



(51) International Patent Classification:

A61B 8/06 (2006.01)

A61B 8/00 (2006.01)

A61B 8/08 (2006.01)

G01S 15/89 (2006.01)

A61B 8/13 (2006.01)

(21) International Application Number:

PCT/EP2020/057584

(22) International Filing Date:

19 March 2020 (19.03.2020)

(25) Filing Language:

English

(26) Publication Language:

English

(30) Priority Data:

62/820,549

19 March 2019 (19.03.2019)

US

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(81) Designated States (unless otherwise indicated, for every

kind of national protection available): AE, AG, AL, AM,

AO, AT, AU, AZ, BA, BB, BG, BH, BN, BR, BW, BY, BZ,

CA, CH, CL, CN, CO, CR, CU, CZ, DE, DJ, DK, DM, DO,

DZ, EC, EE, EG, ES, FI, GB, GD, GE, GH, GM, GT, HN,

HR, HU, ID, IL, IN, IR, IS, JO, JP, KE, KG, KH, KN, KP,

KR, KW, KZ, LA, LC, LK, LR, LS, LU, LY, MA, MD, ME,

MG, MK, MN, MW, MX, MY, MZ, NA, NG, NI, NO, NZ,

OM, PA, PE, PG, PH, PL, PT, QA, RO, RS, RU, RW, SA,

SC, SD, SE, SG, SK, SL, ST, SV, SY, TH, TJ, TM, TN, TR,

TT, TZ, UA, UG, US, UZ, VC, VN, WS, ZA, ZM, ZW.

(84) Designated States (unless otherwise indicated, for every

kind of regional protection available): ARIPO (BW, GH,

GM, KE, LR, LS, MW, MZ, NA, RW, SD, SL, ST, SZ, TZ,

UG, ZM, ZW), Eurasian (AM, AZ, BY, KG, KZ, RU, TJ,

TM), European (AL, AT, BE, BG, CH, CY, CZ, DE, DK,

EE, ES, FI, FR, GB, GR, HR, HU, IE, IS, IT, LT, LU, LV,

MC, MK, MT, NL, NO, PL, PT, RO, RS, SE, SI, SK, SM,

(54) Title: THREE DIMENSIONAL VOLUME FLOW QUANTIFICATION AND MEASUREMENT

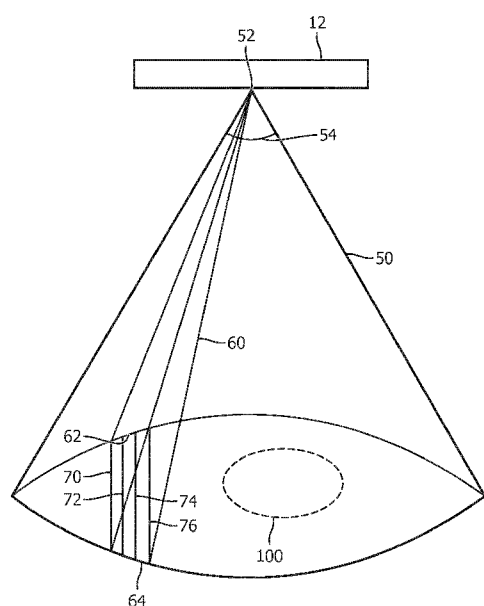


FIG. 1

(57) Abstract: An ultrasonic diagnostic imaging system acquires volume image flow data sets of subvolumes of a blood vessel over at least a cardiac cycle. Image data of the subvolumes is then aligned both spatially and temporally to produce 3D images of the volume flow of the blood vessel over a heart cycle. A volume flow profile curve is produced from the acquired volume image flow data sets. The subvolumes are scanned starting with the center of the blood vessel and proceeding outward therefrom. The blood vessel center may be designated manually by a user or automatically by the ultrasound system by Doppler or other methods. Each subvolume is scanned over a heart cycle, with the systolic phase in the temporal center of the acquisition interval. The subvolumes are scanned in synchronism with the heart cycle and the estimation of a heart cycle is updated during each subvolume data acquisition interval.

TR), OAPI (BF, BJ, CF, CG, CI, CM, GA, GN, GQ, GW,
KM, ML, MR, NE, SN, TD, TG).

Declarations under Rule 4.17:

- *as to applicant's entitlement to apply for and be granted a patent (Rule 4.17(ii))*

Published:

- *with international search report (Art. 21(3))*

THREE DIMENSIONAL VOLUME FLOW
QUANTIFICATION AND MEASUREMENT

5 This invention relates to medical diagnostic
ultrasound systems and, in particular, to ultrasound
systems which produce quantified measurements of the
volume flow of blood through the heart or a blood
vessel.

10 Ultrasound has long been used to assess various
parameters of blood flow in the heart and vascular
system using the Doppler principle. The basic
Doppler response is flow velocity, which can further
be used to determine additional characteristics of
blood flow. One characteristic of interest to
15 cardiologists is the volume flow of blood through a
vessel. Early efforts to estimate volume flow
consisted of multiplying the mean velocity of blood
flow by the nominal cross-sectional area of a blood
vessel. However, these early efforts had
20 shortcomings due to the need to make certain
estimates. One is that the vessel lumen is circular.
Another is the estimation of the mean velocity from a
single Doppler measurement or from a qualitative
assessment of spectral Doppler data. Velocity
25 measurement must also be corrected for the angle
between the ultrasound beam direction and the
direction of flow. Yet another consideration is the
laminar flow profile in the presence of stenosis.

A further complication arises due to the
30 pulsatility of arterial flow. While venous flow is
substantially constant, arterial flow is constantly
changing over the heart cycle. Thus, the standard
techniques often lack for user independency and
repeatability. Some of these demands have been eased
35 by the advent of 3D ultrasound to assess flow

conditions and particularly its ability to acquire volume blood flow information. With 3D imaging, the full vessel lumen can be imaged and a sequence of 3D image data sets acquired for later replay and
5 diagnosis. When data of the full volumetric flow in the vessel is acquired in the data sets, the image data can be examined during post-acquisition diagnosis to assess the flow profile. Different 2D image planes can be extracted from the 3D data in
10 multi-planar reconstruction (MPR), so that an image plane of a desired orientation through a vessel can be examined. Three dimensional imaging thus addresses many of the static imaging challenges which are problematic with 2D flow estimation.

15 In recent years the problem of analyzing the temporal dynamics of blood flow have been addressed by a technique called "spatial-temporal image correlation," or STIC. With STIC, a sweep is made through a blood vessel with ultrasound and many image
20 frames are acquired over a sequence of heart cycles. When done by manually scanning with a 2D ultrasound probe, this image acquisition can take ten seconds or longer. The same acquisition can be performed with a mechanical 3D probe which mechanically sweeps the
25 image plane through the vessel, but 3D mechanical probes often have poorer elevation focus which leads to inaccuracies when constructing MPR images in the elevation dimension. After the acquisition is complete and the image frames are stored, image
30 frames of the desired anatomy, created by MPR reconstruction if necessary, are reassembled into a loop of images according to their phase sequence in the cardiac cycle. This task is made difficult by the fact that the heart cycle may not be uniform over
35 the time required to acquire the data sets.

Consequently, synthetic methods of estimating the heart rate from analysis of the movement of cardiac tissue or blood have been developed, which nonetheless is often difficult to assess and prone to inaccuracy. Accordingly, it is desirable to develop more robust techniques for accurately assessing volume flow in the presence of flow pulsatility and erratic heartbeats. It is further desirable to automate such techniques so as to prioritize and shorten the time interval needed to acquire the volume flow data and reduce the impact of motional effects by both the probe and the anatomy.

In accordance with the principles of the present invention, a diagnostic ultrasound system is described which produces a blood flow profile using volume flow acquisition. A 3D imaging probe is used to acquire one or more image volumes of flow data from a vessel and a B- or C- cross-sectional plane of the vessel is extracted which is processed to determine the blood volume flow rate. When arterial flow is being assessed, it is preferable to acquire multiple subvolumes of the volume flow, with each spatially different volume flow dataset being acquired over all phases of a cardiac cycle to maintain temporal sampling precision. The subvolumes are then spatially assembled in cardiac phase order so as to produce a full volume flow sequence with adequate temporal sampling. The B- or C-plane is then extracted and a volume flow profile estimated through the plane using Gauss's theorem.

In a preferred implementation, volume or subvolume acquisition starts around the center of the vessel where flow signals are stronger and thus cardiac phases are easier to identify. This also results in acquisition of subvolumes with the

greatest contribution to total volume flow earlier in the acquisition process to minimize adverse motional effects. The acquisition process will in this case begin by identifying the vessel center prior to
5 subvolume acquisition, as by user designation or an automated technique such as a rapid Doppler sequence through the vessel. Subvolume acquisition then begins from the vessel center and proceeds outward therefrom.

10 Acquisition in synchronism with the phase of the heart cycle can be achieved by a non-synthetic method of measuring the heart cycle (e.g., an ECG monitor attached to the patient), or by a synthetic method such as user estimation or automatic M-mode or
15 speckle tracking of cardiac or vascular anatomy or blood flow. Preferably the flow profile is calculated on the fly during acquisition, and heart rate estimations based on the cardiac cycles of previously acquired subvolumes are updated each heart
20 cycle to account for irregularities in the heart cycle. Preferably the acquisition of each subvolume is temporally centered about the systolic phase, so that systolic flow, when volume flow is greatest in arterial vessels, is fully sampled.

25 According to aspects of the invention, an ultrasonic diagnostic imaging system for analyzing volume flow of blood includes a 3D imaging probe adapted to acquire volume image flow data sets of a blood vessel, an image data processor responsive to
30 the volume image flow data sets, and a vessel center locator. The vessel center locator is responsive to spatially organized blood vessel data, which is adapted to identify a center of the blood vessel. The system further includes a beamformer controller,
35 responsive to the vessel center locator, which is

adapted to control the 3D imaging probe to acquire volume image flow data sets of the blood vessel commencing around the center of the vessel. In certain embodiments, there is also a volume flow calculator, which is responsive to acquired volume image flow data sets of the blood vessel, which is adapted to calculate volume flow profile data.

In some embodiments, the ultrasonic diagnostic imaging system further includes a heart rate calculator adapted to produce data representing a heart rate and the volume image flow data sets are acquired in timed relation to the heart rate data.

In some embodiments, the heart rate calculator includes one of an ECG monitor or an ultrasound data processor which is adapted to produce estimated heart rate data using ultrasound data.

In some embodiments, the ultrasonic diagnostic imaging system includes a subvolume selector responsive to the vessel center locator. The subvolume selector controls the beamformer controller to acquire volume image flow data sets of subvolumes of a blood vessel commencing around the center of the vessel.

In some embodiments, the beamformer controller acquires volume image flow data sets of a subvolume of a blood vessel over the duration of a heart cycle.

In some embodiments, the beamformer controller acquires volume image flow data sets of a subvolume of a blood vessel over an acquisition interval commencing in the middle of a diastolic portion of a heart cycle and ending in the middle of the next diastolic portion of a heart cycle.

In some embodiments, the beamformer controller acquires volume image flow data sets of a subvolume of a blood vessel during systolic heart phases

occurring in the middle of the acquisition interval.

In some embodiments, the heart rate calculator is further coupled to the volume flow calculator and adapted to calculate a subvolume acquisition time in relation to systolic peaks of a flow profile.

In some embodiments, the ultrasonic diagnostic imaging system further includes a 3D image data memory adapted to store volume image data sets. In some embodiments, the ultrasonic diagnostic imaging system further includes a multi-planar reformatter, which is coupled to the 3D image data memory and selects an image plane intersecting the blood vessel.

In some embodiments, the volume flow calculator is further adapted to calculate volume flow profile data in relation to the image plane intersecting the blood vessel.

In some embodiments, the system further includes a volume renderer coupled to the 3D image data memory and adapted to produce a 3D image of the blood vessel.

In some embodiments, the system further includes a display configured to display one or more of an image plane selected by the multi-planar reformatter, a 3D image of the blood vessel, and a flow profile curve. The display may be coupled to one or more other elements of the system.

The present invention also provides a method of analyzing volume flow of blood by ultrasound data acquisition comprising: identifying a center of a blood vessel; acquiring volume image flow data sets of the blood vessel commencing around the center of the vessel with a 3D ultrasound imaging probe; processing image data sets acquired with the imaging probe for the display of an ultrasound image of the blood vessel; and calculating volume flow profile

data using the acquired volume image flow data sets.

In some embodiments, the method also includes estimating heart cycle timing. In some embodiments, volume image flow data sets of the blood vessel are
5 acquired from subvolumes of the vessel in synchronism with the heart cycle timing.

In some embodiments, the method includes detecting a systolic phase of the heart cycle. In some embodiments, the volume image flow data sets of
10 the blood vessel are acquired from a subvolume of the vessel during an acquisition interval starting in mid-diastole of a heart cycle and ending in mid-diastole of a subsequent heart cycle, wherein acquisition during the systolic phase occurs during
15 the middle of the acquisition interval.

In some embodiments, acquiring the volume image flow data sets of the blood vessel from subvolumes of the vessel further comprises updating the estimated heart cycle timing during a plurality of the
20 subvolume acquisition intervals.

In the drawings:

FIGURE 1 illustrates the acquisition of a subvolume of ultrasound data of a volumetric region
25 using a 3D ultrasonic imaging probe.

FIGURE 2 illustrates a sequence of acquisition of subvolumes of ultrasound data of a blood vessel lumen which starts from the center of the vessel.

FIGURE 3 illustrates a typical series of flow profiles of the acquisition sequence of FIGURE 2 when
30 each subvolume acquisition is temporally centered about the systolic phase.

FIGURE 4 illustrates how volume flow over-estimation or under-estimation can arise when the
35 heart cycle is inaccurately measured or estimated.

FIGURE 5 illustrates the determination of a subsequent subvolume acquisition interval from the sampling of an initial subvolume over two heart cycles.

5 FIGURES 6 illustrates the determination of a subsequent subvolume acquisition interval when sampling of the initial subvolume begins during systole.

10 FIGURE 7 illustrates in block diagram form an ultrasonic diagnostic imaging system constructed in accordance with the principles of the present invention.

15 FIGURE 1 illustrates the acquisition of a subvolume 60 of ultrasound data of a volumetric region 50 by the transducer 12 of a 3D ultrasonic imaging probe. In this example the full volume 50 has a conical shape subtending a sector angle 54 with an apex 52 located at the front surface of an array transducer 12. For subvolume scanning the full
20 volume may be scanned by acquiring ultrasound data from a series of adjacent subvolumes 60. Each subvolume may comprise a series of two-dimensional sector-shaped slices 70, 72, 74 and 76, where each slice comprising a series of adjacent scanlines.
25 While the 3D probe may comprise an oscillating one-dimensional transducer array, preferably the probe comprises a stationary two-dimensional matrix array transducer for improved scanning speed and accuracy. Since beams of a matrix array are electronically
30 steered in both elevation and azimuth, the full volume and the individual subvolumes may have any desired size and shape, and the slices of a subvolume can be scanned in any beam order sequence. The illustration of FIGURE 1 also shows a dashed outline
35 100 where a blood vessel lumen intersects the base of

the volume 50 between the front and back edges of the volume 64 and 62.

It is preferable to acquire subvolume data of a vessel lumen beginning in the center of the lumen where flow signals are strongest and it is easier to reliably identify cardiac phases. More flow signals are contained in a central subvolume and preserved after wall filtering for analysis such as heart rate estimation as discussed below. Central subvolumes also comprise the greatest contributions to total volume flow of the vessel. Such an acquisition sequence is illustrated in FIGURE 2 where subvolumes 1-5 are acquired of a vessel lumen 100 starting with a subvolume from the center of the vessel. The vessel center may be identified by user designation thereof. For instance, the user can view a B mode image of the blood vessel, which shows the vessel wall. The user clicks a cursor at what appears to be the center of the vessel, and this identification is used by the ultrasound system to demarcate the region where acquisition commences. Automated techniques can alternatively be used. For instance, prior to acquisition a series of Doppler beams are transmitted and received from the entire blood vessel and Doppler processed to produce velocity estimates along all of the beams. The spatial location on a beam with the highest velocity may then be used as demarcating the vessel center, and subvolume acquisition then proceeds from this location outward. The vessel center acquired by one of these or another technique is marked with an "X" in FIGURE 2. Acquisition then commences with the acquisition of subvolume 1 which includes the vessel center. Temporally different acquisitions of flow (velocity) data of the subvolume are performed for a complete set of phases of the

heart cycle. The acquisition of subsequent subvolumes then continues outward from the central location, including subvolumes 2, 3, 4 and 5 in this example. As can be seen from the drawing, the initially acquired subvolume includes the greatest amount of flow data, and the peripheral subvolumes 4 and 5 contain the smallest contributions to the total volume flow of the vessel.

FIGURE 3 illustrates these differences in volume flow content, represented by the amplitudes of the flow signal levels of the different subvolumes. The flow signal 81 of the first subvolume is seen to exhibit the greatest amplitude, with the flow signals 82 and 83 of subvolumes 2 and 3 being of slightly lesser amplitudes. The peripheral subvolumes 4 and 5 exhibit flow signals 84 and 85 with the smallest amplitude and hence the smallest contributions to total volume flow, as well as the greatest possibility of flow phase ambiguity. It is for this reason, as well as motional effect minimization, that data from these peripheral subvolumes are the last to be acquired.

It is also seen in FIGURE 3 that the systolic phases, where the flow velocity peaks at peak systole, is in the temporal middle of each subvolume acquisition interval. This sampling timing advantageously results in complete acquisition of flow during systole when volume flow is the greatest. Inaccuracies that can occur with other heart phase timings are illustrated in FIGURE 4, which shows flow profiles 80 of four differently phased acquisitions. FIGURE 4a illustrates a situation where the estimation of the heart rate is greater than the actual heart rate. Since the heart rate is estimated to be more rapid than the actual heart rate, the

interval of subvolume acquisition is shorter than it should be, as illustrated by the portion 80a of the profile shown in bold. Consequently, the acquisition interval omits some of the phases of diastole, and the resulting data is overly dominated by greater systolic flow. As a result, the volume flow rate will be over-estimated for the subvolume. Over-estimation of the heart rate can also result in the situation of FIGURE 4b, where acquisition during the interval of 80b misses some systolic phase information, in this example, the second half of systole. As a result, the volume flow rate is under-estimated for the subvolume. FIGURE 4c shows a situation where the heart rate is estimated to be less (slower) than the actual heart rate. As a result, an acquisition interval will be greater than one heart cycle. In this example systole is oversampled when the subvolume is sampled during the acquisition interval of 80c, and volume flow is over-estimated. FIGURE 4d illustrates another situation where the heart rate is estimated to be less than the actual heart rate. The subvolume is sampled during the acquisition interval of bold flow profile line 80d. In this case, the subvolume is over-sampled during diastole, and as a result volume flow is under-estimated due to the inclusion of excessive diastolic data.

It is thus seen that accurate heart rate information is important for accurate volume flow assessment. In accordance with a further aspect of the present invention, when the heart rate is determined from ultrasonic signal information, the heart rate estimation is continuously updated during each subvolume acquisition to properly adjust for the occurrence of a longer or shorter interval between

heartbeats. A heartrate estimation can be estimated from M-mode data as described in US Pat. 9,357,978 (Dow et al.) This can be done in the background, with the heartrate estimated even before volume flow acquisition begins, thereby enabling the acquisition of the first subvolume in proper synchrony with the heartrate. Another way to determine the heartrate is to continuously calculate the flow profile signal during each subvolume acquisition, and use this updated information to properly synchronize the following acquisition with the phase of the heart. One example of this technique is illustrated in FIGURE 5. In this example the first subvolume is acquired over two heart cycles, the first exhibiting a systolic flow phase 80_1 which peaks at time T_1 and the second exhibiting a systolic flow phase 80_2 which peaks at time T_2 . The two systolic peaks are detected by peak detection of the flow profile signal, and the peak-to-peak interval ΔT is determined. The time of commencement of acquisition of the second subvolume is then set as a function of the ΔT interval and the time of occurrence of T_2 . Preferably the fraction of the ΔT interval is one-half, which causes acquisition of the second subvolume to start midway in the diastolic phase. Thus, calculating $T_2 + r\Delta T$ when r is set to 0.5 will cause acquisition of the subsequent subvolume to commence in the middle of the diastolic phase and result in sampling the next subvolume over the interval of waveform 80_3 in FIGURE 5. This timing will advantageously result in sampling of the second subvolume with its systolic phase occurring during the middle of the acquisition interval, assuring that all of the systolic phase will be sampled.

FIGURE 6 provides a second example of use of the flow profile waveform to establish subvolume

acquisition timing. In this example acquisition of subvolume 1 starts at T_1' , just after a systolic peak of the flow signal. The same level of the next systolic phase is detected, which occurs at time T_2' ,
5 and the interval ΔT between T_1' and T_2' is determined. The equation $T_2 + r\Delta T$ is calculated which yields a starting time for acquiring the second subvolume as shown at 86. While the first subvolume is not sampled with the systolic phase temporally located in
10 the middle of the subvolume acquisition interval, properly phased acquisition will occur for the second and all subsequent subvolumes.

In FIGURE 7, an ultrasound system constructed in accordance with the principles of the present
15 invention is shown in block diagram form. A transducer array 12 is provided in an ultrasound probe 10 for transmitting ultrasonic waves and receiving echo information. The transducer array 12 is an array of transducer elements capable of
20 scanning in three dimensions, in both elevation and azimuth. The transducer array 12 is coupled to a microbeamformer 14 in the probe which controls transmission and reception of signals by the array elements. Microbeamformers are capable of at least
25 partial beamforming of the signals received by groups or "patches" of transducer elements as described in US Pats. 5,997,479 (Savord et al.), 6,013,032 (Savord), and 6,623,432 (Powers et al.) The microbeamformer is coupled by the probe cable to a
30 transmit/receive (T/R) switch 16 which switches between transmission and reception and protects the main beamformer 18 from high energy transmit signals. The transmission of ultrasonic beams from the transducer array 12 under control of the
35 microbeamformer 14 is directed by a beamformer

controller 17 coupled to the T/R switch and the main beamformer 18, which receives input from the user's operation of the user interface or control panel 38. Among the transmit characteristics controlled by the transmit controller are the direction, number, spacing, amplitude, phase, frequency, polarity, and diversity of transmit waveforms. Beams formed in the direction of pulse transmission may be steered straight ahead from the transducer array, or at different angles on either side of an unsteered beam for a wider sector field of view.

The echoes received by a contiguous group of transducer elements are beamformed by appropriately delaying them and then combining them. The partially beamformed signals produced by the microbeamformer 14 from each patch are coupled to the main beamformer 18 where partially beamformed signals from individual patches of transducer elements are combined into a fully beamformed coherent echo signal. For example, the main beamformer 18 may have 128 channels, each of which receives a partially beamformed signal from a patch of 12 transducer elements. In this way the signals received by over 1500 transducer elements of a two-dimensional matrix array transducer can contribute efficiently to a single beamformed signal.

The coherent echo signals undergo signal processing by a signal processor 20, which includes filtering by a digital filter and noise reduction as by spatial or frequency compounding. The signal processor may also perform speckle reduction as by spatial or frequency compounding. The digital filter of the signal processor 20 can be a filter of the type disclosed in U.S. Patent No. 5,833,613 (Averkiou et al.), for example. The echo signals are then coupled to a quadrature bandpass filter (QBP) 22.

The QBP performs three functions: band limiting the r.f. echo signal data, producing in-phase and quadrature pairs (I and Q) of echo signal data, and decimating the digital sample rate. The QBP
5 comprises two separate filters, one producing in-phase samples and the other producing quadrature samples, with each filter being formed by a plurality of multiplier-accumulators (MACs) implementing an FIR filter.

10 The beamformed and processed coherent echo signals are coupled to a pair of image data processors. A B mode processor 26 produces signal data for a B mode image of structure in the body such as tissue. The B mode processor performs amplitude
15 (envelope) detection of quadrature demodulated I and Q signal components by calculating the echo signal amplitude in the form of $(I^2+Q^2)^{1/2}$. The quadrature echo signal components are also coupled to a Doppler processor 24. The Doppler processor 24 stores
20 ensembles of echo signals from discrete points in an image field which are then used to estimate the Doppler shift at points in the image with a fast Fourier transform (FFT) processor. The rate at which the ensembles are acquired determines the velocity
25 range of motion that the system can accurately measure and depict in an image. The Doppler shift is proportional to motion at points in the image field, e.g., blood flow and tissue motion. For color
Doppler image data, the estimated Doppler flow values
30 at each point in a blood vessel are wall filtered and converted to color values using a look-up table. The wall filter has an adjustable cutoff frequency above or below which motion will be rejected such as the low frequency motion of the wall of a blood vessel
35 when imaging flowing blood. The B mode image data

and the Doppler flow values are coupled to a scan converter 28 which converts the B mode and Doppler samples from their acquired R- θ coordinates to Cartesian (x,y) coordinates for display in a desired display format, e.g., a rectilinear display format or a sector display format. Either the B mode image or the Doppler image may be displayed alone, or the two shown together in anatomical registration in which the color Doppler overlay shows the blood flow in B mode processed tissue and vessels in the image. Another display possibility is to display side-by-side images of the same anatomy which have been processed differently. This display format is useful when comparing images. The scan-converted image data, both B mode and Doppler data, is coupled to and stored in a 3D image data memory 30 where it is stored in memory locations addressable in accordance with the spatial locations from which the image data values were acquired. Image data from 3D scanning can be accessed by a volume renderer 32, which converts the data values of a 3D data set into a projected 3D image as viewed from a given reference point as described in US Pat. 6,530,885 (Entrekin et al.) The 3D images produced by the volume renderer 32 and 2D images from data produced by the scan converter 28 are coupled to a display processor 34 for further enhancement, buffering and temporary storage for display on an image display 36. The 3D image data is also coupled to a multi-planar reformatter 48 which, in response to user input from the user controls 38, is able to extract image data for a user-designated image plane from the 3D dataset. This image data is coupled to the display processor 34 for display of a selected MPR image, and

the plane of the MPR image is used in the estimation of volume flow as described below.

In accordance with the present invention, B mode and Doppler data produced by processors 24 and 26 are
5 coupled to a vessel center locator 44. This enables the vessels center locator to do several things. One is to enable a user to click on a point in a B mode image of a blood vessel which the user believes is the center of the vessel. A signal indicating this
10 user action is coupled to the vessel center locator from the user interface 38, and the identified vessel center point is stored in the locator and coupled to subvolume selector 46. Another operation of the vessel center locator 44 is to receive velocity data
15 from the Doppler processor after a rapid Doppler scan of a blood vessel. The locator 44 analyzes this data to determine the spatial location in the vessel with the highest flow velocity. In that case, this spatial location is coupled to the subvolume selector
20 46 and used as the vessel center. It will be appreciated that the B mode and Doppler data coupled to the vessel center locator can be that which is processed by scan conversion, so that the spatial location coordinates will correspond with that used
25 by the display 36. The vessel center locator 44 is thus capable of using either user input or automated methods to determine a vessel center and couple that information to the subvolume selector 46.

The multi-planar reformatter 48 is also coupled
30 to a volume flow calculator 40. The volume flow calculator also receives Doppler velocity data from the Doppler processor 24 and is thus able to compute the volume blood flow through a B- or C-plane of a blood vessel using Gauss's theorem. For volume flow
35 data, Gauss's theorem is calculated as:

$$Q = \int_S v \cdot dA$$

where Q is the volume flow in, e.g., milliliters per second, v is flow velocity, and the surface S is a selected plane through a vessel lumen. A surface
5 integral of velocity v over the enclosing boundary S yields the volume flow Q . Volume flow through a plane intersecting a blood vessel can thus be updated with new data for each new phase of the heart cycle to produce a flow profile curve of Q as a function of
10 time, and the flow volume over the phases of an entire heart cycle can be summed to calculate the volume flow per heart cycle.

The flow data of volume flow produced by the volume flow calculator 40 is coupled to a graphics
15 generator 49, which produces a flow profile curve such as that shown in FIGURE 4 for display on the display 36. The graphics generator also produces graphics for display with the ultrasound image for things such as cursors, measurement dimensions, exam
20 parameters, and patient name.

The flow profile curve data is also coupled to a heart rate calculator 42, where it is used to estimate the heartrate in the absence of ECG monitor signals or user input of a heartrate value. The
25 heartrate calculator uses the flow profile curve data to detect systolic peaks of the flow profile, to detect the interval ΔT between systolic peaks, and to calculate the start times for successive subvolume acquisitions as described above. The heartrate
30 timing data is coupled to the subvolume selector 46, which determines when to acquire each subvolume needed to scan the entire volumetric region of a vessel. The data from the vessel center locator 44 informs the subvolume selector of where the first

subvolume is to be acquired (*i.e.*, around the vessel center,) and the data from the heart rate calculator 42 informs the subvolume selector of the timing of each subvolume acquisition so that the systolic phase
5 will be acquired in the middle of each subvolume acquisition for at least the second and subsequent subvolume acquisitions. Acting on this information, the subvolume selector informs the beamformer controller of when and where each subvolume
10 acquisition is to be performed. The ultrasound system of FIGURE 7 thereby performs acquisition and assembly of a volume image of volume flow data for the accurate estimation of a volume flow profile curve.

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WHAT IS CLAIMED IS:

1. An ultrasonic diagnostic imaging system for analyzing volume flow of blood comprising:
 - 5 a 3D imaging probe adapted to acquire volume image flow data sets of a blood vessel;
 - an image data processor responsive to the volume image flow data sets;
 - a vessel center locator, responsive to spatially organized blood vessel data, which is adapted to identify a center of the blood vessel;
 - 10 a beamformer controller, responsive to the vessel center locator, which is adapted to control the 3D imaging probe to acquire volume image flow data sets of the blood vessel commencing around the center of the vessel; and
 - 15 a volume flow calculator, responsive to acquired volume image flow data sets of the blood vessel, which is adapted to calculate volume flow profile data.
2. The ultrasonic diagnostic imaging system of Claim 1, further comprising a heart rate calculator adapted to produce data representing a heart rate,
 - 25 wherein the volume image flow data sets are acquired in timed relation to the heart rate data.
3. The ultrasonic diagnostic imaging system of Claim 2, wherein the heart rate calculator further comprises one of an ECG monitor or an ultrasound data processor which is adapted to produce estimated heart rate data using ultrasound data.
4. The ultrasonic diagnostic imaging system of Claim 1, further comprising a subvolume selector

responsive to the vessel center locator which is adapted to control the beamformer controller to acquire volume image flow data sets of subvolumes of a blood vessel commencing around the center of the vessel.

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5. The ultrasonic diagnostic imaging system of Claim 4, wherein the beamformer controller is further adapted to acquire volume image flow data sets of a subvolume of a blood vessel over the duration of a heart cycle.

6. The ultrasonic diagnostic imaging system of Claim 5, wherein the beamformer controller is further adapted to acquire volume image flow data sets of a subvolume of a blood vessel over an acquisition interval commencing in the middle of a diastolic portion of a heart cycle and ending in the middle of the next diastolic portion of a heart cycle.

7. The ultrasonic diagnostic imaging system of Claim 6, wherein the beamformer controller is further adapted to acquire volume image flow data sets of a subvolume of a blood vessel during systolic heart phases occurring in the middle of the acquisition interval.

8. The ultrasonic diagnostic imaging system of Claim 2, wherein the heart rate calculator is further coupled to the volume flow calculator and adapted to calculate a subvolume acquisition time in relation to systolic peaks of a flow profile.

9. The ultrasonic diagnostic imaging system of Claim 1, further comprising:

a 3D image data memory adapted to store volume image data sets; and

a multi-planar reformatter, coupled to the 3D image data memory, and adapted to select an image plane intersecting the blood vessel,

wherein the volume flow calculator is further adapted to calculate volume flow profile data in relation to the image plane intersecting the blood vessel.

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10. The ultrasonic diagnostic imaging system of Claim 9, further comprising a volume renderer coupled to the 3D image data memory and adapted to produce a 3D image of the blood vessel.

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11. The ultrasonic diagnostic imaging system of Claim 10, further comprising a display, coupled to the multi-planar reformatter, the volume renderer, and the volume flow calculator, which is adapted to display one or more of an image plane selected by the multi-planar reformatter, a 3D image of the blood vessel, and a flow profile curve.

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12. A method of analyzing volume flow of blood by ultrasound data acquisition comprising:

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identifying a center of a blood vessel;

acquiring volume image flow data sets of the blood vessel commencing around the center of the vessel with a 3D ultrasound imaging probe;

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processing image data sets acquired with the imaging probe for the display of an ultrasound image of the blood vessel; and

calculating volume flow profile data using the acquired volume image flow data sets.

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13. The method of Claim 12, further comprising:
estimating heart cycle timing;

5 wherein acquiring volume image flow data sets
further comprises acquiring volume image flow data
sets of the blood vessel from subvolumes of the
vessel in synchronism with the heart cycle timing.

14. The method of Claim 13, further comprising:
detecting a systolic phase of the heart cycle;

10 wherein acquiring volume image flow data sets
further comprises acquiring volume image flow data
sets of the blood vessel from a subvolume of the
vessel during an acquisition interval starting in
mid-diastole of a heart cycle and ending in mid-
15 diastole of a subsequent heart cycle,

wherein acquisition during the systolic phase
occurs during the middle of the acquisition interval.

15. The method of Claim 14,

20 wherein acquiring volume image flow data sets of
the blood vessel from subvolumes of the vessel
further comprises updating the estimated heart cycle
timing during a plurality of the subvolume
acquisition intervals.

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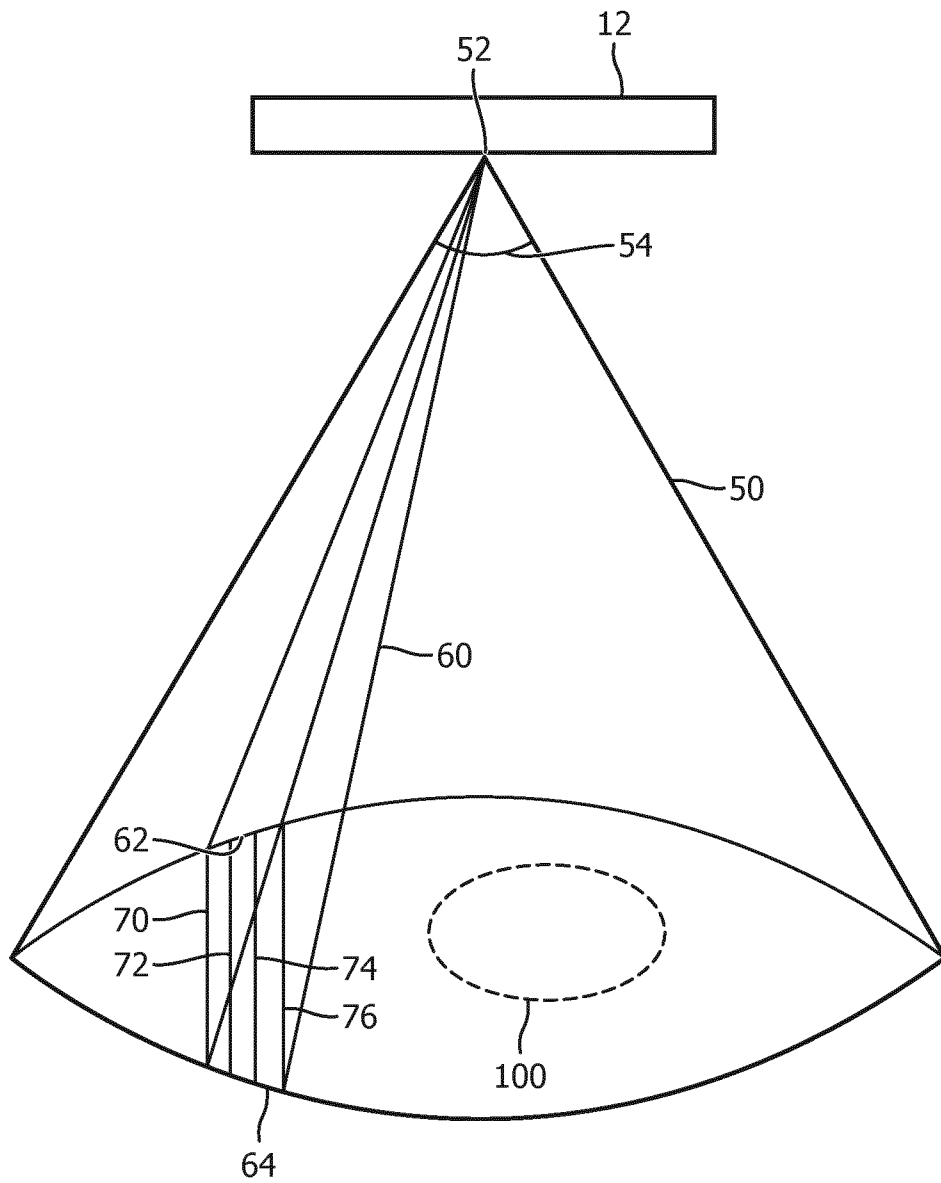


FIG. 1

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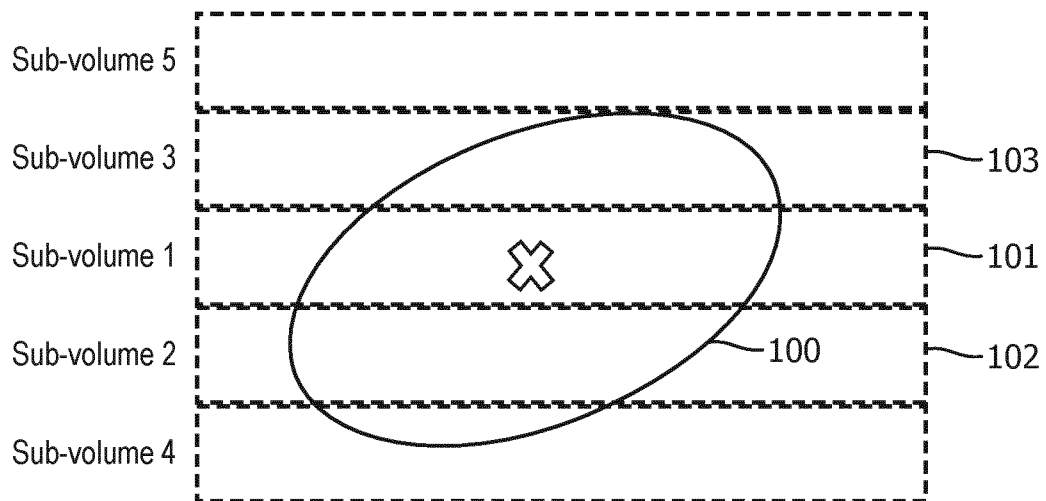


FIG. 2

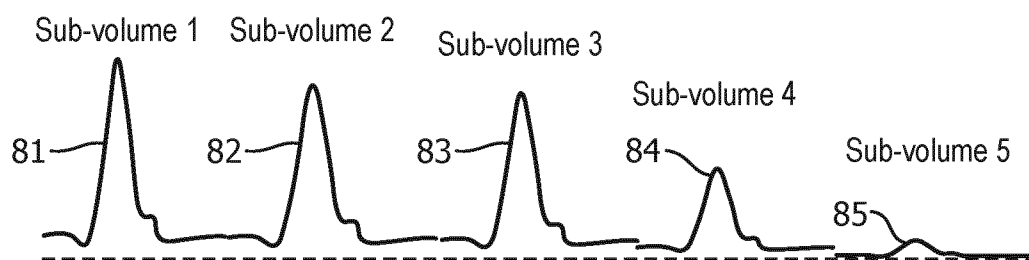


FIG. 3

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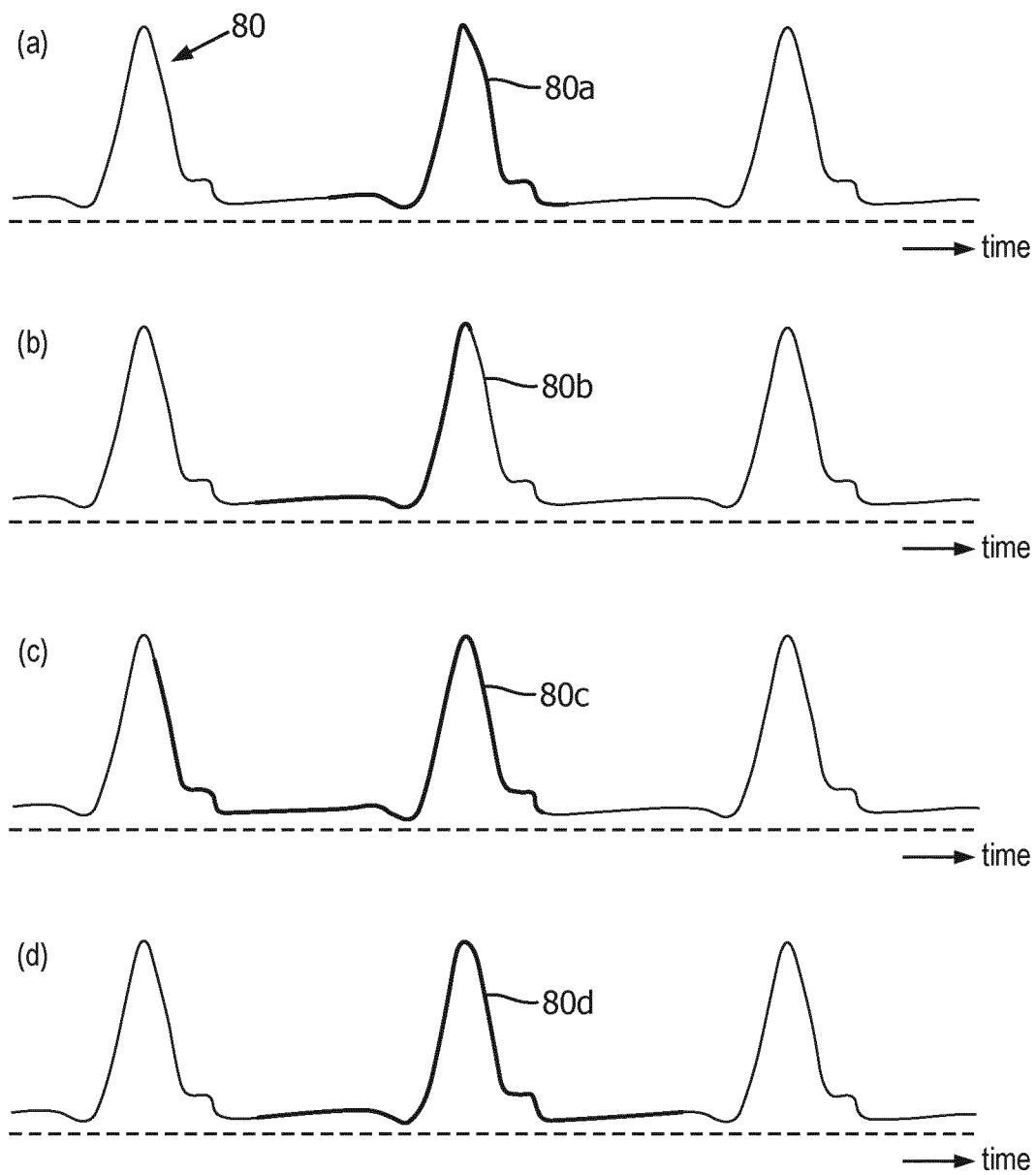


FIG. 4

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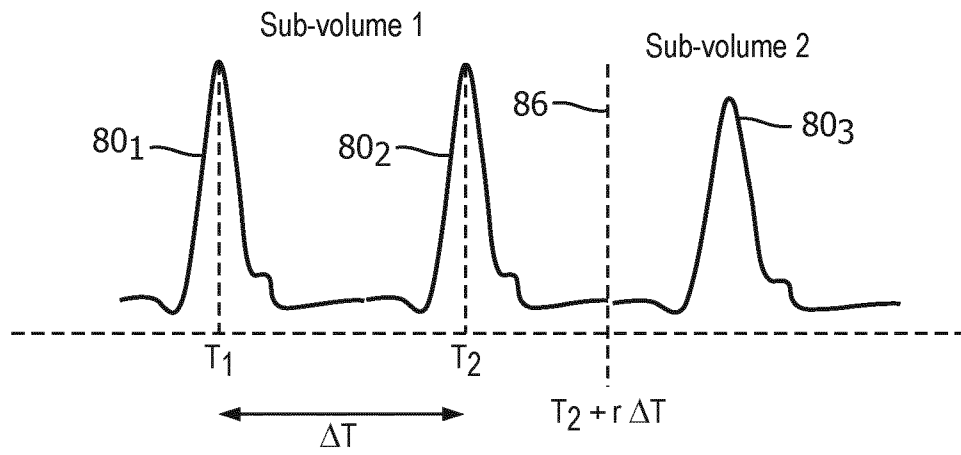


FIG. 5

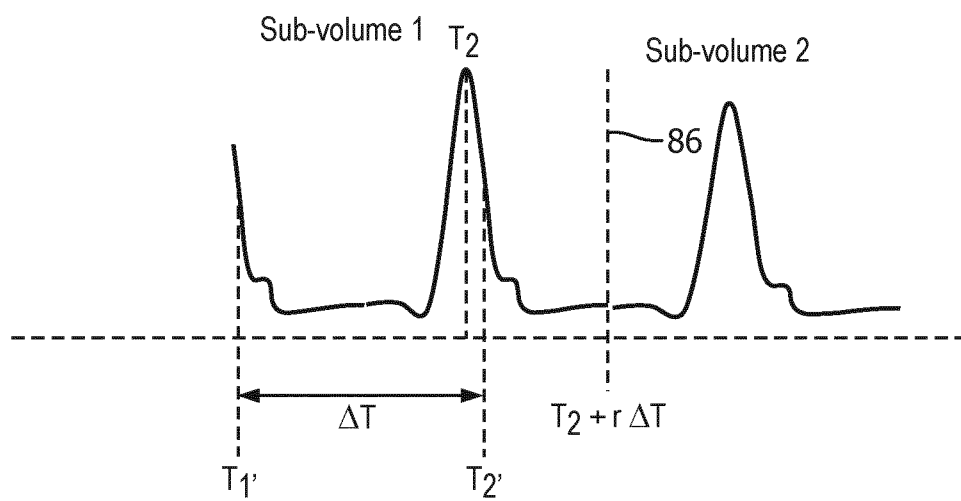


FIG. 6

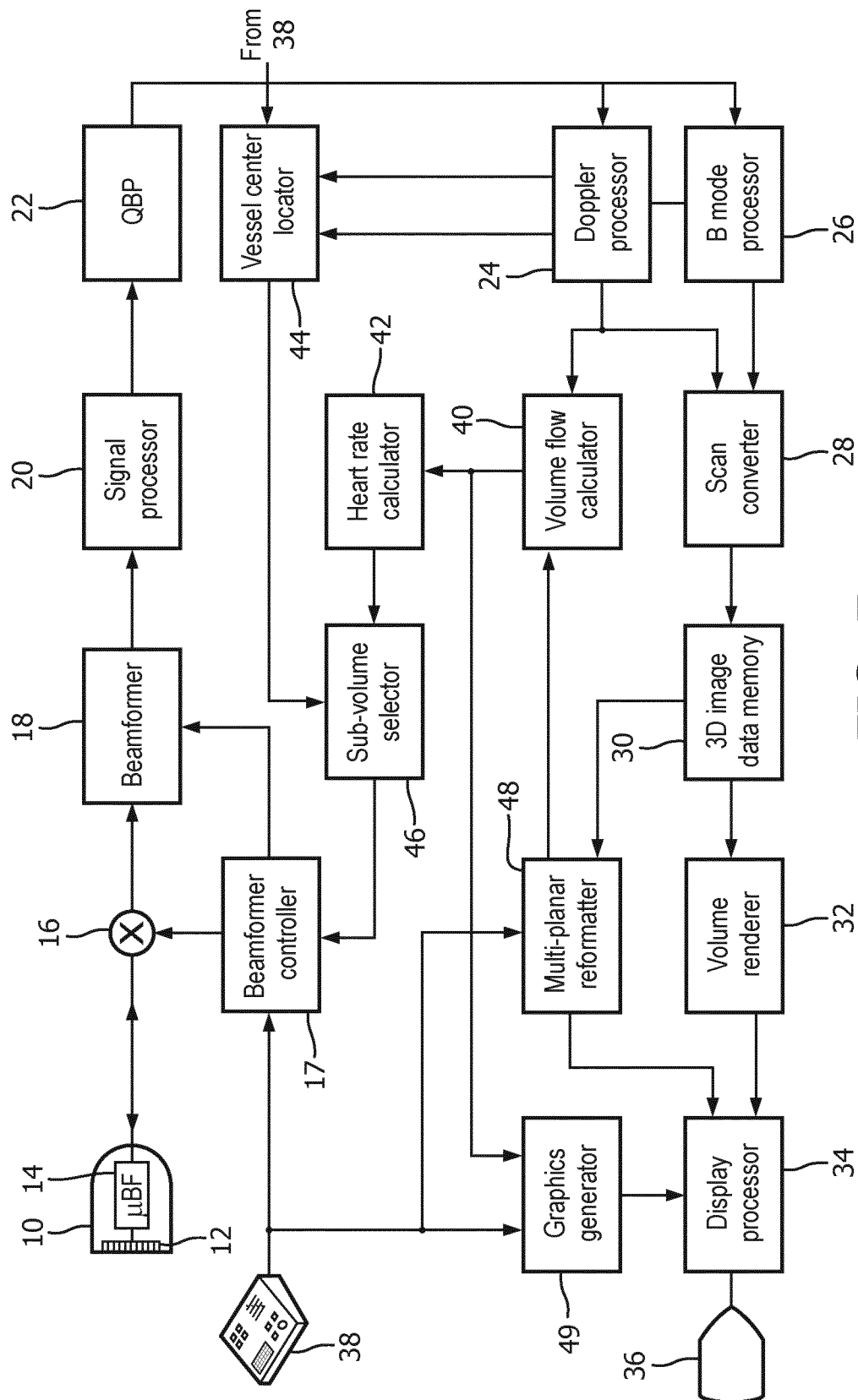


FIG. 7

INTERNATIONAL SEARCH REPORT

International application No
PCT/EP2020/057584

A. CLASSIFICATION OF SUBJECT MATTER INV. A61B8/06 A61B8/08 A61B8/13 A61B8/00 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		
Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	US 2008/287799 A1 (HALL ANNE LINDSAY [US] ET AL) 20 November 2008 (2008-11-20) paragraph [0027] paragraph [0029] paragraph [0042] - paragraph [0045] paragraph [0041] figures 1,15 -----	1,4,9-12
X	US 6 503 202 B1 (HOSSACK JOHN A [US] ET AL) 7 January 2003 (2003-01-07) column 6, lines 38-53 column 11, lines 14-33 column 7, lines 21-31 column 8, lines 26-39 ----- -/-	1-15
☒ Further documents are listed in the continuation of Box C. ☒ See patent family annex.		
* Special categories of cited documents : "A" document defining the general state of the art which is not considered to be of particular relevance "E" earlier application or patent but published on or after the international filing date "L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified) "O" document referring to an oral disclosure, use, exhibition or other means "P" document published prior to the international filing date but later than the priority date claimed "T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention "X" document of particular relevance; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone "Y" document of particular relevance; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art "&" document member of the same patent family		
Date of the actual completion of the international search 18 June 2020		Date of mailing of the international search report 26/06/2020
Name and mailing address of the ISA/ European Patent Office, P.B. 5818 Patentlaan 2 NL - 2280 HV Rijswijk Tel. (+31-70) 340-2040, Fax: (+31-70) 340-3016		Authorized officer De la Hera, Germán

INTERNATIONAL SEARCH REPORT

International application No

PCT/EP2020/057584

C(Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT

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