referring to Figure 8, an endoscopy capsule 81 is shown.

30

The capsule 81 comprises a generally cylindrically-shaped sealed case 82 having rounded ends and having dimensions and shape which allows the capsule to be swallowed by a subject, such as a patient. The ultrasonic imaging device 3 is housed within the case 82 such that the transducer array 11 (Figure 1) is suitably positioned for imaging surrounding tissue, for example, in an annular section (not shown) around the perimeter of the case. The case 82 has an outer diameter, d , and a length, l . The case 82 may have dimensions which allow a capsule to be swallowed. For example, d may be 1 mm and l may be about 2 cm. The capsule 81 may be swallowed by patient and while it passes through the gastrointestinal tract (not shown), it can perform ultrasound and transmit signals to the processing device 4 (Figure 1). The capsule 81

35

may be used in veterinary applications, in other words, the subject may be a non-human animal. A capsule-like device may be used in non-medical or non-veterinary applications, such as inspection of pipes.

5 Referring to Figure 9, a wearable patch 91 is shown.

The patch 91 may comprises a flexible substrate 92 and the ultrasonic imaging device 3 is housed within a pocket (not shown) formed in part by the flexible substrate 92 or is attached to the flexible substrate 92. A face (not shown) of the substrate 92 may be
10 coated with an adhesive (not shown) which allows the patch 91 to be attached to a surface 93 of a subject, for example, the skin of a patient. The transducer array 11 (Figure 1) suitably positioned for imaging through the surface 93. The ultrasonic imaging device 3 has lateral dimensions, w , which allow the patch to be worn. For example, w may be between 1 to 4 cm.

15

Referring to Figure 10, a wand 101 is shown.

The wand 101 comprise a cylindrical or bar-like case 102. The ultrasonic imaging device 3 is housed within the case 102, for example, occupying the full length 103 of the
20 cylindrical case 102. The transducer array 11 (Figure 1) is suitably positioned, for example on the front end or tip of the device 102 facing the surface 104. The transducer array 11 may be either a 1D or 2D array providing 2D or 3D imaging respectively. The wand 101 may be scanned over the surface 104 of a subject, such as a patient. Thus, the wand 101 can be used by medical workers (such as doctors, nurses or paramedics) as a
25 convenient point-of-care diagnostic device. The wand 101 may have similar dimensions as a pen, for example, having a diameter, $2r$, of about 1 cm or a handheld instrument, for example, having a diameter, $2r$, of about 1 to 3 cm.

Modifications

30 It will be appreciated that various modifications may be made to the embodiments hereinbefore described. Such modifications may involve equivalent and other features which are already known in the design, manufacture and use of ultrasonic imaging devices and component parts thereof and which may be used instead of or in addition to features already described herein. Features of one embodiment may be replaced or
35 supplemented by features of another embodiment.

The ultrasonic imaging device 3 may transmit the signals directly to the processing system 4. However, the ultrasonic imaging device 3 may transmit the signals via, for example, an antenna (not shown) which is located remotely from the processing device 4, via a wireless repeater (not shown), via a wired link (not shown) and/or wireless link (not shown) such as a wireless local area network.

Although claims have been formulated in this application to particular combinations of features, it should be understood that the scope of the disclosure of the present invention also includes any novel features or any novel combination of features disclosed herein either explicitly or implicitly or any generalization thereof, whether or not it relates to the same invention as presently claimed in any claim and whether or not it mitigates any or all of the same technical problems as does the present invention. The applicants hereby give notice that new claims may be formulated to such features and/or combinations of such features during the prosecution of the present application or of any further application derived therefrom.

Claims

1. An ultrasonic imaging device comprising:
an ultrasonic transducer array (11) comprising a plurality of transducer elements
5 (12);
a multiplexer (15) comprising a plurality of inputs and at least one output, each
input coupled to a respective transducer element;
an analogue front-end (23) comprising at least one channel, each channel coupled
to a respective output of the multiplexer and each channel including a respective
10 analogue demodulator (44, 45) arranged to extract in-phase and quadrature
components of a signal from the transducer and to provide demodulated in-phase and
quadrature signal components (50, 51);
at least two analogue-to-digital converters (60, 61) configured to receive signals
comprising or obtained from the in-phase and quadrature signal components and to
15 provide digitised demodulated in-phase and quadrature signal components; and
a wireless interface (32) configured to transmit signals comprising or obtained
from the digitised demodulated in-phase and quadrature signal components.
2. An ultrasonic imaging device comprising:
20 an ultrasonic transducer array (11) comprising a plurality of transducer elements
(12);
a multiplexer (15) comprising a plurality of inputs and at least one output, each
input coupled to a respective transducer element;
an analogue front-end (23) comprising at least one channel, each channel coupled
25 to a respective output of the multiplexer, each channel arranged to provide an RF signal
(40) in dependence on a signal from the transducer;
at least one analogue-to-digital converters (60) configured to receive a signal
comprising or obtained from the RF signal and to provide a digitised RF signal;
a digital processor (28) arranged to extract in-phase and quadrature components
30 from the digitised RF signal and to provide digitised demodulated in-phase and
quadrature signal components, and
a wireless interface (32) configured to transmit signals comprising or obtained
from the digitised demodulated in-phase and quadrature signal components.

3. An ultrasonic imaging device according to claim 1, wherein each channel of the analogue front-end (23) includes first and second low-pass filters (56, 57) arranged to filter the in-phase and quadrature signal components respectively before the in-phase and quadrature signal components are digitised by the analogue-to-digital converters (60, 61).
5
4. An ultrasonic imaging device according to claim 3, wherein the first and second low-pass filters (56, 57) have respective bandwidths which are less than the Nyquist cut-off frequency of a transducer element (12).
10
5. An ultrasonic imaging device according to claim 1, wherein the least two analogue-to-digital converters (60, 61) are configured to sample at a non-uniform rate.
6. An ultrasonic imaging device according to any one of claims 1 or 2 to 4, wherein the analogue-to-digital converters are configured to supply digitised demodulated in-phase and quadrature signal components and wherein wireless interface (32) is configured to transmit the digitised demodulated in-phase and quadrature signal components.
15
7. An ultrasonic imaging device according to any one of claims 1 to 5, further comprising:
20
a processor (28) arranged, for each channel, to receive the digitised demodulated in-phase and quadrature signal components and to perform beamforming based on the digitised demodulated in-phase and quadrature signal components.
25
8. An ultrasonic imaging device according to claim 7, wherein the processor (28) is arranged to carry out interpolation and to provide interpolated in-phase and quadrature signal values.
9. An ultrasonic imaging device according to claim 7 or 8, wherein the processor (28) is configured to apply a time delay to the interpolated in-phase and quadrature signal values.
30
10. An ultrasonic imaging device according to claim 8 or 9, wherein the processor (28) is arranged to carry out phase rotation of the interpolated in-phase and quadrature signal values to produce first and second RF signal values.
35

11. An ultrasonic imaging device according to claim 10, wherein the processor (28) is arranged to carry out summation of the first and second RF signal values to a given memory location corresponding to a given scan-line sample or pixel.
- 5 12. An ultrasonic imaging device according to claim 11, wherein the processor (28) is arranged to repeat interpolation, phase rotation and summation for a series of digitised RF signals or digitised demodulated in-phase and quadrature signal components.
- 10 13. An ultrasonic imaging device according to any one of claims 7 to 12, further comprising:
memory (28) for storing a two-dimensional image comprising pixel values received from the processor.
- 15 14. An ultrasonic imaging device according to any one of claims 7 to 13, wherein wireless interface (32) is configured to transmit two- or three-dimensional image.
15. An ultrasonic imaging device according to any one of claims 1 to 14, further comprising:
20 an excitation circuit (17) comprising a pulser (20), a transmit beamformer (21) and a processor configured to control the beamformer.
16. An ultrasonic imaging device according to claim 15, wherein the processor of the excitation circuit is configured to perform synthetic aperture beamforming or to
25 perform transmit focusing to form focused, plane wave or defocused transmit beams for synthetic aperture imaging.
17. An ultrasonic imaging device according to any one of claims 1 to 16, wherein the analogue front-end (23) comprises one channel.
- 30 18. An ultrasonic imaging device according to any one of claims 1 to 17, comprising:
a housing which contains the ultrasonic transducer array (11), the multiplexer (15), the analogue front-end (23), the at least one analogue-to-digital converter or the at least two analogue-to-digital converters (60, 61) and the wireless interface (32) and
35 wherein the housing has a volume less than 2 cm³.

19. An ultrasonic imaging device according to any one of claims 1 to 18, which adapted to be a capsule (81) for swallowing by a human subject or a non-human animal subject or for passing through a passage of a non-animal subject.
- 5 20. An ultrasonic imaging device according to any one of claims 1 to 18, which adapted to be a patch (91) for applying to a surface (93) of a subject.
21. An ultrasonic imaging device according to any one of claims 1 to 18, which adapted to be a hand-held wand for scanning over a surface (93) of a subject.
- 10 22. A processing device comprising:
a network interface (33); and
a processor (34) coupled to the network interface;
wherein the processor is configured to receive parametric or filtered quasi in-phase and
15 quadrature signals, to reconstruct in-phase and quadrature signals and to perform
synthetic aperture beamforming.
23. An ultrasonic imaging system (1) comprising:
an ultrasonic imaging device (3) according to any one of claims 1 to 22; and
20 a processing device (4) comprising a network interface (33), a processor (34),
storage (35) and a display (36),
wherein the ultrasonic imaging device and the processing device are in wireless
communication.

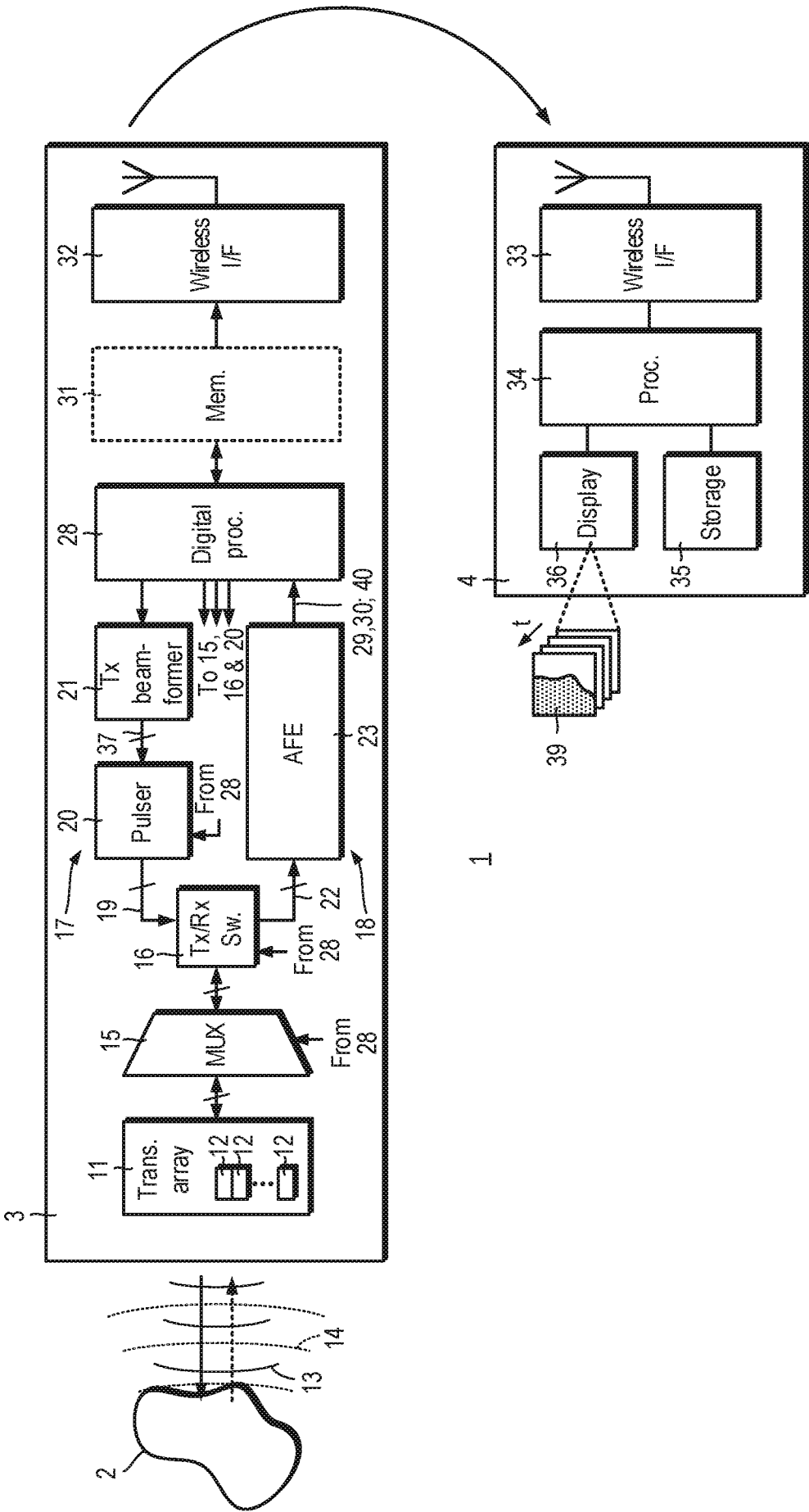


Fig. 1

Receive synthetic aperture processing (where $M = 1$)

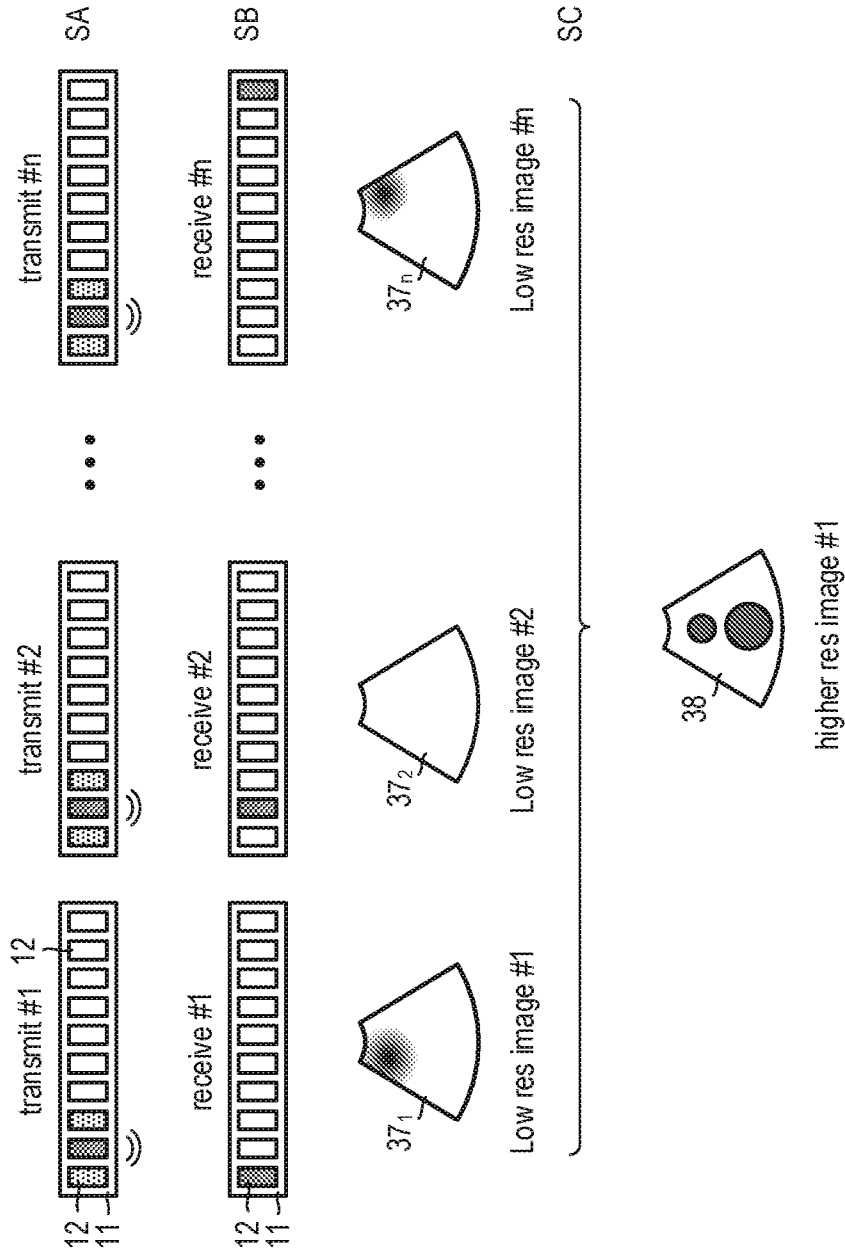


Fig. 2

3/11

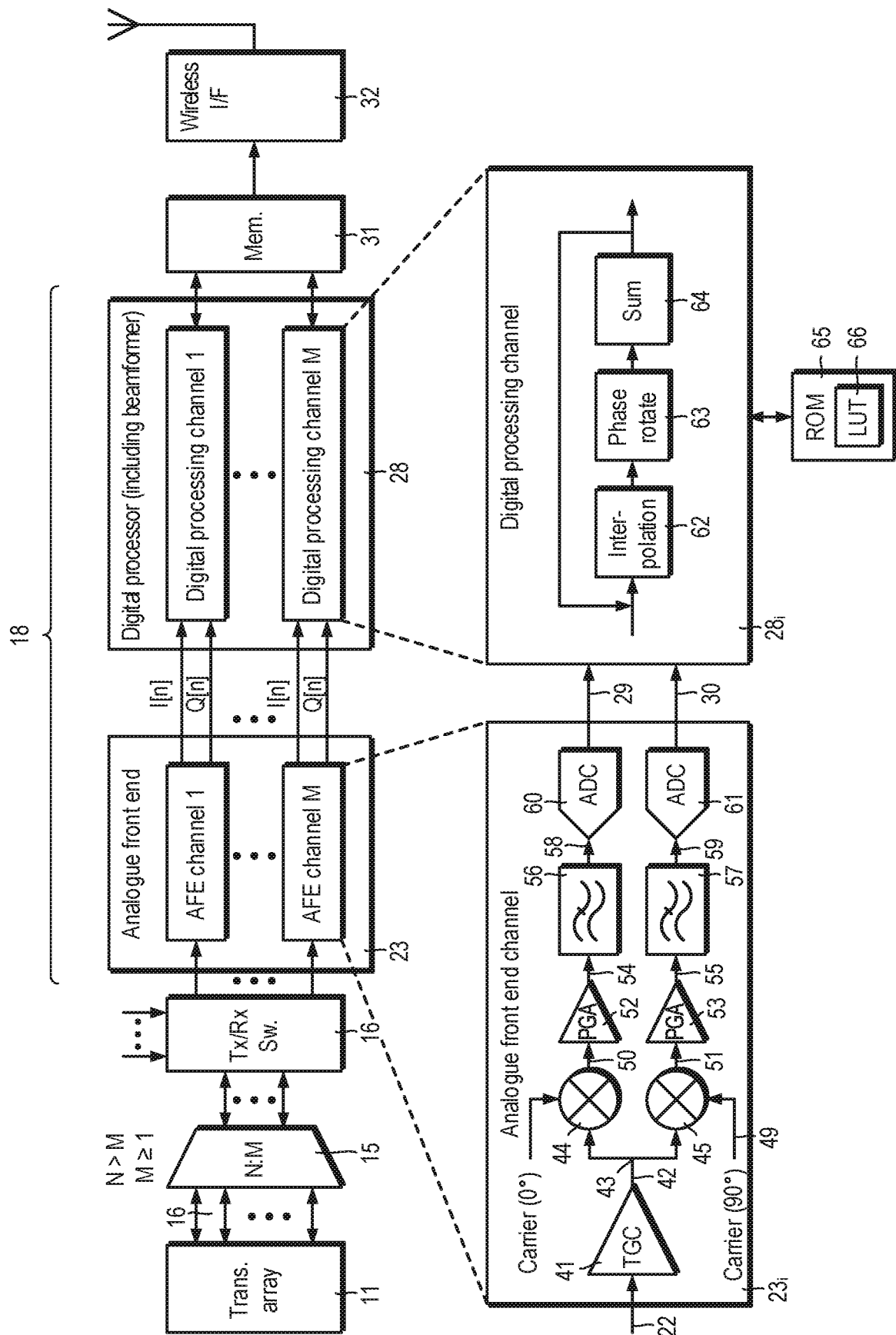


Fig. 3

4/11

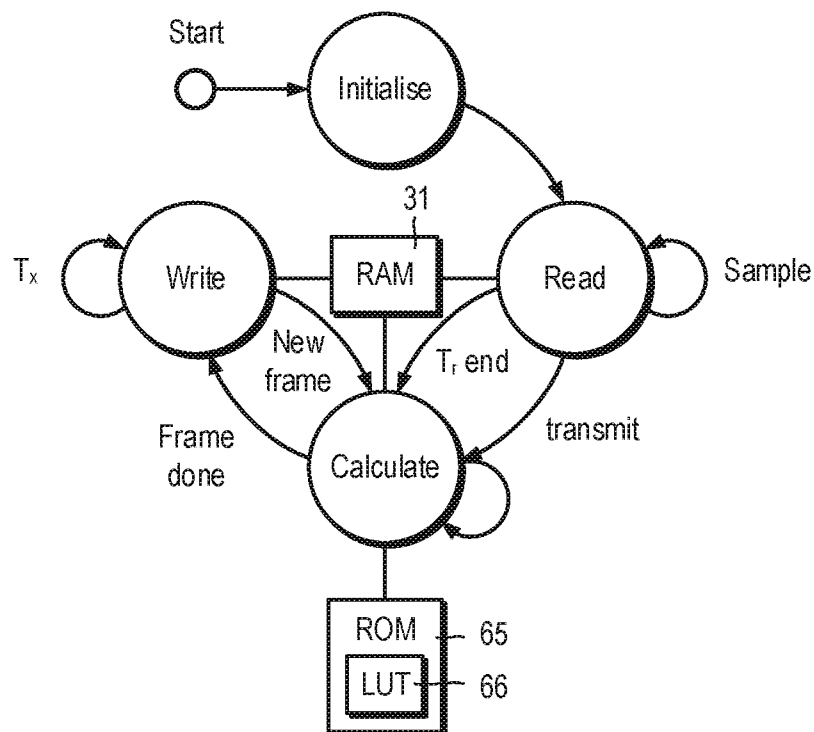


Fig. 4

5/11

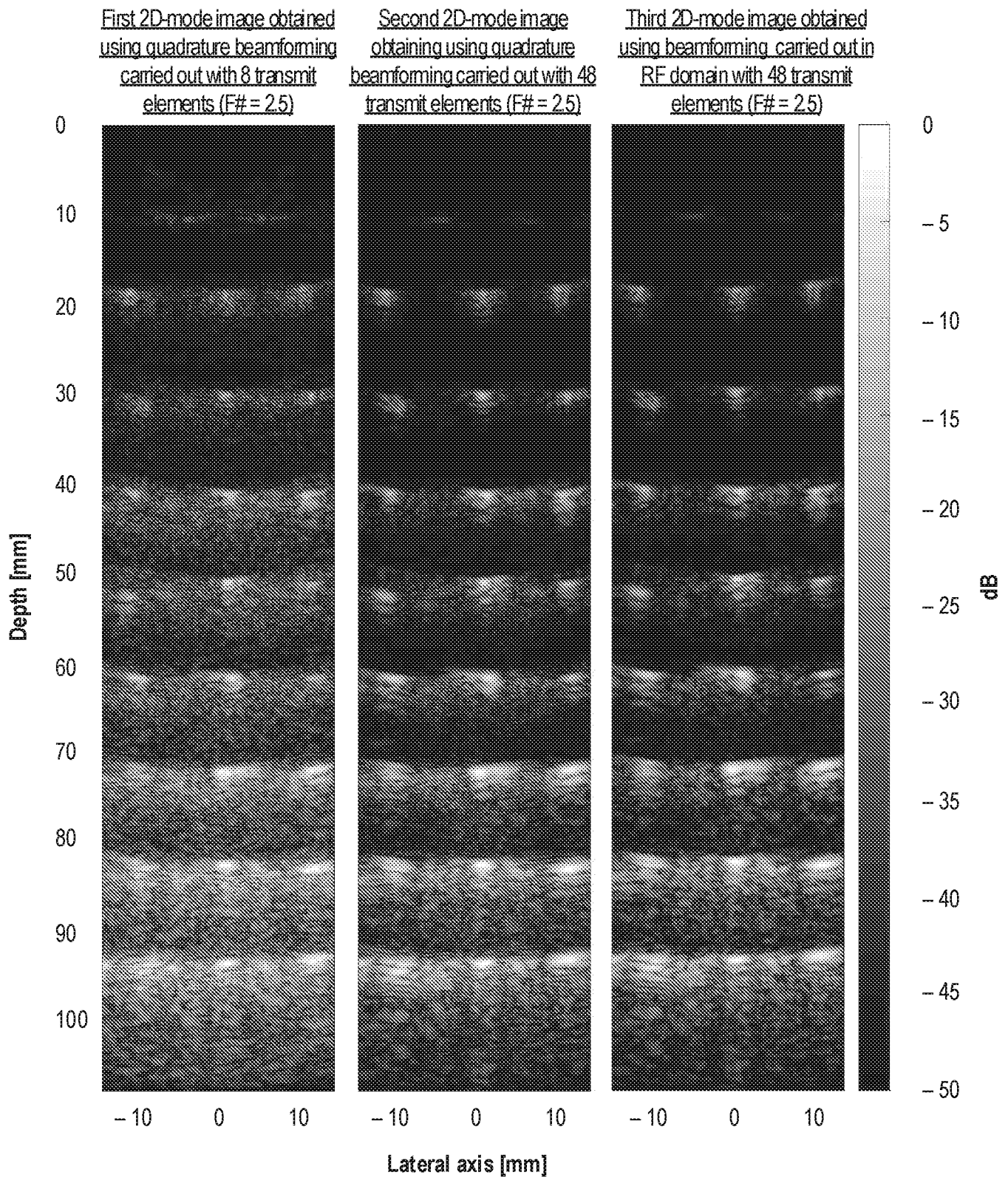


Fig. 5

6/11

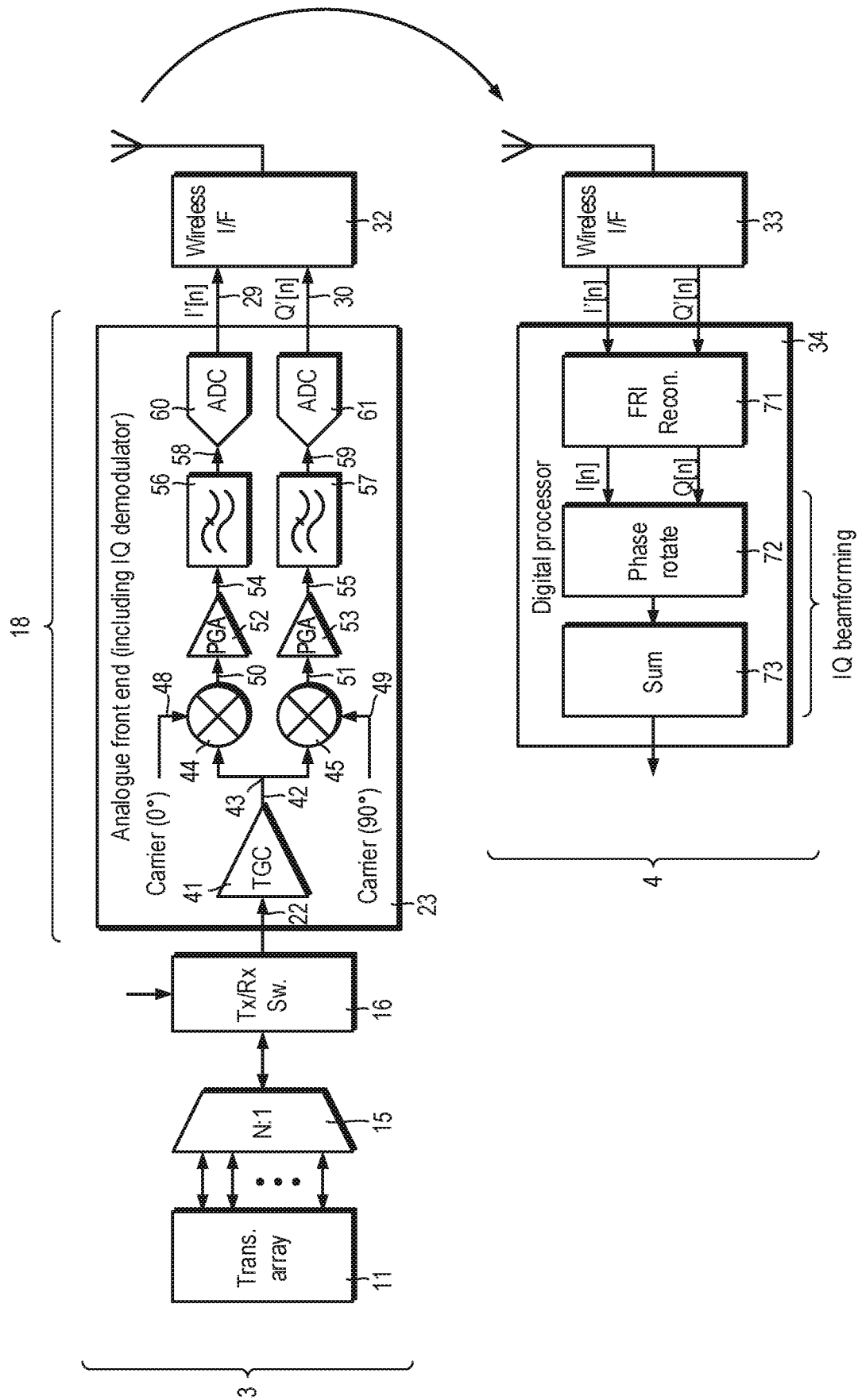


Fig. 6

7/11

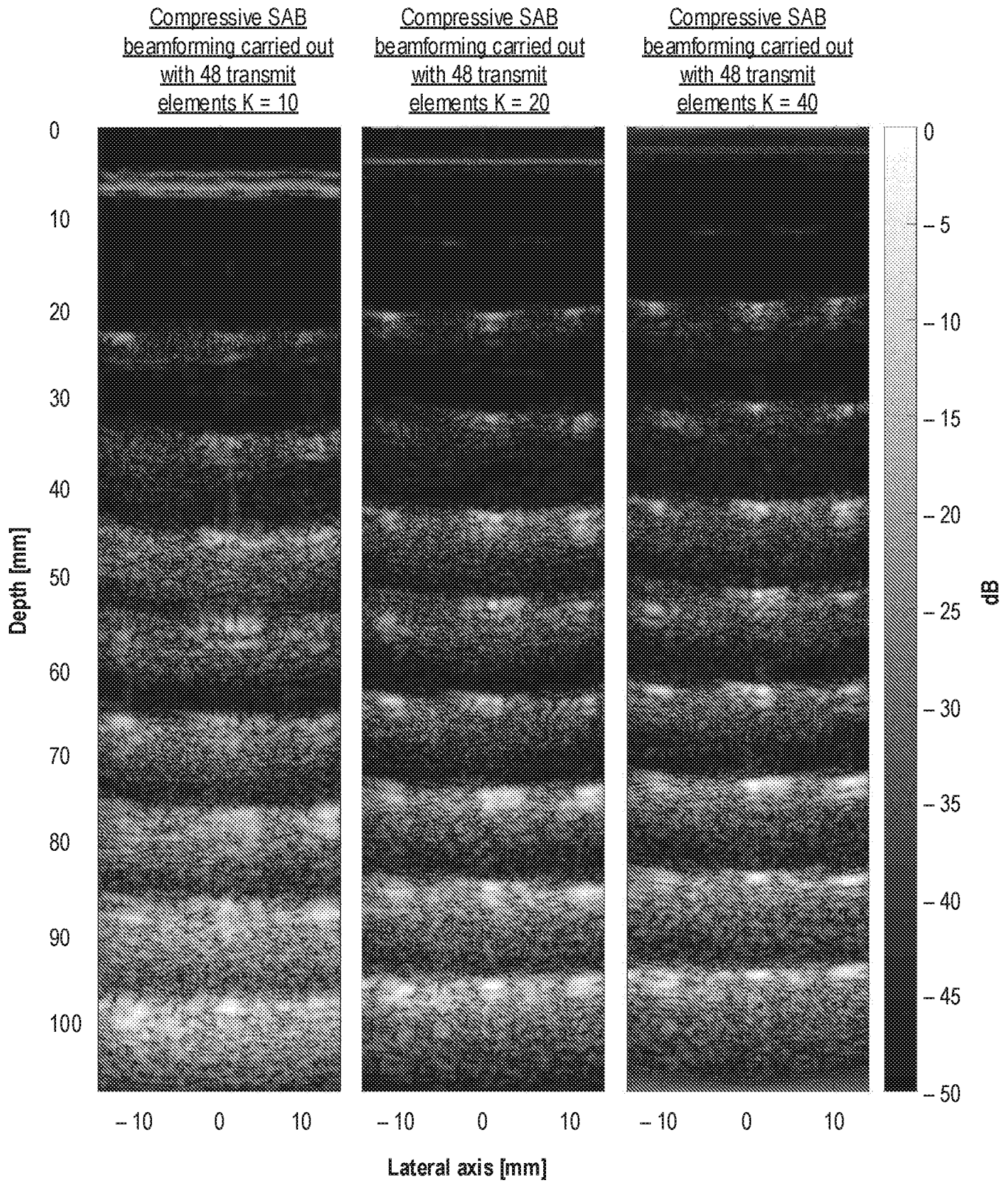


Fig. 7

8/11

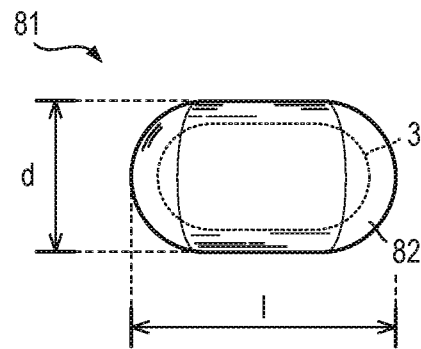


Fig. 8

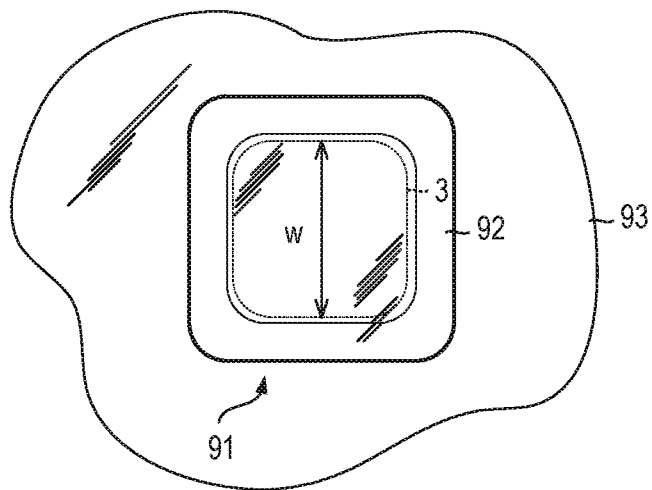


Fig. 9

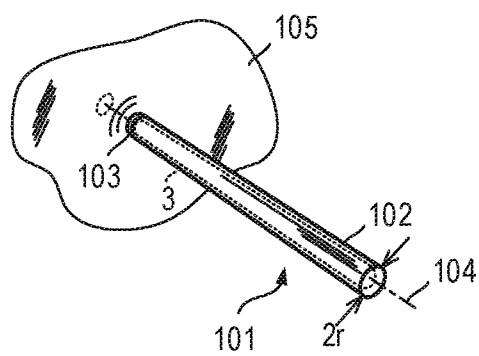


Fig. 10

Synthetic phased subarray beamforming

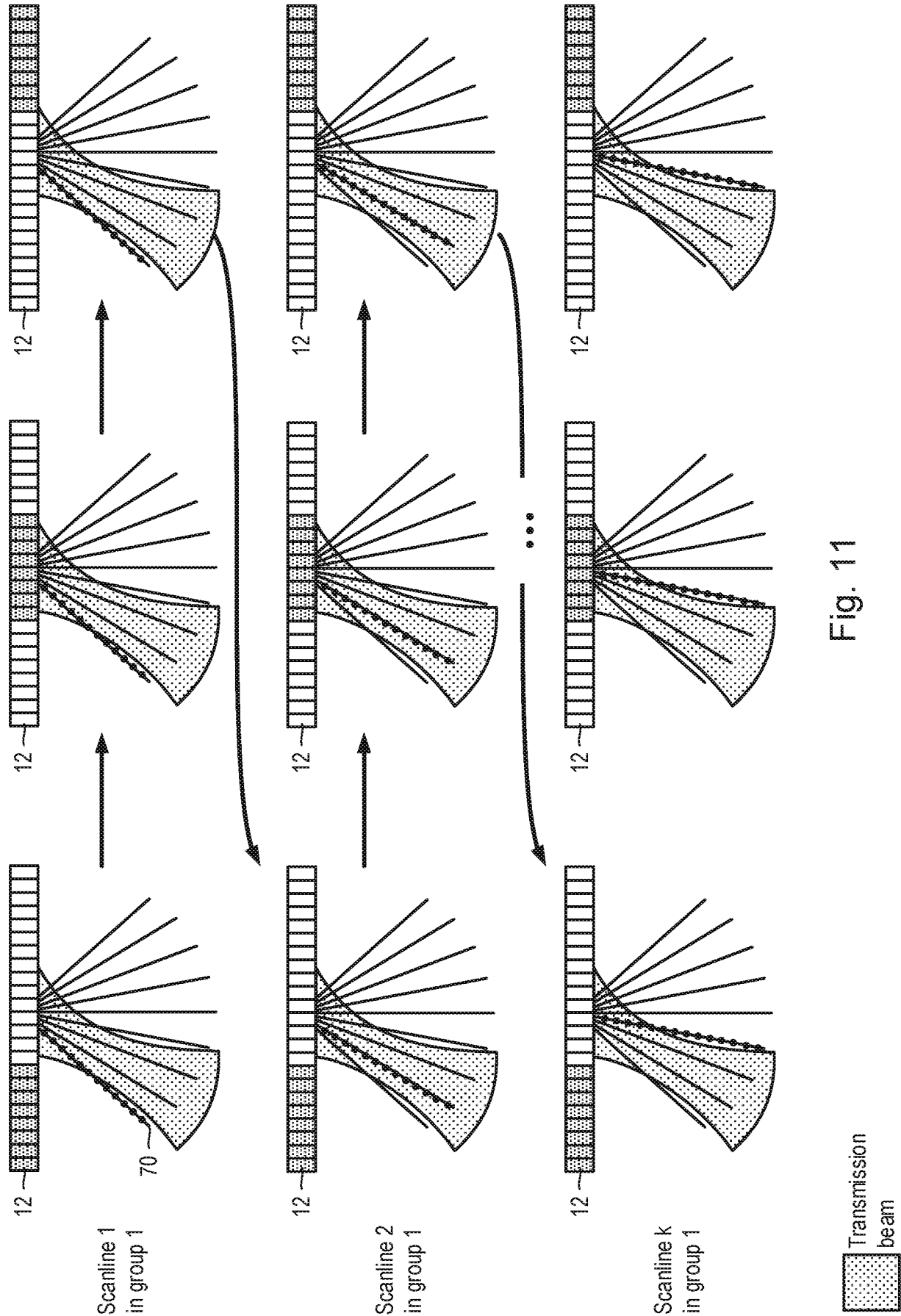


Fig. 11

10/11

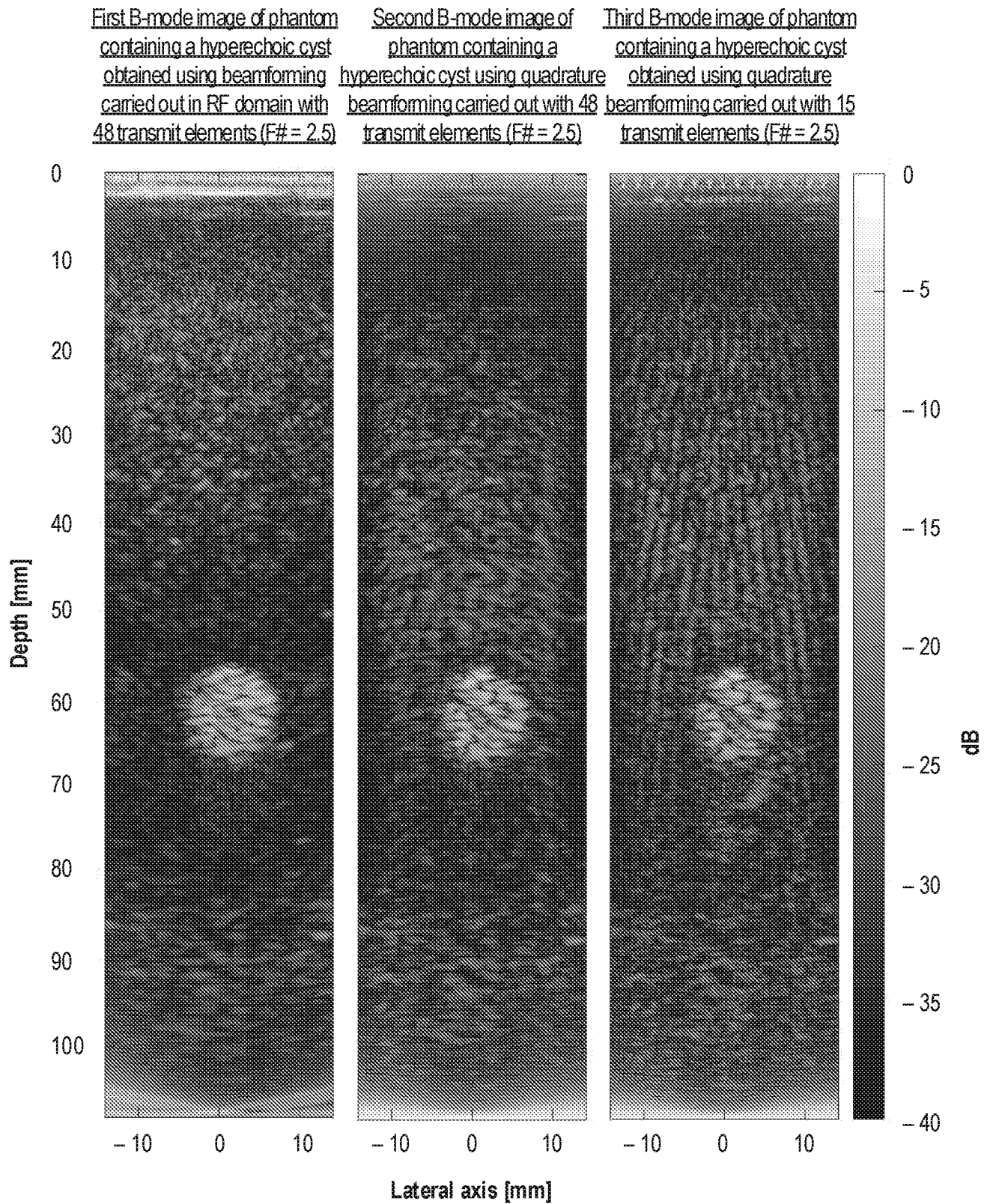


Fig. 12

11/11

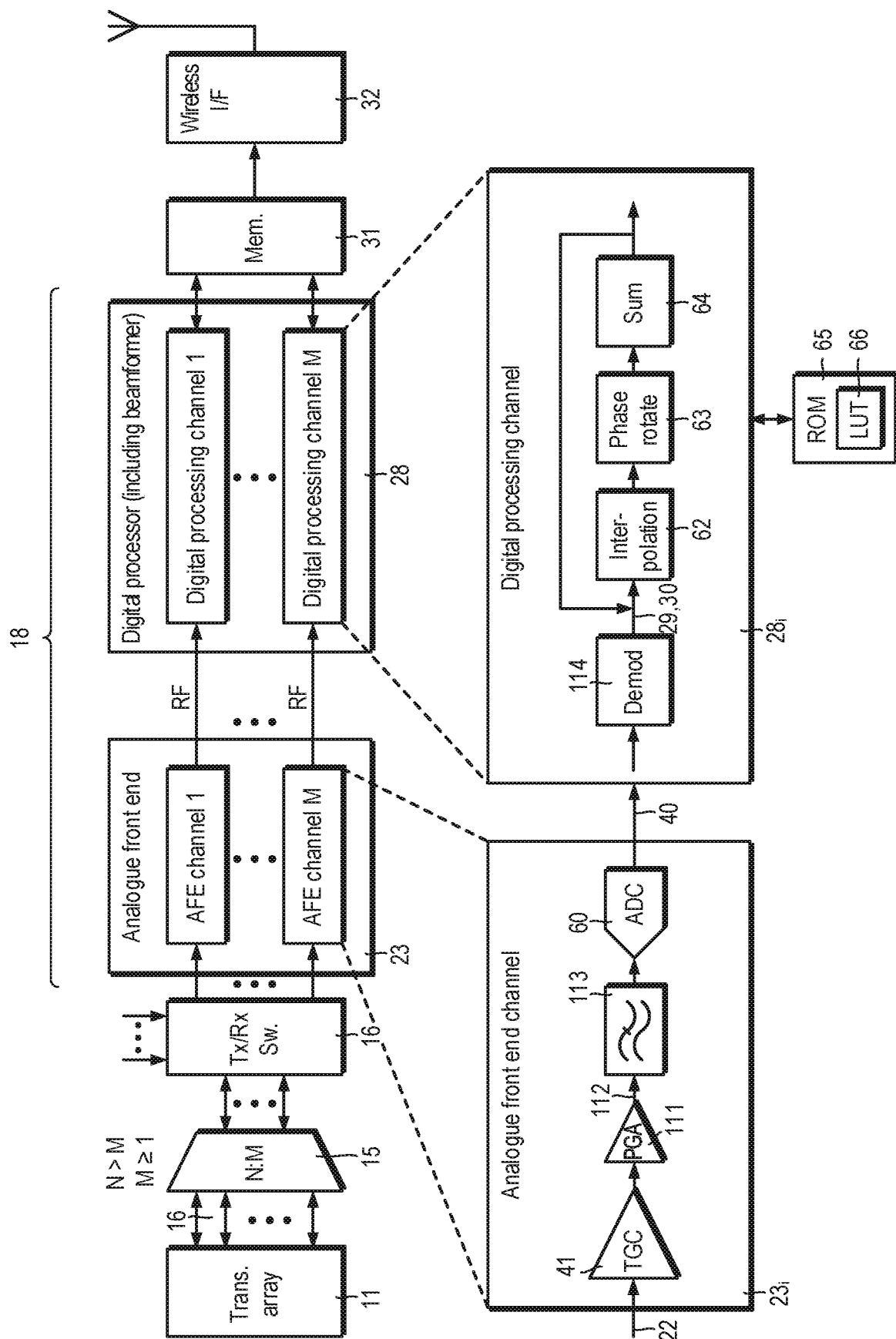


Fig. 13

INTERNATIONAL SEARCH REPORT

International application No
PCT/GB2017/053761

A. CLASSIFICATION OF SUBJECT MATTER INV. A61B8/00 A61B8/12 G01S7/52 A61B8/08 G01S15/89 ADD.											
According to International Patent Classification (IPC) or to both national classification and IPC											
B. FIELDS SEARCHED Minimum documentation searched (classification system followed by classification symbols) A61B G01S Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched Electronic data base consulted during the international search (name of data base and, where practicable, search terms used) EPO-Internal, WPI Data											
C. DOCUMENTS CONSIDERED TO BE RELEVANT <table border="1"> <thead> <tr> <th>Category*</th> <th>Citation of document, with indication, where appropriate, of the relevant passages</th> <th>Relevant to claim No.</th> </tr> </thead> <tbody> <tr> <td>X</td> <td>PARK MINSUK ET AL: "A new post-phase rotation based dynamic receive beamforming architecture for smartphone-based wireless ultrasound imaging", PROGRESS IN BIOMEDICAL OPTICS AND IMAGING, SPIE - INTERNATIONAL SOCIETY FOR OPTICAL ENGINEERING, BELLINGHAM, WA, US, vol. 9790, 1 April 2016 (2016-04-01), pages 97901V-97901V, XP060065679, ISSN: 1605-7422, DOI: 10.1117/12.2216705 ISBN: 978-1-5106-0027-0</td> <td>2,6-14, 17-23</td> </tr> <tr> <td>Y</td> <td>abstract; figures 1,2 pages 97901V-2, paragraph 1 - pages 97901V-4, paragraph 7 ----- -/-</td> <td>1,3-5, 15,16</td> </tr> </tbody> </table>			Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.	X	PARK MINSUK ET AL: "A new post-phase rotation based dynamic receive beamforming architecture for smartphone-based wireless ultrasound imaging", PROGRESS IN BIOMEDICAL OPTICS AND IMAGING, SPIE - INTERNATIONAL SOCIETY FOR OPTICAL ENGINEERING, BELLINGHAM, WA, US, vol. 9790, 1 April 2016 (2016-04-01), pages 97901V-97901V, XP060065679, ISSN: 1605-7422, DOI: 10.1117/12.2216705 ISBN: 978-1-5106-0027-0	2,6-14, 17-23	Y	abstract; figures 1,2 pages 97901V-2, paragraph 1 - pages 97901V-4, paragraph 7 ----- -/-	1,3-5, 15,16
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INTERNATIONAL SEARCH REPORT

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