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(54) **ULTRASONIC DEVICE FOR DETECTING THE HEARTBEAT OF A PATIENT**

ULTRASCHALLVORRICHTUNG ZUR DETEKTION DES HERZSCHLAGS EINES PATIENTEN

DISPOSITIF ULTRASONORE DESTINÉ À DÉTECTER LE RYTHME CARDIAQUE D'UN PATIENT

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Description

[0001] The invention relates to an ultrasonic device for detecting the heartbeat of a patient that can preferably be used while the patient is being examined in an imaging apparatus, e.g., a magnetic resonance imaging, MRI, apparatus.

Background of the invention

[0002] When the heart of a patient is examined using an MRI or other imaging apparatus, it is important to eliminate motion artifacts caused by the rapid motion of the heart. To date, the image acquisition of the MRI is synchronized to the moment of the patient's heartbeat by means of an external trigger signal. It is customary in the art to derive such a trigger signal from an electrocardiogram that is acquired simultaneously with the MRI imaging. The high magnetic and electromagnetic fields involved in the MRI imaging superimpose a rather high level of interference onto the weak electrical signals that have been picked up from the skin of the patient. In addition, electrolytes in the blood of the patient represent moving charges that generate even more interference in combination with the high magnetic field. This interference is termed magnetohydrodynamic effect. Both types of interference deteriorate the detection of the heartbeat to the point where it can no longer serve as a reliable trigger signal for the MRI imaging. This causes the motion artifacts, which the triggering with the heartbeat was intended to avoid, to reappear in the resulting MRI images. In addition, when the heart of an unborn fetus is to be imaged using MRI, the skin of the fetus is not accessible for picking up an electrocardiogram, so no trigger signal for the MRI imaging is available at all.

[0003] To overcome these drawbacks (F. Kording, B. Schoennagel, G. Lund, F. Ueberle, C. Jung, G. Adam, J. Yamamura, "Doppler Ultrasound Compared with Electrocardiogram and Pulse Oximetry Cardiac Triggering: A Pilot Study", Magnetic Resonance in Medicine, doi: 10.1002/mm.25502 (2014)) propose, and demonstrate an MR-compatible cardiotocograph, CTG, to acquire the heartbeat of adult patients during MRI. The heart of the patient is examined with ultrasonic waves, and the reflected ultrasonic waves that are Doppler-shifted by the motion of the heart are detected. From this signal, the exact moment of the heartbeat is extracted using a wavelet-based peak detection. (J. Yamamura, I. Kopp, M. Frisch, R. Fischer, K. Valett, K. Hecher, G. Adam, U. Wedegärtner, "Cardiac MRI of the Fetal Heart Using a Novel Triggering Method: Initial Results in an Animal Model", Journal of Magnetic Resonance Imaging 35, 1071-1076 (2012)) propose, and demonstrate on sheep, MRI imaging of a fetal heart that is triggered by a CTG signal. The Doppler frequency shift is proportional to the speed of movement of the heart; therefore, the Doppler-shifted ultrasonic signal can serve as a suitable triggering signal for the image acquisition in the MRI.

[0004] US 2008/015439 A1 discloses a device comprising the features of the preamble of claim 1.

[0005] When the heart of a fetus is to be imaged using MRI, a high level of noise is superimposed on the CTG signal by the blood flow, breathing of the mother and by the MRI. As a robust trigger signal has to be processed in real-time without false negative or false positive trigger detections, there is therefore a need for an improved ultrasonic detection of the heartbeat during MRI.

Disclosure of the invention

[0006] The invention is defined in the appended claims. The inventors have developed a device for detecting the heartbeat of a patient. This device may preferably be used during imaging in a magnetic resonance imaging, MRI, apparatus because it can be very easily constructed to be MRI compatible. However, its basic working is not dependent on the presence of an MRI apparatus in any way. Specifically, it can be used in conjunction with another type of imaging apparatus, such as computer tomography, CT, or positron emission tomography, PET, to provide the heartbeat of the patient as a trigger signal for the imaging. The slower the imaging method is compared with the rhythm of the heartbeat, the more the image quality is improved by the triggering. However, the device can be used even without any imaging taking place.

[0007] The device comprises a transmitter for ultrasonic waves and a receiver for Doppler-shifted ultrasonic waves. The transmitter and the receiver may be implemented in one and the same transducer that performs both functions, but they may also be separate entities. Preferably, the transmitter and/or the receiver are piezoelectric transducers that may convert back and forth between an electrical signal and an ultrasonic signal.

[0008] The device further comprises evaluating means that are coupled to the receiver and configured to evaluate the moment t_1, t_2, t_3, t_4 of a heartbeat of the patient out of the time dependent signal from the receiver. This time dependent signal may be a raw signal, but it may also have been pre-processed in order to extract only a frequency component that has been Doppler-shifted from the frequency of the transmitter. This pre-processing may, for example, be accomplished using a demodulator.

[0009] The evaluating means comprise spectrum analyzing means that are configured to extract at least one time-varying frequency component from the time-dependent signal delivered by the receiver. The evaluating means further comprise determining means configured to evaluate the moment t_1, t_2, t_3, t_4 of the heartbeat from the variation of at least one of the frequency components in time.

[0010] Preferably, the device comprises means that prevent coupling to the radio frequency field of the MRI at RF frequency ω_R. Such means may, for example, comprise structures that are designed not to be resonant at the frequency ω_R, the use of non-metallic materials, or

standing wave traps configured to block the frequency ω_R.

[0011] The inventors have found that a single heartbeat of the patient is an aperiodic event that generates a plurality of frequency components in the time-dependent signal from the receiver. However, the heartbeat is a periodically recurring event. Therefore, any frequency component generated by the heartbeat is a periodic function in time, with the period corresponding to the heart rate of the patient. This can be exploited to derive the moment t_1, t_2, t_3, t_4 of every single heartbeat from said function. The inventors have found that, surprisingly, this permits a very accurate detection of this moment t_1, t_2, t_3, t4 even if the original time-dependent signal from the receiver is superimposed with noise that contains peaks about as high as, or even higher than, the peaks produced by the heartbeat. The only a priori knowledge used in this heartbeat detection is the fact that the heartbeat is a periodically recurring event.

[0012] Therefore, the variation of the frequency component in time that is evaluated by the determining means is most preferably a periodic variation that corresponds to the heart rate of the patient.

[0013] In a first very simple embodiment, not forming part of the present invention, the determining means may be configured to determine the moment t_a, t_b, t_c, t_d where the at least one time-dependent frequency component assumes a maximum m_a, m_b, m_c, m_d as the moment t_1, t_2, t_3, t_4 of the heartbeat. However, because the periodically varying frequency component more or less resembles a sine wave, its maximum may be a lot less sharp than a peak caused by the heartbeat in the original time-dependent signal from the receiver. To recover this sharpness and to improve the accuracy with which the moment t_1, t_2, t_3, t_4 can be detected, according to the invention, a hybrid detection of the heartbeat is used. The determining means are configured to determine a window of interest, WOI, in time from the variation of the at least one frequency component in time. This window is subsequently used to determine the moment where the time-dependent signal from the receiver assumes a maximum b_1, b_2, b_3, b_4 within the WOI as the moment t_1, t_2, t_3, t_4 of the heartbeat. Thus, basically, a frequency component of the time-dependent signal is analyzed to determine a time frame that contains the desired heartbeat event, and this time frame is used to gate the actual detection of this event in the original signal in the time domain.

[0014] The window of interest, WOI, may be a window of pre-set length Δt around the moment t_a, t_b, t_c, t_d where the at least one time-dependent frequency component of the signal assumes a maximum m_a, m_b, m_c, m_d. In the alternative or in combination with this, the WOI may also be, for example, a window that extends between two moments t_e, t_f, t_g, t_h where the at least one time-dependent frequency component (51a) assumes a minimum m_e, m_f, m_g, m_h.

[0015] For the detection of a fetal heart beat, multiple receivers may be necessary, entailing more electric wiring that is prone to pick up interference from the radio frequency field of the MRI. To reduce this interference and improve the compatibility of the device with MRI, resonant paths, especially circular closed conductive paths, should be avoided. Preferably, the response of the complete signal path from the receiver to the evaluating means at the frequency ω_R of the radio frequency field of the MRI is at least -30 dB lower than the response of a structure that is resonant at the frequency ω_R. To this end, the inventors propose a further specially advantageous embodiment of the invention that is directed at optimizing the electrical wiring in this way.

[0016] The inventors stress that while this embodiment integrates perfectly with the features relating to the evaluation of the signal presented above, it is not tied in any way to this mode of evaluation.

[0017] In this embodiment, the device for detecting the heartbeat of a patient comprises a transceiver unit with a transmitter for ultrasonic waves and a plurality of receivers for Doppler-shifted ultrasonic waves. The transmitters and the receivers may be one and the same element, for example, piezo-electric transducers that convert back and forth between an electrical voltage signal and an ultrasonic wave. The transceiver unit may therefore be a transducer unit. Each receiver has a first output terminal and a second output terminal. The transmitters and receivers may, for example, also be separate piezoelectric elements.

[0018] Each of the first output terminals of at least two receivers is connected by a branch line to a first common bus, and each of the second output terminals of the at least two receivers is connected by a branch line to a second common bus. The inventors have found that this topology of wiring minimizes the induction of radio frequency pulses to the common buses when the transceiver unit is used inside the radio frequency field of an MRI.

[0019] Likewise, multiple transmitters may be connected with common buses using the same topology. If transducers are used that combine the functions of a transmitter and a receiver in one single element, only one first common bus and one second common bus are necessary to wire all transducers. If the transducers are piezoelectric elements, each having a plus pole and a minus pole, then the first common bus may interconnect the plus poles, while the second common bus may interconnect the minus poles. The second common bus may also be connected to a ground potential.

[0020] The transmitter and the receivers may be mounted on a common dielectric substrate, preferably a flexible substrate. If the substrate is flexible, it may, for example, adapt to the shape of a mother's womb when placed on that womb. Exemplary mounting methods that are especially favorable for MRI compatibility are gluing and heat-sealing, which require no additional metal. For the device to be MRI-compatible, no ferromagnetic components may be introduced into the high magnetic field of the MRI, since it may heat up, and/or the magnetic

field may apply a high force to it. It is very advantageous to use as little metal in the field of the MRI as possible, since the rapidly varying magnetic field of the MRI may cause eddy currents in metal. The magnetic field that is associated with these eddy currents disturbs the field of the MRI and causes artifacts. The imaging quality of the MRI strongly depends on the homogeneity of the magnetic field and on the ability of radio frequency pulses to propagate throughout the patient without being obstructed by metallic matter.

[0021] In a specially advantageous embodiment of the invention, the dielectric substrate is at least partially surrounded by a conductive shield. This conductive shield prevents external electromagnetic fields from being picked up by the first and/or second common bus to which the receivers are connected, in the manner of a Faraday cage. This is especially advantageous in MRI imaging, where the patient is interrogated with radio-frequency fields. For compatibility with MRI imaging, the shield should be made of a non-ferromagnetic but highly conductive material, such as copper, silver or gold.

[0022] The shield may, for example, be a thin foil made of these metals. The shield may also, for example, comprise a non-conducting carrier, such as a plastic foil, carrying conductive structures.

[0023] The shield may surround the receivers in addition to surrounding the substrate. This improves the shielding of the first and second common buses at the price of introducing a slight mechanical damping of the ultrasonic waves captured by the receivers. If maximizing the coupling of the received ultrasonic waves to the receivers is more important, then the receivers may alternatively protrude out of the shield. In this case, the surface of the receivers that protrude out of the shield may comprise conductive structures that are electrically connected to the shield, so as to prevent electromagnetic waves from passing through the shield in the places where the receivers protrude out of the shield.

[0024] Preferably, at least part of the conductive shield is grounded. This grounding may be effected through the same cable that connects the first and second common buses to the outside world. For example, the cable may comprise at least two concentric conductors, wherein one conductor is connected to the first common bus and the other conductor is connected to the second common bus. At least a part of the conductive shield is grounded via the outermost conductor of the cable. For example, the cable may be a coaxial cable. The grounded outer conductor then provides the ground potential to the second common bus as a reference potential for the signal acquisition, while at the same time providing the ground to the shield.

[0025] The cable may also be, for example, a triaxial cable. In this case, for example, the first common bus may be connected to the innermost conductor, the second common bus may be connected to the middle conductor, and the shield may connect the outermost conductor to ground. At the price of having one more conductor, this provides the freedom of arbitrarily choosing the reference potential for the signal detection.

[0026] For MRI imaging, it is important that the shield provides substantially no circular structures that may heat up or lead to image artifacts. Therefore, in a further specially advantageous embodiment of the invention, at least part of the shield is divided into a plurality of segments. To make the different segments work together as one shield nonetheless, in a further specially advantageous embodiment of the invention, at least two of the segments are electrically connected by means of a capacitor.

[0027] This capacitor is preferably dimensioned in a way that it blocks lower frequencies which are relevant for the generation of eddy currents within the shield, while being conductive for higher frequencies that may be picked up by the first and second common buses. In this manner, eddy currents are avoided, but the shield is effective for the frequencies that may obscure the measured signals.

[0028] In a further specially advantageous embodiment of the invention, the first common bus and the second common bus are arranged on different faces of the substrate, and/or the first common bus and the second common bus are interwoven along at least part of the signal path between at least one receiver and the spectrum analyzing means. This reduces induced common-mode currents by the MRI and parasitic magnetic fields.

[0029] Preferably, the substrate (circuit board) is protected by a plastic casing that is biocompatible and sealed in a watertight manner.

[0030] When the heartbeat of a human fetus is to be detected, an additional challenge is that the fetus may move within the uterus so that its heart is no longer being examined by the ultrasonic waves from the transmitter. If this happens, the trigger signal for the MR imaging of the fetal heart is gone. Because the mother has to lie in a very uncomfortable position inside the MRI apparatus, which is not possible for an extended period of time, the loss of the trigger signal usually means that the whole examination of the fetal heart must be aborted and repeated at a later time. The intra-uterine movement of the fetus may especially be triggered by the loud intimidating noise generated by the MRI.

[0031] To overcome this drawback, in a further specially advantageous embodiment of the invention, the transceiver unit comprises a plurality of transmitters and/or receivers, and the evaluating means comprise tracking means to determine which transmitters and/or receivers cover the present position of the heart of the patient, so that the Doppler-shifted ultrasonic waves can be registered by at least one receiver and evaluated by the evaluating means. The transmitters may, for example, be arranged in groups, and they may be activated either by group or individually. The transmitters, and/or the receivers, may, for example, be distributed over a large area in which a fetus is expected to be moving. Preferably, the transmitters, and/or the receivers, are ar-

ranged on a flexible mat that can be placed on the mother's womb, so that every point in the uterus where the fetal heart can be is covered by at least one transmitter and/or receiver. Therefore, a constant detection of the heartbeat of the fetus, and thus a continuous trigger signal for the MRI, will be available, no matter which way the fetus moves.

[0032] Advantageously, the evaluating means comprise feedback means that control operation of the plurality of transmitters, wherein the feedback means are configured to power on only those transmitters that have been identified to cover the present position of the heart of the patient by the tracking means. No unnecessary ultrasound, which could result in artifacts, is then transmitted.

[0033] In a further advantageous embodiment of the invention, the signal path between at least one receiver and the spectrum analyzing means comprises at least one standing wave trap configured to block the radio frequency ω_R of the MRT apparatus, and/or to prevent coupling between the signal path and the radio frequency ω_R of the MRI. This avoids the introduction of common-mode signals into the wiring and avoids potential heating of the wire. Such heating could injure the patient by burning of the skin, or it might even be a fire hazard. Furthermore, the MRI image may be significantly disturbed, if the transmitted radio frequency ω_R of the MRI is induced into the signal path.

[0034] Preferably, the standing wave trap is formed from a coaxial cable with an inner conductor and an outer conducting shield. This may especially be the same coaxial cable that is carrying the signal. The coaxial cable is wound to a coil so that a first coil is formed in the inner conductor and a second coil is formed in the conducting shield. The second coil in the outer conducting shield is bridged by a capacitor that forms the standing wave trap in combination with said second coil in the outer conducting shield.

[0035] Moreover, if several piezo-electric elements are used, for example on a flexible substrate to create a large area for signal reception, multiple signals paths may be required. The signals paths may also couple to the radio frequency of the MRI, which has to be prevented in order to prevent heating of the signal paths and MRI image artifacts. This may especially be achieved by resonant parallel circuits, consisting of a parallel capacitor and inductor, tuned to the resonant frequency of the MRI, which are placed within each signal path.

[0036] In a specially advantageous embodiment of the invention, the signal path between at least one receiver and the spectrum analyzing means comprises a plurality of standing wave traps, so that in the radio frequency ω_R field of the MRI apparatus, the signal travels no more than a quarter of the wavelength λ corresponding to said radio frequency ω_R before traversing a first or a further standing wave trap. The magnetic field inside the MRI apparatus is very inhomogeneous and may result in localized heating of wiring.

[0037] If any currents coupled into the wiring by the radio frequency field cannot travel any further than $\lambda/4$ and if these are minimally attenuated by -30 dB, the heating and corresponding image artifacts can be effectively avoided.

[0038] Preferably, the signal path between at least one receiver and the spectrum analyzing means comprises a bandpass filter that rejects both the switching frequency ω_G of the gradient field of the MRI apparatus and the radio frequency ω_R of the MRI apparatus. The switching frequency ω_G is on the order of 200 kHz or below, while the frequency of the ultrasonic waves used to examine the heart of the patient is on the order of 1 MHz, and the radio frequency ω_R is on the order of 64 MHz and above. The bandpass may be preferably centered on the frequency of the ultrasonic waves. The width of the bandpass filter needs to be at least large enough to accommodate both the frequency of the transmitted ultrasonic wave and the frequency of the received ultrasonic wave with the maximum possible Doppler shift.

[0039] Advantageously, at least part of the evaluating means are shielded by a housing that comprises conductive carbon fiber. While such a housing serves as a Faraday cage like a metallic housing, the generation of eddy currents and associated parasitic magnetic fields is greatly reduced.

[0040] The disclosure presented above also teaches to provide a system that comprises a magnetic resonance tomography, MRI, apparatus and a device for detecting the heartbeat of a patient according to the present invention. The trigger input of the MRI apparatus that triggers image acquisition is coupled to the determining means of said device to receive the moment t_1, t_2, t_3, t_4 of the heartbeat of the patient, so that image acquisition is triggered at this moment t_1, t_2, t_3, t_4. The improved precision of the heartbeat detection and the improved resilience of this detection to interference of the MRI makes the triggering of the image acquisition more reliable. The reliablility of the triggering in turn improves the quality of the final MR image. In addition, the triggering is also resilient against intra-uterine movement of the fetus, so that this movement no longer results in the MRI examination being aborted.

Description of the drawings

[0041] Further advantageous embodiments of the invention are detailed in the description of the Figures, where this description shall not limit the scope of the invention. The Figures show:

Figure 1: Embodiment of the device 1 and of the system 100 according to the present invention.

Figure 2: Extraction of the moment t_1, t_2, t_3, t_4 of the heartbeat 2a out of the time-dependent signal 51 from the receiver 5.

Figure 3: Transceiver unit 13 optimized for compatibility with the MRI apparatus 3.

Figure 4: Transceiver unit 13, further optimized for tracking of intra-uterine movement of a fetus.

Figure 5: Assembly of a transceiver unit 13 with a flexible substrate 16 suitable for application on a mother's womb 2c.

Figure 6: A further embodiment of a transceiver unit 13 adapted for use on an adult patient 2.

Figure 7: Interfacing of the evaluating means 6 with the receiver 5 and the trigger input 31 of the MRI apparatus 3.

Figure 8: Shielding 20 of the substrate 16, connected to a coaxial cable 21.

Figure 9: Division of shield 20 into segments 20a-20d connected by capacitors 20e-20h.

[0042] Figure 1 shows an embodiment of the device 1 for detecting the heartbeat 2a of a patient 2 to synchronize the data acquisition of an MRI apparatus 3. The patient 2 is located within the MRI apparatus 3, where his heart 2b is to be imaged. To trigger this imaging, the device 1 is used. The device 1 comprises a transmitter 4 for ultrasonic waves 4a and a receiver 5 for Doppler-shifted ultrasonic waves 5a. The transmitter 4 and the receiver 5 are integrated in one single component 4, 5. The time-dependent signal 51 from the receiver 5 is fed into the evaluating means 6 by means of a coaxial cable 81 with an inner conductor 82 and an outer conductive shield 83. A stretch of this coaxial cable 81 is wound into a coil 81a, thereby forming a first coil 84 in the inner conductor 82 and a second coil 85 in the outer conducting shield 83. The second coil 85 in the outer conducting shield 83 is bridged by a capacitor 86, so that this second coil 85 and the capacitor 86 together form a standing wave trap 8 that is part of the device 1. The resonance frequency of the standing wave trap 8 is set to the radio frequency ω_R of the MRI apparatus 3.

[0043] The evaluating means 6 are shielded in a housing 65 that comprises conductive carbon fiber, so that the generation of eddy currents and associated magnetic fields is reduced. The time-dependent signal 51 contains interference from two sources in the MRI apparatus 3, namely from the switching of the gradient fields at the frequency ω_G and from the radio frequency pulses with frequency ω_R. To eliminate both sources of interference, the signal 51 is first filtered by means of an analog band-pass filter 61 that only lets a frequency band $\omega_U \pm \Delta\omega$ pass, wherein ω_U is the frequency of the transmitted ultrasonic waves 4a and $\Delta\omega$ is the Doppler shift that the received ultrasonic waves 5a have experienced due to the fast motion of the heart 2b of the patient 2. The signal 51 then passes through an analog front-end 62 that receives amplifies, demodulates and filters the signal 51 again, hence only signals with the frequency components of the Doppler shift $\Delta\omega$ remain in the signal 51a. The demodulation may also be performed in the digital domain.

[0044] The filtered signal 51a then passes through spectrum analyzing means 63 that decompose it into frequency components. Since a single heartbeat 2a of the patient 2 is an aperiodic event that is periodically recurring with a period corresponding to the heart rate of the patient 2, some frequency components 51b will be periodically time-varying with the same period. These frequency components are evaluated further by the determining means 64 to yield the moment t_1, t_2, t_3, t_4 of the heartbeat 2a of the patient 2. By contrast, frequency components in the signal 51a that result from noise or spontaneous movement are not periodic and can be neglected. The signal path from the receiver 5 to the input of the spectrum analyzing means 63 is labeled 52.

[0045] The filtered signal 51a, the time-varying frequency components 51b and the moment t_1, t_2, t_3, t_4 of the heartbeat are fed into tracking means 66 that monitor whether the heart 2b of the patient 2 is still in range of the transceiver 4, 5. If the heart 2b has left this range, so that the heartbeat 2a can no longer be detected, the feedback means 67 that are combined with the tracking means 66 power this transceiver 4, 5 off and power on a different transceiver 4, 5 that covers the new position of the heart 2b, so that the detection of the heartbeat 2a can be resumed.

[0046] The moment t_1, t_2, t_3, t_4 of the heartbeat 2a is delivered to a display unit 7 where it can be directly monitored by the attending physician. The main purpose of the evaluated moment t_1, t_2, t_3, t_4 of the heartbeat 2a, however, is to serve as a trigger signal for the imaging in the MRI apparatus 3. To this end, the moment t_1, t_2, t_3, t_4 of the heartbeat 2a is fed into a trigger port 31 of the MRI apparatus 3 by means of a standard cable connection or by means of a wireless connection. A wireless connection has the additional benefit that it cannot create any ground loops. For the same reason, the evaluating means 6 are battery-powered.

[0047] Figure 2 illustrates the extraction of the moment t_1, t_2, t_3, t_4 of the heartbeat 2a from the time-dependent filtered signal 51a in more detail. Diagram (a) shows the amplitude A of the filtered Doppler signal 51a over the time t. From this raw signal, the spectrum analyzing means 63 extract frequency components 51b with a time-varying amplitude A. To this end, the signal 51a is decomposed into frequency components at sampling time t, and it is monitored which of those frequency components vary in time. Diagram (b) shows the decomposition of the signal 51a into frequency components 51b at one point in time t_0. The decomposition can be done by discrete Fast Fourier Transform (FFT), filter banks or any other appropriate means.

[0048] Diagram (b) in Figure 2 shows only those frequency components 51b that are time-varying, with amplitudes $A(t_0)$ that correspond to the spectral power density of the respective frequency components 51b. It should be noted that frequency components caused by noise may well have amplitudes $A(t_0)$ that are at least as large, or even larger, than the amplitudes $A(t_0)$ of the time-varying frequency components 51b. However, the frequency components of the noise can be easily dis-

missed because they are not time-varying periodically and they are time-varying in the range of > 3 Hz, whereas the range of the heartbeat is < 2.5 Hz.

[0049] Diagram (c) in Figure 2 shows the variation in time of the amplitude A(t) of one time-varying frequency component 51b. Within the range of diagram (c), the amplitude A(t) assumes four maxima m_a, m_b, m_c, and m_d at times t_a, t_b, t_c and t_d, respectively. The amplitude A(t) also assumes four minima m_e, m_f, m_g and m_h at times t_e, t_f, t_g and t_h.

[0050] As shown in diagram (d) in Figure 2, windows of interest 53a, 53b, 53c and 53d in time are generated from the times t_a, t_b, t_c and t_d at which the amplitude A(t) assumes a maximum in diagram (c). Each window 53a, 53b, 53c and 53d has a fixed length Δt. Within each window 53a, 53b, 53c and 53d, the determining means 64 examine the original time-dependent signal 51a for the occurrence of the highest maximum. In each of the windows 53a, 53b, 53c and 53d, this maximum occurs at times t_1, t_2, t_3 and t_4 with amplitude A(t) values b_1, b_2, b_3 and b_4, respectively. It represents the E wave and is followed by a smaller maximum that represents the A wave. The times t_1, t_2, t_3 and t_4 are the final outcome of the determining means 64. The data analyzed in Figure 2 were acquired on an adult patient.

[0051] Figure 3 details a transceiver unit 13 that is optimized for compatibility with the MRI apparatus 3. This embodiment is adapted for detection of the heartbeat 2a of a fetus 2, but it has no compensation for intra-uterine movement of the fetus 2. The transceiver unit 13 comprises a substrate 16 with an upper face 16a and a lower face 16b. Sub-figure (a) shows the components visible on both faces 16a and 16b of the substrate 16. Sub-figure (b) only shows the components visible on the upper face 16a, with the exception of the transceivers 4, 5 that are mounted on the lower face 16b, but shown in sub-figure (b) with dashed contours for clarity. Sub-figure (c) only shows the components visible on the lower face 16b.

[0052] Each transceiver 4, 5, which combines the function of a transmitter 4 for ultrasonic waves 4a and the function of a receiver 5 for Doppler-shifted ultrasonic waves 5a, has a first terminal 9 and a second terminal 10. The first terminals 9 are connected by branch lines 14 to a first common bus 11. The second terminals 10 are connected by branch lines 15 to a second common bus 12. The first terminals 9 and the branch lines 14 are all on the upper face 16a of the substrate 16, while the second terminals 10 and the branch lines 15 are all on the lower face 16b of the substrate 16.

[0053] The first terminals 9 of the transceivers 4, 5 are positive (plus) terminals, while the second terminals 10 of the transceivers 4, 5 are negative (minus) terminals. The second common bus 12 that interconnects the second terminals 10 is connected via a standing wave trap 8 to the outside world.

[0054] As shown in a sectional view in sub-figure (d), and in sub-figure (e) in a perspective view, both common buses 11 and 12 straddle the substrate 16 and are interwoven with each other. This arrangement of the wiring contains as few symmetries and circular structures as possible, so that both the coupling of radio frequency pulses into the buses 11 and 12 and the generation of parasitic magnetic fields by the currents flowing through the buses 11 and 12 are minimized.

[0055] Figure 4 shows another embodiment of a transceiver unit 13 that is specifically adapted to compensate intra-uterine movement of a fetus 2 during an MRI examination. The substrate 16 is a large flexible mat designed to adapt to the shape of a mother's womb 2c. The transceivers 4, 5 are arranged in a plurality of groups 17a-17h, but may also be arranged individually. Even if transceivers are arranged in groups, their transmitting functions may be activated individually, and their receivers may be read out individually. In this grouped example, each group 17a-17h has a separate first common bus 11a-11h that connects the first terminals 9 of all transceivers 4, 5 in the group 17a-17h. All groups 17a-17h share one second common bus 12 that is connected to a ground potential. Every group 17a-17h has its own standing wave trap 8 through which all signals leaving this group 17a-17h on both buses 11a-11h, 12 have to pass. On its further path in the horizontal direction in the perspective of Figure 4, the signals have to pass further standing wave traps 8. The standing wave traps 8 are arranged so that no signal can travel more than one quarter of the wavelength corresponding to the radio frequency ω_R of the MRI apparatus 3 without hitting the first or another standing wave trap 8. This ensures that no matter where on the extended substrate 16 a radio frequency pulse couples into some wiring, this will not cause a potentially dangerous local heating of this wiring and will reduce MRI related image artifacts.

[0056] Figure 5 details the assembly of the embodiment shown in Figure 4, as well as its use. The flexible substrate 16 consists of a flexible plastic mat 16c and a flexible printed circuit board 16d onto which all wiring (for example, branch lines 14) is printed. The transceivers 4, 5 are set into both the flexible plastic mat 16c and the flexible printed circuit board 16d. The plastic mat 16c and the printed circuit board 16d are joined together in a water-tight manner by gluing.

[0057] To monitor the heartbeat 2a of a fetus 2 inside an uterus 2d, the transceiver unit 13 is placed on the mother's womb 2c with the flexible plastic mat 16c facing towards the womb 2c. When the fetus 2 moves inside the uterus 2d, the heart 2b of the fetus 2 may move out of the detecting range of one of the transceivers 4, 5 and into the detecting range of another one of the transceivers 4, 5, of which only three are shown in Figure 5 for clarity. The tracking means 66 register this intra-uterine movement and instruct the feedback means 67 to power on the correct transceivers 4, 5 in order to ensure an uninterrupted detection of the heartbeat 2a.

[0058] Figure 6 illustrates a further embodiment of a transceiver unit 13 that is adapted for use on an adult patient 2. Sub-figure (a) shows a sectional side view,

sub-figure (b) shows a top view.

[0059] Since an adult's heartbeat produces a much stronger signal, only one transceiver 4, 5 is required. The transceiver 4, 5 is set in a printed circuit board 16 inside a housing 19 and connected by an acoustic matching layer 18 to this housing 19. The first (positive, plus) terminal 9 of the transceiver 4, 5 is connected by a branch line 14 to the inner conductor 82 of the coaxial signal cable 81. The second (negative, minus) terminal 10 of the transceiver 4, 5 is connected by a branch line 15 to the outer conducting shield 83 of the coaxial signal cable 81. A standing wave trap 8 is installed between the second terminal 10 of the transceiver 4, 5 and the outer conducting shield 83.

[0060] Figure 7 details the interfacing of the evaluating means 6 with the receiver 5 and the trigger input 31 of the MRI apparatus 3. The signal 51 from the receiver 5 first passes through the analog bandpass filter 61 that only lets frequencies in the range $\omega_U \pm \Delta\omega$ pass. The analog front end 62 receives, amplifies, demodulates and filters the signal 51 to signal 51a. The demodulation may also be performed in the digital domain downstream of the analog front end 62, as long as it is performed before the decomposition of the signal 51a into frequency components. The analog front end 62 is the entry and exit point of a microcontroller 70 that further comprises a combined analog-digital and digital-analog converter 68, as well as a digital signal processor 69 that implements the function of the spectrum analyzing means 63 and the determining means 64.

[0061] Figure 8 shows a transceiver unit 13 according to an embodiment where the common dielectric substrate 16 onto which the transceivers 4, 5 are mounted is surrounded by a conductive shield 20. The transceiver unit 13 is connected to a coaxial cable 21 comprising an inner conductor 21a, a grounded outer conductor 21b that serves to shield the inner conductor 21a, and a plastic cladding 21c. The inner conductor 21a of the cable 21 is connected to the first common bus 11 on the substrate 16, while the outer conductor 21b of the cable 21 is connected to the shield 20. The shield 20 is in turn connected to the second common bus 12 on the substrate 16.

[0062] Two variants are shown in Figure 8. Figure 8a shows a first variant where the transceivers 4, 5 are completely covered by the shield 20. Figure 8b shows a second variant where the transceivers 4, 5 protrude through the shield 20, so that the shield 20 does not impede the coupling of transmitted and reflected ultrasonic waves in any way.

[0063] Figure 9 details the division of a shield 20 into four segments 20a-20d. The segments are connected to each other via capacitors 20e-20h. Lower frequencies that are mainly relevant for the generation of eddy currents are blocked by the capacitors 20e-20h, so that the shield 20 does not heat up. Higher frequencies that have the potential to interfere with the measurement pass the capacitors, and are therefore shorted by the shield 20.

List of reference signs

[0064]

1	device
2	patient (adult or fetus)
2a	heartbeat of patient 2
2b	heart of patient 2
2c	mother's womb
2d	uterus
3	MRI apparatus
4	ultrasonic transmitter
4a	ultrasound transmitted by transmitter 4
5	ultrasonic receiver
5a	ultrasound received by receiver 5
6	evaluating means
7	display unit
8	standing wave trap
9	first terminal
10	second terminal
11	first common bus
12	second common bus
13	transceiver unit
14	branch lines to first common bus 11
15	branch lines to second common bus 12
16	substrate
16a, 16b	top and bottom faces of substrate 16
16c	flexible plastic mat of substrate 16
16d	flexibe printed circuit board of substrate 16
17a-17h	groups of transceivers 4, 5
18	acoustic matching layer
19	housing
20	conductive shield
20a-20d	segments of conductive shield 20
20e-20h	capacitors between segments 20a-20d
21	cable to connect to common buses 11, 11a-h, and 12
21a, 21b	concentric conductors of cable 21
21c	plastic cladding of cable 21
31	trigger input of MRI apparatus 3
51	time-dependent signal from receiver 5
51a	time-varying filtered Doppler signal derived from signal 51
51b	frequency components of signal 51a
52	signal path from receiver 5 to spectrum analyzing means 63
53a-53d	window of interest, WOI
61	analog bandpass filter
62	analog front end
63	spectrum analyzing means
64	determining means
65	housing of evaluating means 6
66	tracking means

67	feedback means
68	analog-digital converter
69	digital signal processor
70	microcontroller
81	coaxial cable
81a	coil formed in coaxial cable 81
82	inner conductor of coaxial cable 81
83	outer conducting shield of coaxial cable 81
84	first coil formed in inner conductor 81
85	second coil formed in outer conducting shield 83
86	capacitor
100	system
A	amplitude (time domain) / spectral power density (ω domain)
b_1, b_2, b_3, b_4	amplitudes of E wave maxima in signal 51
m_a, m_b, m_c, m_d	amplitudes of maxima in frequency component 51a
m_e, m_f, m_g, m_h	amplitudes of minima in frequency component 51a
t	time
Δt	length of windows 53a-53d
t_a, t_b, t_c, t_d	moments of maxima in frequency component 51a
t_e, t_f, t_g, t_h	moments of minima in frequency component 51a
to	point in time for generation of diagram (b) in Figure 2
t_1, t_2, t_3, t_4	moment of heartbeat 2a of patient 2
ω_G	switching frequency of gradient field in MRI apparatus 3
ω_R	radio frequency field in MRI apparatus 3
ω_U	frequency of ultrasound 4a
$\Delta\omega$	maximum Doppler shift of ultrasound 5a

Claims

1. A device (1) for detecting the heartbeat (2a) of a patient (2), preferably during imaging in a magnetic resonance imaging, MRI, apparatus (3) or other imaging apparatus, comprising a transmitter (4) for ultrasonic waves (4a), a receiver (5) for Doppler-shifted ultrasonic waves (5a), and evaluating means (6) coupled to the receiver (5) and configured to evaluate the moment t_1, t_2, t_3, t_4 of a heartbeat (2a) of the patient (2) out of the time-dependent signal (51) from the receiver (5), wherein said evaluating means (6) comprise:

   • spectrum analyzing means (63) configured to extract at least one time-varying frequency component (51b) from the time-dependent signal (51, 51a) delivered by the receiver (5); and

   • determining means (64) configured to evaluate the moment t_1, t_2, t_3, t_4 of the heartbeat (2a) from the variation of at least one of the frequency components (51b) in time,

   characterized in that
   the determining means (64) are configured to determine a window of interest, WOI, (53a, 53b, 53c, 53d) in time from the variation of the at least one frequency component (51b) in time, and to determine the moment where the time-dependent signal (51, 51a) from the receiver (5) assumes a maximum b_1, b_2, b_3, b_4 within the WOI (53a, 53b, 53c, 53d) as the moment t_1, t_2, t_3, t_4 of the heartbeat (2a).

2. The device (1) according to claim 1, **characterized in that** the determining means (64) are configured to determine the WOI (53a, 53b, 53c, 53d) as a window of pre-set length Δt around the moment t_a, t_b, t_c, t_d where the at least one time-dependent frequency component (51a) assumes a maximum m_a, m_b, m_c, m_d, and/or as a window that extends between two moments t_e, t_f, t_g, t_h where the at least one time-dependent frequency component (51b) assumes a minimum m_e, m_f, m_g, m_h.

3. The device (1) according to any one of claims 1 to 2, **characterized in that** the device (1) further comprises a transceiver unit (13) with at least one transmitter (4) and a plurality of receivers (5), wherein

   • each receiver (5) has a first output terminal (9) and a second output terminal (10),
   • each of the first output terminals (9) of at least two receivers (5) is connected by a branch line (14) to a first common bus (11, 11a-11h), and
   • each of the second output terminals (10) of the at least two receivers (5) is connected by a branch line (15) to a second common bus (12).

4. The device (1) according to claim 3, **characterized in that** the transmitter (4) and the receivers (5) are mounted on a common dielectric substrate (16), preferably a flexible substrate (16).

5. The device (1) according to claim 4, **characterized in that** the dielectric substrate (16) is at least partially surrounded by a conductive shield (20).

6. The device (1) according to claim 5, **characterized in that** at least part of the conductive shield (20) is grounded.

7. The device (1) according to claim 6, **characterized in that** the first common bus (11, 11a-11h) and the second common bus (12) are connected to two different concentric conductors (21a, 21b) of a cable (21), and at least part of the conductive shield (20)

is grounded via the outermost conductor (21b) of the cable (21).

8. The device (1) according to any one of claims 5 to 7, **characterized in that** at least part of the shield (20) is divided into a plurality of segments (20a-20d).

9. The device (1) according to claim 8, **characterized in that** at least two of the segments (20a-20d) are electrically connected by means of a capacitor (20e-20h).

10. The device (1) according to any one of claims 4 to 9, **characterized in that** the first common bus (11, 11a-11h) and the second common bus (12) are arranged on different faces (16a, 16b) of the substrate (16), and/or that the first common bus (11, 11a-11h) and the second common bus (12) are interwoven along at least part of the signal path (52) between at least one receiver (5) and the spectrum analyzing means (63).

11. The device (1) according to any one of claims 3 to 10, **characterized in that** the transceiver unit (13) comprises a plurality of transmitters (4) and that the evaluating means (6) comprise tracking means (66) to determine which transmitters (4) and/or receivers (5) cover the present position of the heart (2b) of the patient (2), so that the Doppler-shifted ultrasonic waves (5a) can be registered by at least one receiver (5) and evaluated by the evaluating means (6).

12. The device (1) according to claim 11, **characterized in that** the evaluating means (6) comprise feedback means (67) that control operation of the plurality of transmitters (4), wherein the feedback means (67) are configured to power on only those transmitters (4) that have been identified to cover the present position of the heart (2b) of the patient (2) by the tracking means (66).

13. The device (1) according to any one of claims 1 to 12, **characterized in that** the signal path (52) between at least one receiver (5) and the spectrum analyzing means (63) comprises at least one standing wave trap (8) configured to block the radio frequency ω_R of the MRI apparatus (3).

14. The device (1) according to claim 13, **characterized in that** the standing wave trap (8) is formed from a coaxial cable (81) with an inner conductor (82) and an outer conducting shield (83), wherein the coaxial cable (81) is wound to a coil (81a) so that a first coil (84) is formed in the inner conductor (82) and a second coil (85) is formed in the conducting shield (83) and wherein the second coil (85) in the outer conducting shield (83) is bridged by a capacitor (86) that forms the standing wave trap (8) in combination with said second coil (85) in the outer conducting shield (83).

15. The device (1) according to any one of claims 13 to 14, **characterized in that** said signal path (52) comprises a plurality of standing wave traps (8), so that in the radio frequency ω_R field of the MRI apparatus (3), the signal (51) travels no more than a quarter of the wavelength λ corresponding to said radio frequency ω_R before traversing a first or a further standing wave trap (8).

16. The device (1) according to any one of claims 1 to 15, **characterized in that** the signal path (52) between at least one receiver (5) and the spectrum analyzing means (63) comprises a bandpass filter (61) that rejects both the switching frequency ω_G of the gradient field of the MRI apparatus (3) and the radio frequency ω_R of the MRI apparatus (3).

17. The device (1) according to any one of claims 1 to 16, **characterized in that** at least part of the evaluating means (6) are shielded by a housing (65) that comprises conductive carbon fiber.

18. A system (100), comprising a magnetic resonance tomography, MRI, apparatus (3) and a device (1) according to any one of claims 1 to 17, wherein the trigger input (31) of the MRI apparatus (3) that triggers image acquisition is coupled to the determining means (64) of the device (1) to receive the moment t_1, t_2, t_3, t_4 of the heartbeat (2a) of the patient (2), so that image acquisition is triggered at said moment t_1, t_2, t_3, t_4.

Patentansprüche

1. Eine Vorrichtung (1) zur Erkennung des Herzschlags (2a) eines Patienten (2), vorzugsweise während der Abbildung in einem Magnetresonanzabbildungsgerät (3), MRI, oder einem anderen Abbildungsgerät, umfassend einen Sender (4) für Ultraschallwellen (4a), einen Empfänger (5) für Doppler-verschobene Ultraschallwellen (5a), und Auswertungsmittel (6), die an den Empfänger (5) gekoppelt sind und dazu ausgebildet sind, den Moment t_1, t_2, t_3, t_4 eines Herzschlags (2a) des Patienten aus dem zeitabhängigen Signal (51) vom Empfänger (5) auszuwerten, wobei besagte Auswertungsmittel (6) umfassen:

- Spektrumanalysiermittel (63), dazu ausgebildet, mindestens eine zeitveränderliche Frequenzkomponente (51b) aus dem vom Empfänger (5) gelieferten zeitabhängigen Signal (51, 51a) zu extrahieren; und
- Bestimmungsmittel (64), dazu ausgebildet, den Moment t_1, t_2, t_3, t_4 des Herzschlags (2a)

aus der Änderung der mindestens einen Frequenzkomponente (51b) mit der Zeit auszuwerten,

dadurch gekennzeichnet, dass
die Bestimmungsmittel (64) dazu ausgebildet sind, ein Interessenfenster, WOI, (53a, 53b, 53c, 53d) in der Zeit aus der Änderung der mindestens einen Frequenzkomponente (51b) mit der Zeit zu bestimmen, und den Moment, wo das zeitabhängige Signal (51, 51a) vom Empfänger (5) innerhalb des WOI (53a, 53b, 53c, 53d) ein Maximum b_1, b_2, b_3, b_4 annimmt, als den Moment t_1, t_2, t_3, t_4 des Herzschlags (2a) zu bestimmen.

2. Die Vorrichtung (1) nach Anspruch 1, **dadurch gekennzeichnet, dass** die Bestimmungsmittel (64) dazu ausgebildet sind, das WOI (53a, 53b, 53c, 53d) als ein Fenster von vorgegebener Länge Δt um den Moment t_a, t_b, t_c, t_d, wo die mindestens eine zeitabhängige Frequenzkomponente (51a) ein Maximum m_a, m_b, m_c, m_d annimmt, und/oder als Fenster, das sich zwischen zwei Momenten t_e, t_f, t_g, t_h erstreckt, wo die mindestens eine zeitabhängige Frequenzkomponente (51b) ein Minimum m_e, m_f, m_g, m_h annimmt, zu bestimmen.

3. Die Vorrichtung (1) nach einem der Ansprüche 1 bis 2, **dadurch gekennzeichnet, dass** die Vorrichtung (1) weiterhin eine Transceiver-Einheit (13) mit mindestens einem Sender (4) und einer Mehrzahl von Empfängern (5) umfasst, wobei

- jeder Empfänger (5) einen ersten Ausgabeanschluss (9) und einen zweiten Ausgabeanschluss (10) hat,
- jeder der ersten Ausgabeanschlüsse (9) von mindestens zwei Empfängern (5) durch eine Abzweigleitung (14) mit einem ersten gemeinsamen Bus (11, 11a-11h) verbunden ist und
- jeder der zweiten Ausgabeanschlüsse (10) von mindestens zwei Empfängern (5) durch eine Abzweigleitung (15) mit einem zweiten gemeinsamen Bus (12) verbunden ist.

4. Die Vorrichtung (1) nach Anspruch 3, **dadurch gekennzeichnet, dass** der Sender (4) und die Empfänger (5) auf einem gemeinsamen dielektrischen Substrat (16), vorzugsweise einem flexiblen Substrat (16), montiert sind.

5. Die Vorrichtung (1) nach Anspruch 4, **dadurch gekennzeichnet, dass** das dielektrische Substrat (16) zumindest teilweise von einer leitenden Abschirmung (20) umgeben ist.

6. Die Vorrichtung (1) nach Anspruch 5, **dadurch gekennzeichnet, dass** mindestens ein Teil der leitenden Abschirmung (20) geerdet ist.

7. Die Vorrichtung (1) nach Anspruch 6, **dadurch gekennzeichnet, dass** der erste gemeinsame Bus (11, 11a-11h) und der zweite gemeinsame Bus (12) mit zwei verschiedenen konzentrischen Leitern (21a, 21b) eines Kabels (21) verbunden sind und mindestens ein Teil der leitenden Abschirmung (20) über den äußersten Leiter (21b) des Kabels (21) geerdet ist.

8. Die Vorrichtung (1) nach einem der Ansprüche 5 bis 7, **dadurch gekennzeichnet, dass** mindestens ein Teil der Abschirmung (20) in eine Mehrzahl von Segmenten (20a-20d) unterteilt ist.

9. Die Vorrichtung (1) nach Anspruch 8, **dadurch gekennzeichnet, dass** mindestens zwei der Segmente (20a-20d) mittels eines Kondensators (20e-20h) elektrisch verbunden sind.

10. Die Vorrichtung (1) nach einem der Ansprüche 4 bis 9, **dadurch gekennzeichnet, dass** der erste gemeinsame Bus (11, 11a-11h) und der zweite gemeinsame Bus (12) auf verschiedenen Seiten (16a, 16b) des Substrats (16) angeordnet sind, und/oder dass der erste gemeinsame Bus (11, 11a-11h) und der zweite gemeinsame Bus (12) entlang mindestens eines Teils des Signalpfads (52) zwischen mindestens einem Empfänger (5) und den Spektrumanalysiermitteln (63) miteinander verwoben sind.

11. Die Vorrichtung nach einem der Ansprüche 3 bis 10, **dadurch gekennzeichnet, dass** die Transceivereinheit (13) eine Mehrzahl von Sendern (4) umfasst und dass die Auswertungsmittel (6) Verfolgungsmittel (66) umfassen, um zu bestimmen, welche Sender (4) und/oder Empfänger (5) die gegenwärtige Position des Herzens (2b) des Patienten (2) abdecken, so dass die Doppler-verschobenen Ultraschallwellen (5a) von mindestens einem Empfänger (5) registriert und durch die Auswertungsmittel (6) ausgewertet werden können.

12. Die Vorrichtung nach Anspruch 11, **dadurch gekennzeichnet, dass** die Auswertungsmittel (6) Rückkopplungsmittel (67) umfassen, die den Betrieb der Mehrzahl von Sendern (4) steuern, wobei die Rückkopplungsmittel (67) dazu ausgebildet sind, nur diejenigen Sender (4) einzuschalten, die durch die Verfolgungsmittel (66) identifiziert worden sind, die gegenwärtige Position des Herzens (2a) des Patienten (2) abzudecken.

13. Die Vorrichtung (1) nach einem der Ansprüche 1 bis 12, **dadurch gekennzeichnet, dass** der Signalpfad (52) zwischen mindestens einem Empfänger (5) und den Spektrumanalysiermitteln (63) mindestens eine

Stehwellenfalle (8) umfasst, die dazu ausgebildet ist, die Radiofrequenz ω_R des MRI-Geräts (3) zu blockieren.

14. Die Vorrichtung (1) nach Anspruch 13, **dadurch gekennzeichnet, dass** die Stehwellenfalle (8) aus einem Koaxialkabel (81) mit einem inneren Leiter (82) und einer äußeren leitenden Abschirmung (83) gebildet ist, wobei das Koaxialkabel (81) zu einer Spule (81a) gewunden ist, so dass eine erste Spule (4) im inneren Leiter (82) gebildet wird und eine zweite Spule (85) in der leitenden Abschirmung (83) gebildet wird, und wobei die zweite Spule (85) in der äußeren leitenden Abschirmung (83) mit einem Kondensator (86) überbrückt ist, der in Kombination mit der besagten zweiten Spule (85) in der äußeren leitenden Abschirmung (3) die Stehwellenfalle (8) bildet.

15. Die Vorrichtung nach einem der Ansprüche 13 bis 14, **dadurch gekennzeichnet, dass** besagter Signalpfad (52) eine Mehrzahl von Stehwellenfallen (8) umfasst, so dass das Signal im Feld des MRI-Geräts (3) mit der Radiofrequenz ω_R nicht mehr als ein Viertel der zu der Radiofrequenz ω_R korrespondierenden Wellenlänge λ zurücklegt, bevor es eine erste oder weitere Stehwellenfalle (8) überquert.

16. Die Vorrichtung (1) nach einem der Ansprüche 1 bis 15, **dadurch gekennzeichnet, dass** der Signalpfad (52) zwischen mindestens einem Empfänger (5) und den Spektrumanalysiermitteln (63) einen Bandpassfilter (61) umfasst, der sowohl die Schaltfrequenz ω_G des Gradientenfelds des MRI-Geräts (3) als auch die Radiofrequenz ω_R des MRI-Geräts (3) zurückweist.

17. Die Vorrichtung (1) nach einem der Ansprüche 1 bis 16, **dadurch gekennzeichnet, dass** mindestens ein Teil der Auswertungsmittel (6) durch ein Gehäuse, das leitende Kohlefaser umfasst, abgeschirmt ist.

18. Ein System (100), umfassend ein Magnetresonanztomographie-Gerät (3), MRI, und eine Vorrichtung (1) nach einem der Ansprüche 1 bis 17, wobei der Triggereingang (31) des MRI-Geräts (3), der die Bildaufnahme auslöst, an die Bestimmungsmittel (64) der Vorrichtung (1) gekoppelt ist, um den Moment t_1, t_2, t_3, t_4 des Herzschlags (2a) des Patienten (2) zu empfangen, so dass die Bildaufnahme zu diesem Moment t_1, t_2, t_3, t_4 ausgelöst wird.

Revendications

1. Dispositif (1) pour détecter la pulsation cardiaque (2a) d'un patient (2), de préférence pendant une imagerie dans un appareil d'imagerie par résonance magnétique, IRM, (3) ou autre appareil d'imagerie, comprenant un émetteur (4) d'ondes ultrasonores (4a), un récepteur (5) d'ondes ultrasonores à décalage par effet Doppler (5a), et des moyens d'évaluation (6) couplés au récepteur (5) et configurés pour évaluer le moment t_1, t_2, t_3, t_4 d'une pulsation cardiaque (2a) du patient (2) à partir du signal dépendant du temps (51) provenant du récepteur (5), dans lequel lesdits moyens d'évaluation (6) comprennent :

- des moyens d'analyse de spectre (63) configurés pour extraire au moins une composante de fréquence variant dans le temps (51b) du signal dépendant du temps (51, 51a) délivré par le récepteur (5) ; et
- des moyens de détermination (64) configurés pour évaluer le moment t_1, t_2, t_3, t_4 de la pulsation cardiaque (2a) à partir de la variation d'au moins l'une des composantes de fréquence (51b) dans le temps,

caractérisé en ce que
les moyens de détermination (64) sont configurés pour déterminer une fenêtre d'intérêt, WOI, (53a, 53b, 53c, 53d) dans le temps à partir de la variation de l'au moins une composante de fréquence (51b) dans le temps, et pour déterminer le moment où le signal dépendant du temps (51, 51a) provenant du récepteur (5) adopte un maximum b_1, b_2, b_3, b_4, au sein de la WOI (53a, 53b, 53c, 53d) en tant que moment t_1, t_2, t_3, t_4 de la pulsation cardiaque (2a).

2. Dispositif (1) selon la revendication 1, **caractérisé en ce que** les moyens de détermination (64) sont configurés pour déterminer la WOI (53a, 53b, 53c, 53d) en tant que fenêtre de longueur préétablie Δt autour du moment t_a, t_b, t_c, t_d où l'au moins une composante de fréquence dépendant du temps (51a) adopte un maximum m_a, m_b, m_c, m_d, et/ou en tant que fenêtre qui s'étend entre deux moments t_e, t_f, t_g, t_h où l'au moins une composante de fréquence dépendant du temps (51b) adopte un minimum m_e, m_f, m_g, m_h.

3. Dispositif (1) selon l'une quelconque des revendications 1 et 2, **caractérisé en ce que** le dispositif (1) comprend en outre une unité émetteur-récepteur (13) avec au moins un émetteur (4) et une pluralité de récepteurs (5), dans lequel chaque récepteur (5) comporte une première borne de sortie (9) et une seconde borne de sortie (10),
chacune des premières bornes de sortie (9) d'au moins deux récepteurs (5) est connectée par une ligne de dérivation (14) à un premier bus commun (11, 11a à 11h), et chacune des secondes bornes de sortie (10) des au moins deux récepteurs (5) est connectée par une ligne de dérivation (15) à un second bus commun (12).

4. Dispositif (1) selon la revendication 3, **caractérisé en ce que** l'émetteur (4) et les récepteurs (5) sont montés sur un substrat diélectrique commun (16), de préférence un substrat flexible (16).

5. Dispositif (1) selon la revendication 4, **caractérisé en ce que** le substrat diélectrique (16) est entouré au moins partiellement d'un blindage conducteur (20).

6. Dispositif (1) selon la revendication 5, **caractérisé en ce qu'**au moins une partie du blindage conducteur (20) est mise à la terre.

7. Dispositif (1) selon la revendication 6, **caractérisé en ce que** le premier bus commun (11, 11a à 11h) et le second bus commun (12) sont connectés à deux conducteurs concentriques (21a, 21b) différents d'un câble (21), et au moins une partie du blindage conducteur (20) est mise à la terre via le conducteur le plus externe (21b) du câble (21).

8. Dispositif (1) selon l'une quelconque des revendications 5 à 7, **caractérisé en ce qu'**au moins une partie du blindage (20) est divisée en une pluralité de segments (20a à 20d).

9. Dispositif (1) selon la revendication 8, **caractérisé en ce qu'**au moins deux des segments (20a à 20d) sont connectés électriquement au moyen d'un condensateur (20e à 20h).

10. Dispositif (1) selon l'une quelconque des revendications 4 à 9, **caractérisé en ce que**
le premier bus commun (11, 11a à 11h) et le second bus commun (12) sont agencés sur des faces (16a, 16b) différentes du substrat (16), et/ou **en ce que** le premier bus commun (11, 11a à 11h) et le second bus commun (12) sont entrelacés le long d'au moins une partie du trajet de signal (52) entre au moins un récepteur (5) et les moyens d'analyse de spectre (63).

11. Dispositif (1) selon l'une quelconque des revendications 3 à 10, **caractérisé en ce que**
l'unité émetteur-récepteur (13) comprend une pluralité d'émetteurs (4) et **en ce que** les moyens d'évaluation (6) comprennent des moyens de suivi (66) pour déterminer quels émetteurs (4) et/ou récepteurs (5) couvrent la position actuelle du cœur (2b) du patient (2), de sorte que les ondes ultrasonores à décalage par effet Doppler (5a) puissent être enregistrées par au moins un récepteur (5) et évaluées par les moyens d'évaluation (6).

12. Dispositif (1) selon la revendication 11, **caractérisé en ce que** les moyens d'évaluation (6) comprennent des moyens de rétroaction (67) qui commandent un fonctionnement de la pluralité d'émetteurs (4), dans lequel les moyens de rétroaction (67) sont configurés pour alimenter uniquement les émetteurs (4) qui ont été identifiés comme couvrant la position actuelle du cœur (2b) du patient (2) par les moyens de suivi (66).

13. Dispositif (1) selon l'une quelconque des revendications 1 à 12, **caractérisé en ce que**
le trajet de signal (52) entre au moins un récepteur (5) et les moyens d'analyse de spectre (63) comprend au moins un piège à ondes stationnaires (8) configuré pour bloquer la fréquence radio ϖ_R de l'appareil IRM (3).

14. Dispositif (1) selon la revendication 13, **caractérisé en ce que** le piège à ondes stationnaires (8) est formé d'un câble coaxial (81) avec un conducteur interne (82) et un blindage conducteur externe (83), dans lequel le câble coaxial (81) est enroulé sur une bobine (81a) de sorte qu'une première bobine (84) est formée dans le conducteur interne (82) et une seconde bobine (85) est formée dans le blindage conducteur (83) et dans lequel la seconde bobine (85) dans le blindage conducteur externe (83) est pontée par un condensateur (86) qui forme le piège à ondes stationnaires (8) en combinaison avec ladite seconde bobine (85) dans le blindage conducteur externe (83).

15. Dispositif (1) selon l'une quelconque des revendications 13 et 14, **caractérisé en ce que**
ledit trajet de signal (52) comprend une pluralité de pièges à ondes stationnaires (8), de sorte que dans le champ de radiofréquence ω_R de l'appareil IRM (3), le signal (51) circule sur pas plus d'un quart de la longueur d'onde λ correspondant à ladite fréquence radio ϖ_R avant de traverser un premier ou un autre piège à ondes stationnaires (8).

16. Dispositif (1) selon l'une quelconque des revendications 1 à 15, **caractérisé en ce que**
le trajet de signal (52) entre au moins un récepteur (5) et les moyens d'analyse de spectre (63) comprend un filtre passe-bande (61) qui rejette à la fois la fréquence de commutation ϖ_G du champ de gradient de l'appareil IRM (3) et la fréquence radio ϖ_R de l'appareil IRM (3).

17. Dispositif (1) selon l'une quelconque des revendications 1 à 16, **caractérisé en ce que**
au moins une partie des moyens d'évaluation (6) est blindée par un boîtier (65) qui comprend une fibre en carbone conductrice.

18. Système (100), comprenant un appareil de tomographie par résonance magnétique, IRM, (3) et un dispositif (1) selon l'une quelconque des revendications

1 à 17, dans lequel l'entrée de déclenchement (31) de l'appareil IRM (3) qui déclenche une acquisition d'image est couplée aux moyens de détermination (64) du dispositif (1) pour recevoir le moment t_1, t_2, t_3, t_4 de la pulsation cardiaque (2a) du patient (2), de sorte qu'une acquisition d'image est déclenchée audit moment t_1, t_2, t_3, t_4.

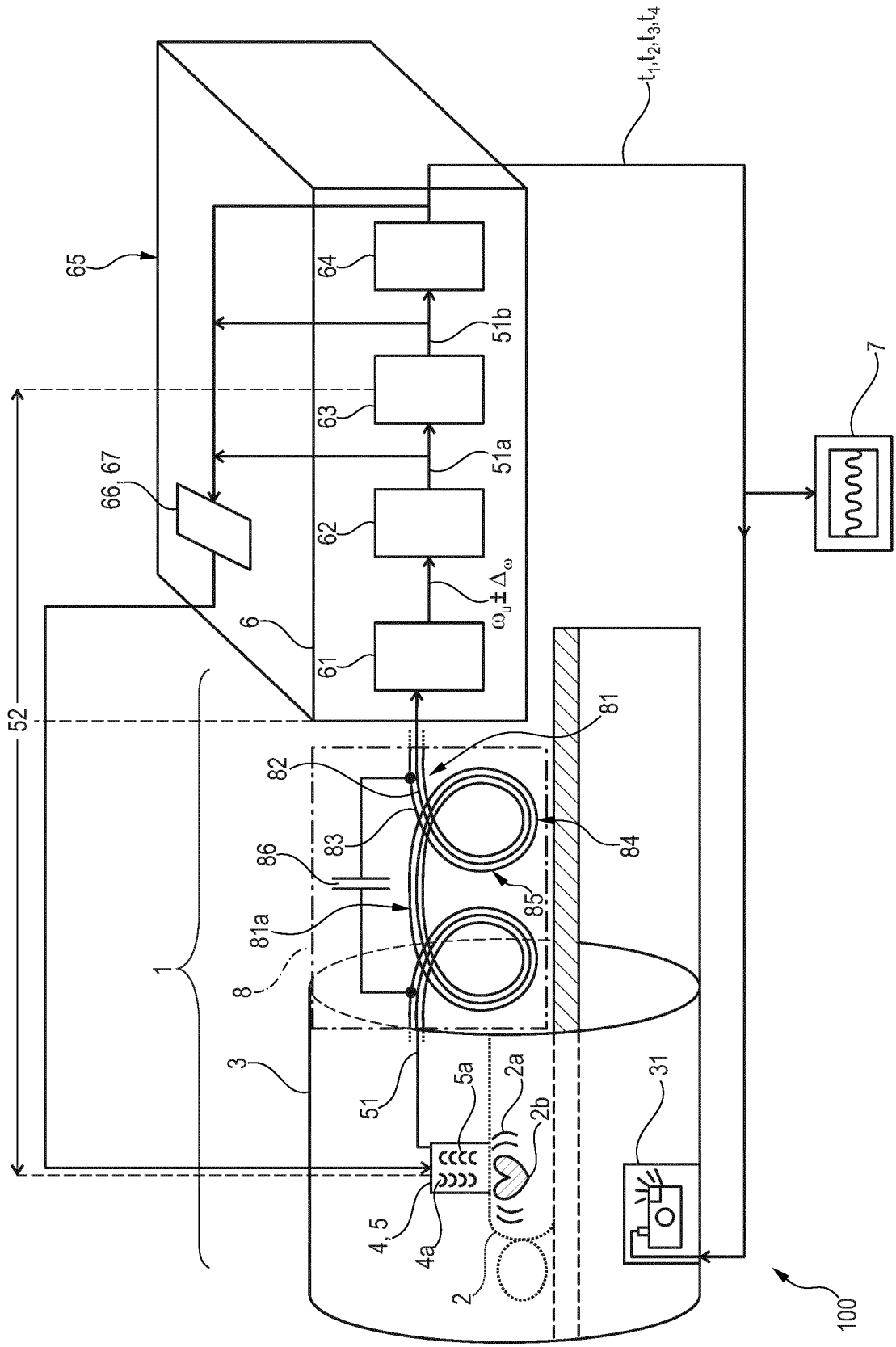
t_1,t_2,t_3,t_4
65
64
51b
63
51a
66, 67
62
$\omega_u \pm \Delta_\omega$
6
61
7
52
81
82
83
84
86
85
81a
8
1
3
51
5a
2a
2b
31
4, 5
4a
2
100

Fig. 1

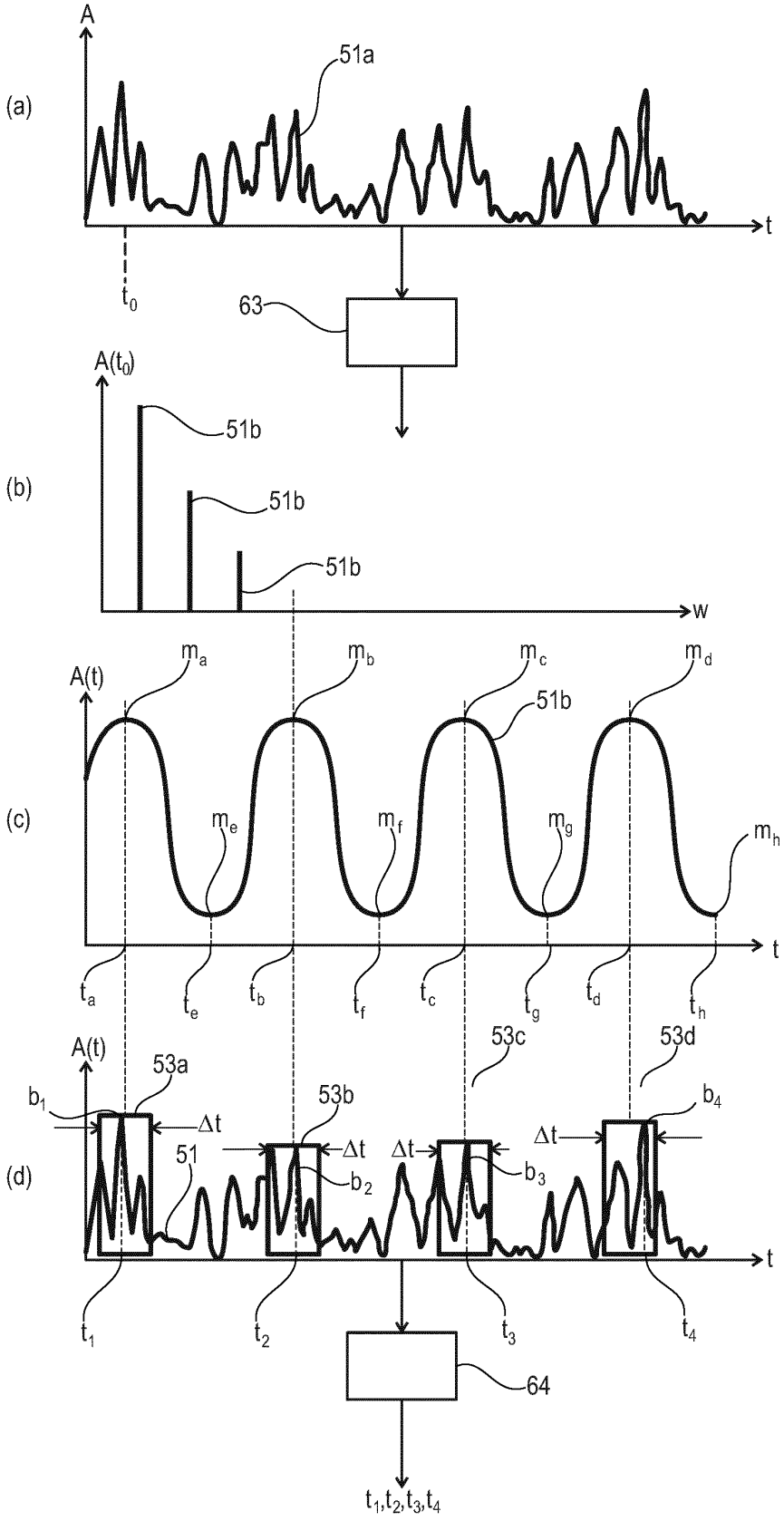
(a)
A
51a
t
t_0
63
(b)
$A(t_0)$
51b
51b
51b
w
(c)
A(t)
m_a
m_b
m_c
m_d
51b
m_e
m_f
m_g
m_h
t
t_a
t_e
t_b
t_f
t_c
t_g
t_d
t_h
(d)
A(t)
53a
53b
53c
53d
b_1
b_2
b_3
b_4
Δt
Δt
Δt
Δt
51
t
t_1
t_2
t_3
t_4
64
t_1,t_2,t_3,t_4

Fig. 2

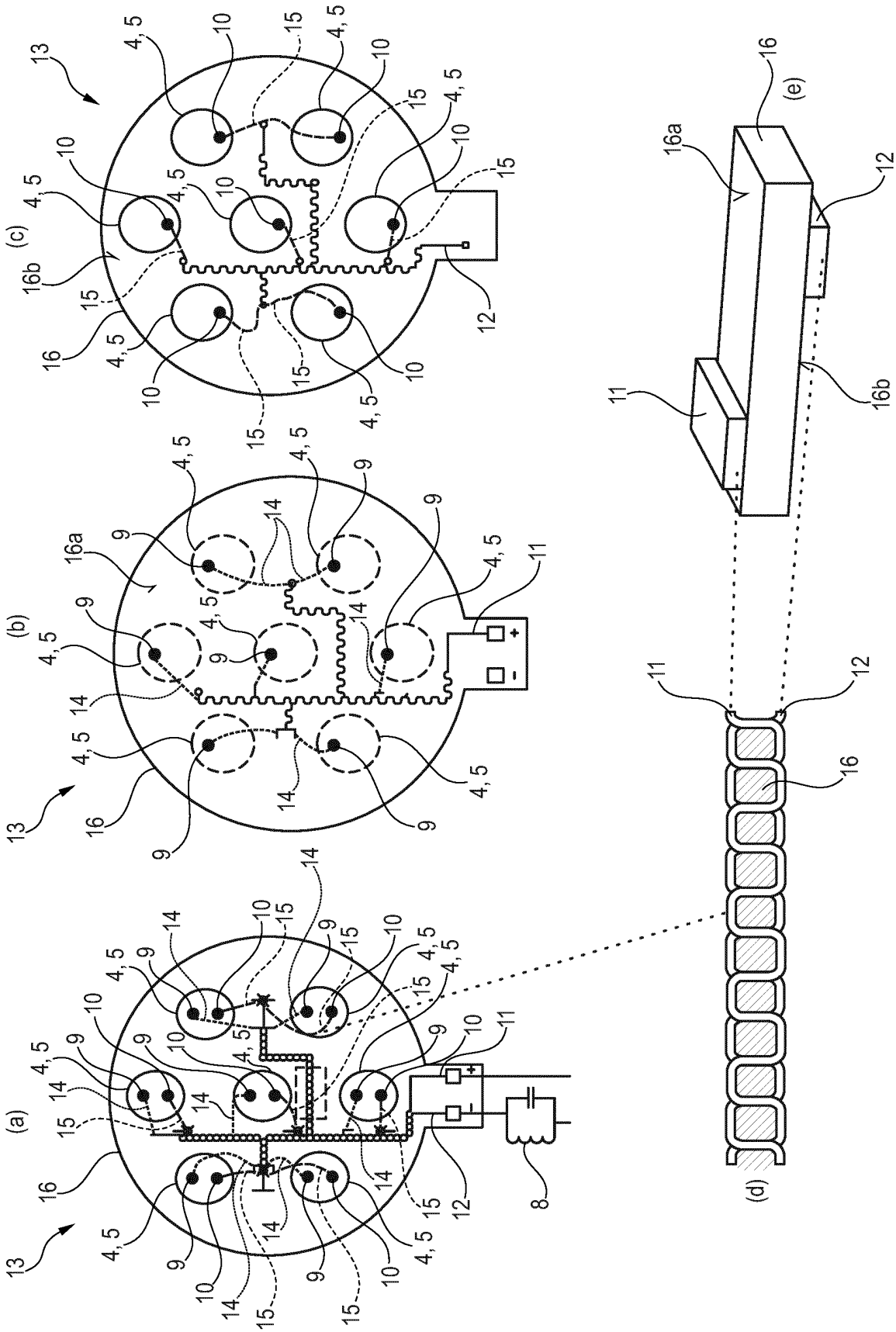

Fig. 3

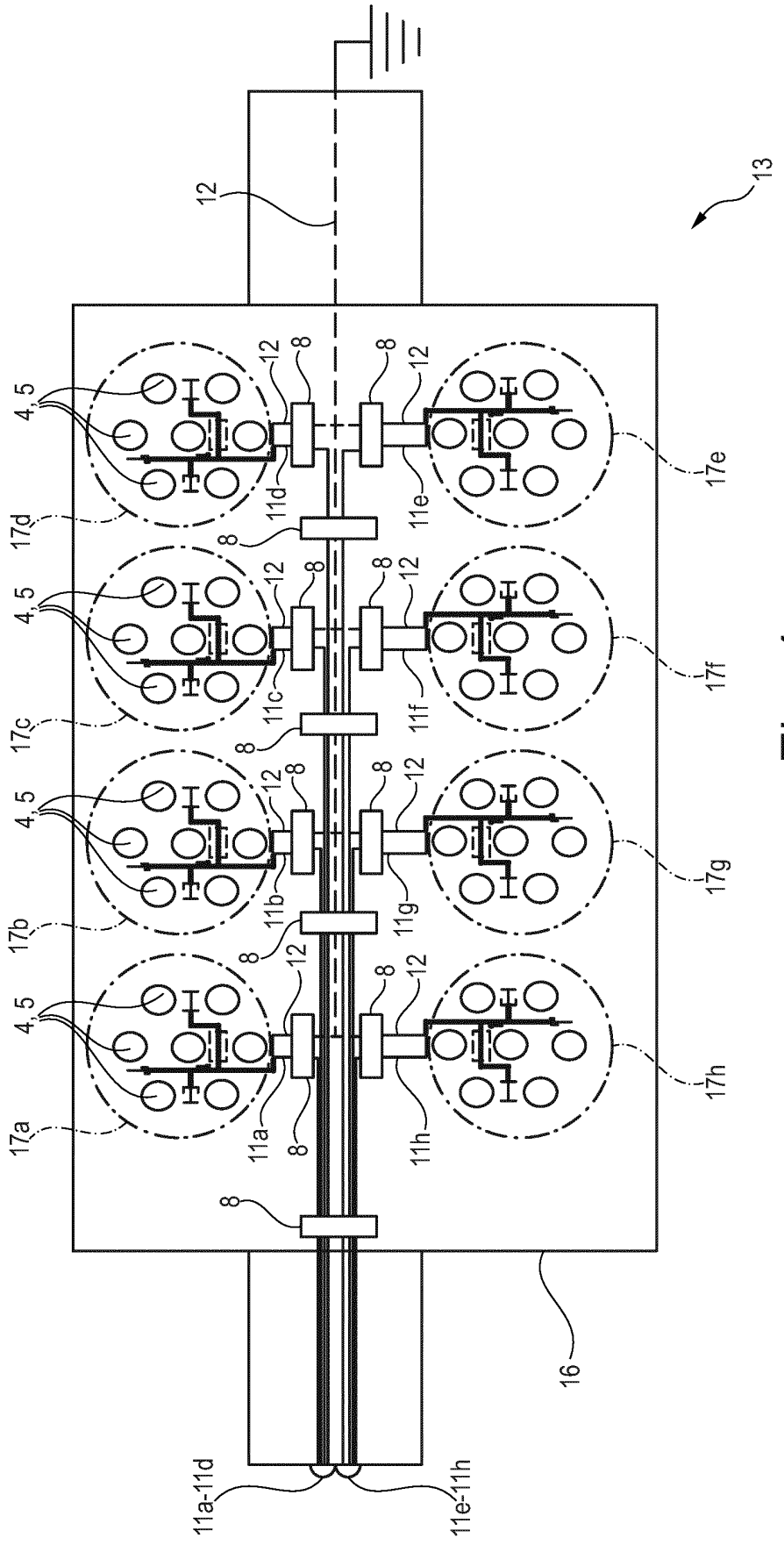
12
13
4,5
8
11d
11e
17d
17e
11c
11f
17c
17f
11b
11g
17b
17g
11a
11h
17a
17h
16
11a-11d
11e-11h

Fig. 4

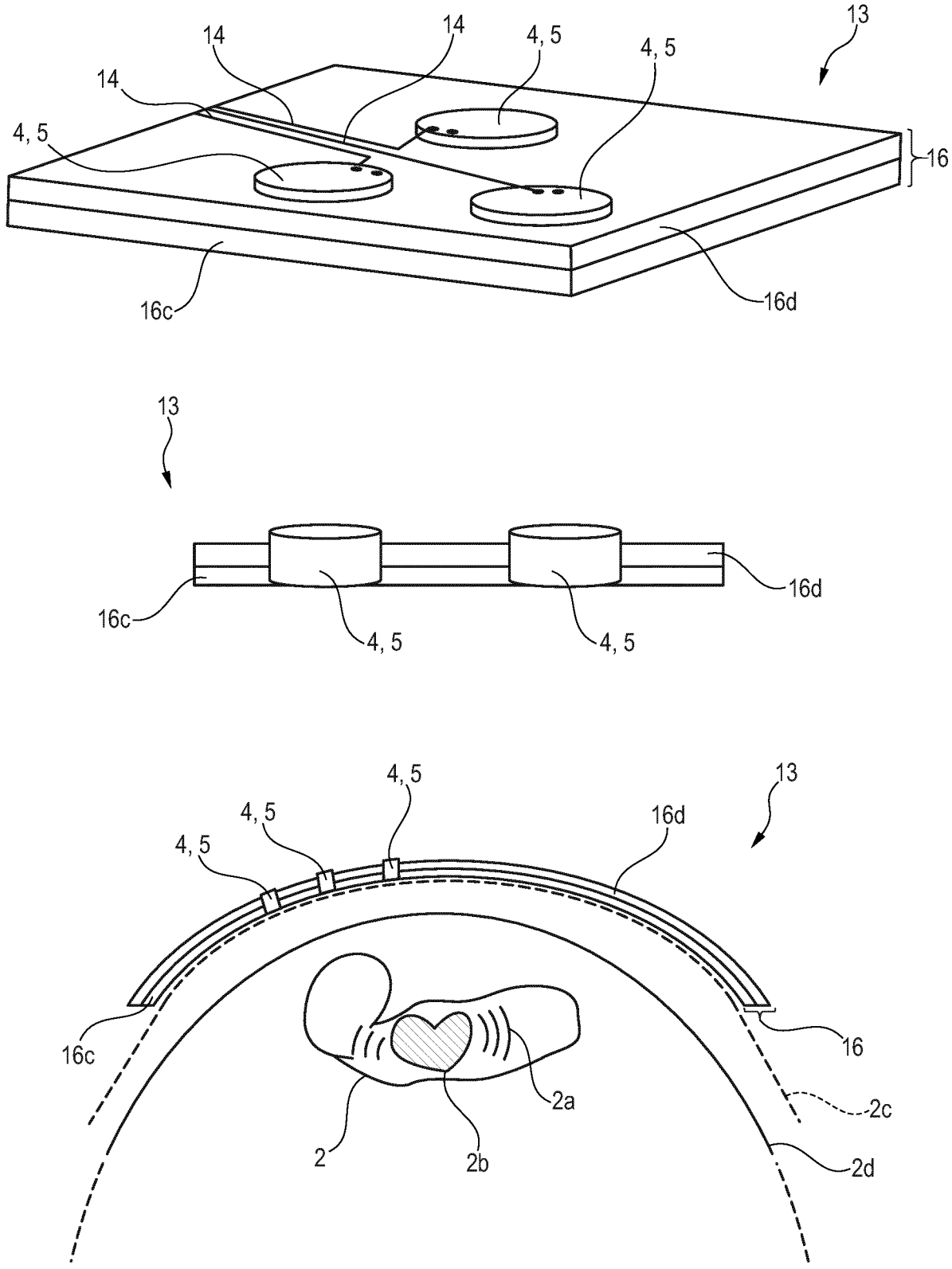
13
14
14
4, 5
4, 5
14
4, 5
16
16c
16d
13
16c
4, 5
4, 5
16d
4, 5
4, 5
4, 5
13
16d
16c
16
2c
2d
2a
2
2b

Fig. 5

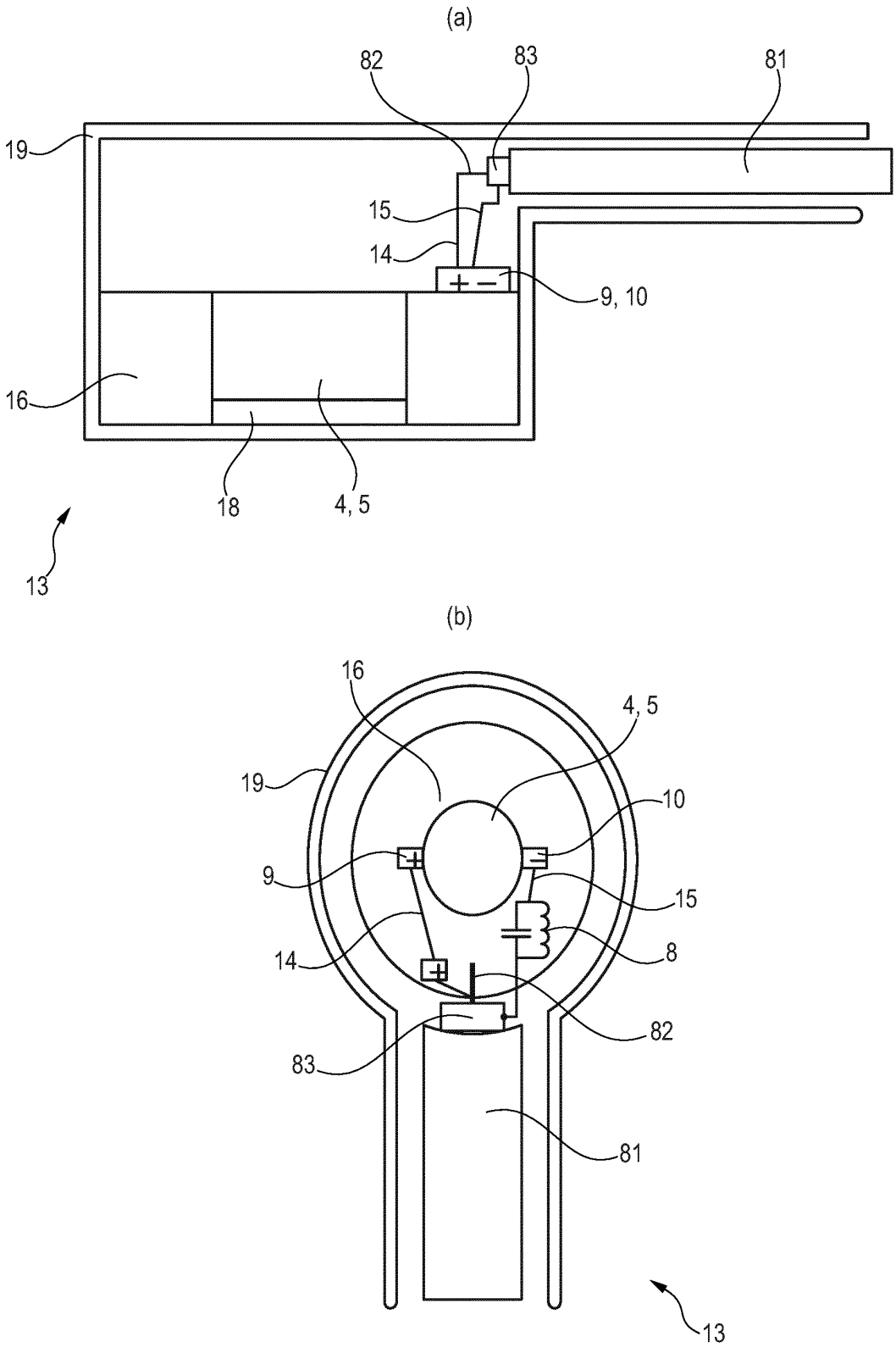

Fig. 6

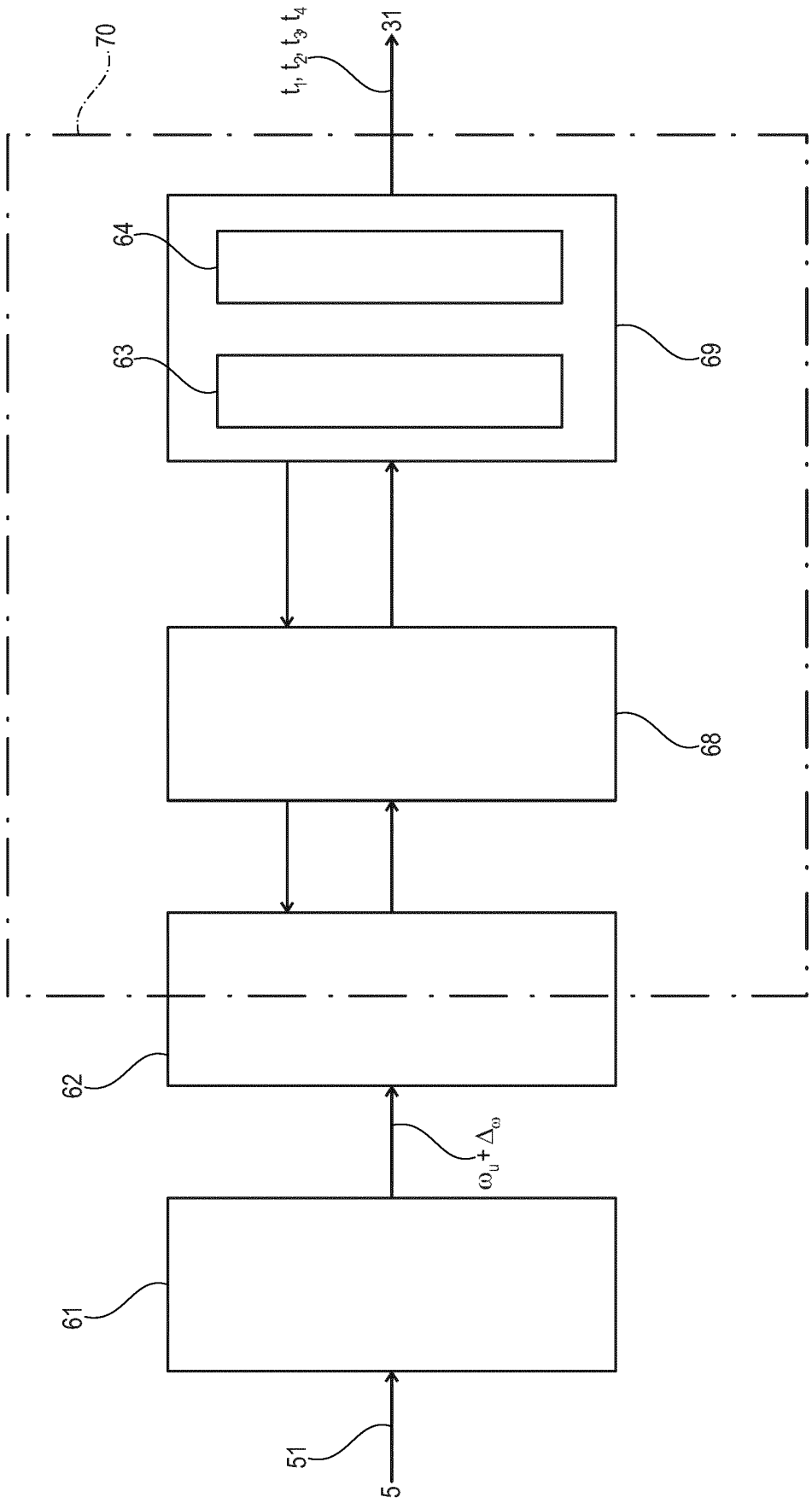

Fig. 7

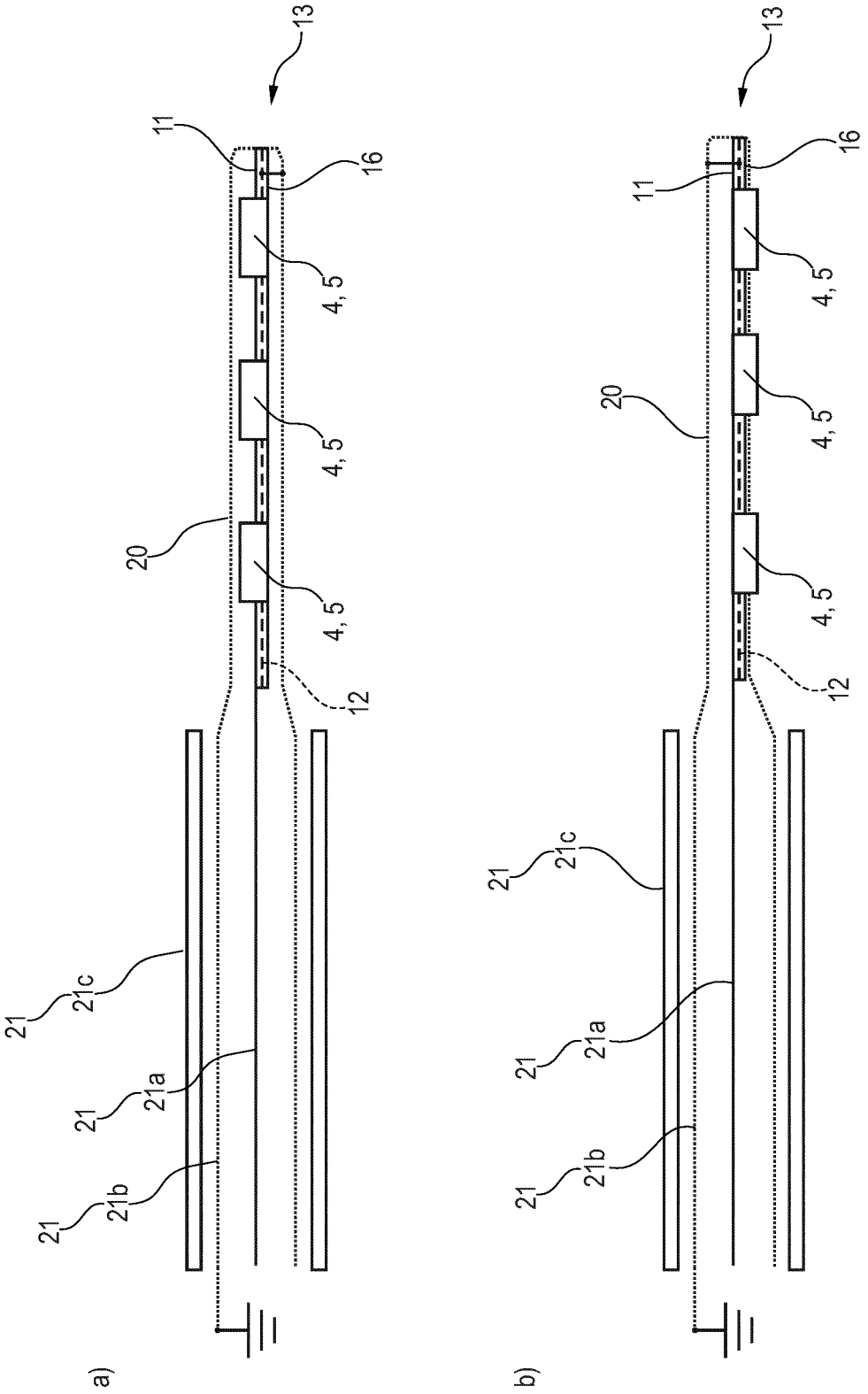

Fig. 8

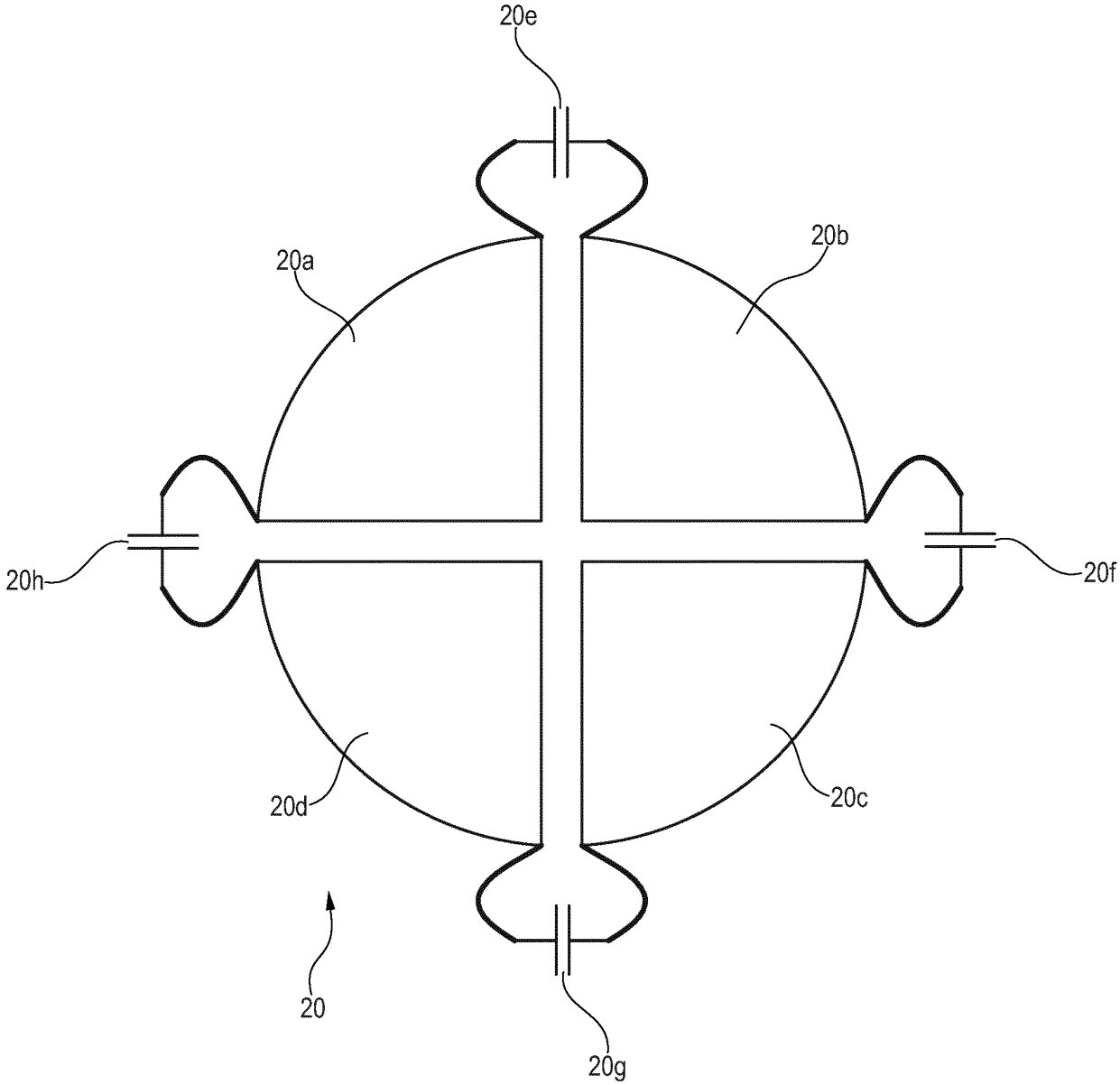

Fig. 9

REFERENCES CITED IN THE DESCRIPTION

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专利名称(译)	用于检测患者心跳的超声设备		
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优先权	2015200597 2015-12-16 EP		
其他公开文献	EP3389501A1		
外部链接	Espacenet		

摘要(译)

用于检测患者（2）的心跳（2a）的设备（1），包括用于超声波（4a）的发射器（4），用于多普勒频移的超声波（5a）的接收器（5）以及评估装置 配置为提取至少一个随时间变化的频率分量，并根据至少一个频率分量的时间变化来评估心跳的时刻t 1，t 2，t 3，t 4。 所述时刻被确定为所述至少一个时变频率分量假定为最大值的时刻，或者被确定为原始时变信号在基于时变频率生成的感兴趣窗口内假定为最大值的时刻。 零件。 心跳可以用于触发图像采集，特别是在MRI设备中。 可以提供跟踪装置以确定哪些发射器和/或接收器覆盖心脏，尤其是用于胎儿的成像。