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(54) Title: ULTRASONIC BLOOD FLOW DOPPLER AUDIO WITH PITCH SHIFTING

Fig. 4

(57) Abstract: An ultrasonic diagnostic imaging system produces audio Doppler from detected Doppler signals. The Doppler signals are detected in a band of frequencies which corresponds to the velocity of blood flow signals, and Doppler information is displayed based on the detected band of frequencies. The audio Doppler system produces Doppler audio in a frequency band which is shifted in pitch from the detected band of fre-quencies. The operator of the ultrasound system is provided with a user control by which the degree of pitch shifting can be controlled. The ultra-sound system displays Doppler blood flow velocities referenced to a trans-mit Doppler frequency f_0, with the audio Doppler being shifted in pitch from the frequencies corresponding to the blood flow velocities.
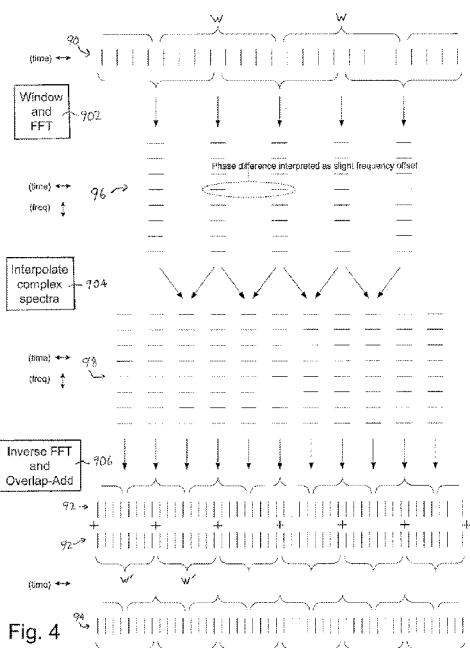
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ULTRASONIC BLOOD FLOW DOPPLER AUDIO
WITH PITCH SHIFTING

This invention relates to medical diagnostic
ultrasound systems and, in particular, to ultrasound
systems which produce Doppler audio which can be
controlled in pitch.

The use of Doppler audio to diagnose blood flow
extends back for many decades. In the years before
real time video presentations of Doppler flow
characteristics were possible, audio was the only
means of ultrasonically diagnosing blood flow. The
clinician would aim the Doppler probe toward the
organ or vessel of interest, unaided by video, and
listen for the characteristic pulsatile "whooshing"
sound of blood flow. Since the Doppler frequency
shift is generally in the kilohertz range or lower,
the amplified Doppler frequency signal could be used
to directly drive an audio loudspeaker. This remains
the way that ultrasonic Doppler signals from blood
flow are produced today. The diagnostic use of audio
Doppler has declined, however, as the live video
presentations of Doppler flow such as spectral
Doppler and colorflow Doppler have provided more
precise and spatially specific ways of ultrasonically
diagnosing blood flow characteristics. Nevertheless,
Doppler audio remains a staple of diagnostic
ultrasound and is still used today to help guide and
confirm proper probe and sample volume placement.
When the visual display viewed by the sonographer and
the Doppler audio heard by the sonographer are both
producing corresponding information, the
sonographer's confidence in the validity and accuracy
of the diagnostic information is reinforced.
The Doppler audio signal is played through a

sound system which is part of the ultrasound system. In cart-mounted systems the loudspeakers of the sound system can be located at a variety of positions on the cart, such as in the system mainframe, on the control panel, or on the display. Since the loudspeakers are carried by the cart and can be positioned in various places, a wide variety of commercially available loudspeakers are available for use by the system designer. But with the recent advent of more compact, hand-carried ultrasound systems, the space for loudspeakers is much more greatly restricted. In compact systems, size and weight are at a premium and the system designer is often restricted to using very small, very compact loudspeakers. In the CX50 compact ultrasound system of Philips Healthcare, for example, the loudspeakers are mounted in the carrying handle, mandating the use of very small, very thin profile speakers. By their nature, small loudspeakers will generally have a higher range of frequency response than larger speakers, an outcome dictated by both their small size and restricted acoustic environment. These speakers are not capable of the higher fidelity low frequency (bass) response characteristic of larger loudspeakers in larger acoustic enclosures.

As indicated above, the loudspeakers of an ultrasound system must reproduce audio sounds of the Doppler signal frequency range. The frequencies of the Doppler signal are proportional to the speed or velocity of blood flow. For relatively higher velocity blood flow, good fidelity reproduction is generally within the capability of the small speakers of a portable system, as the higher frequency sounds from the higher velocity flow are within the reproducible frequency range of the speakers. These

higher blood flow velocities are usually encountered
when diagnosing arterial blood flow. But in venous
flow diagnosis, the blood flow velocities are much
lower. Venous blood flow in the saphenous veins of
the leg may be in the range of only a few centimeters
per second, for instance, or lower. Consequently the
Doppler audio reproduced from these low flow rates
will be low frequencies which are poorly reproduced
by small speakers. The Doppler audio will be of low
volume, garbled, and difficult to comprehend. Hence
it would be desirable to provide better audio
reproduction of the lower frequencies encountered
especially in venous ultrasound exams.

One approach to the problem of low frequency
reproduction has been tried for the tissue Doppler
mode. In tissue Doppler, the motion of tissue, such
as the motion of the myocardium of the heart, is
detected by Doppler processing. Unlike blood flow,
the motion of tissue is the movement of a continuous
section of tissue, in which the tissue cells are
moving in unison since they are physically joined
together. Thus, the tissue motion will be dominated
by the unitary speed of the joined tissue cells and
not the range of velocities produced by turbulent
blood flow. As a result, the Doppler signal will be
characterized by a predominant single frequency at
any short interval of time. The mean frequency value
is used to generate a synthetic sinusoid at the mean
frequency. To reproduce the Doppler signal at a
higher frequency it is only necessary to multiply the
mean frequency by a frequency scaling term to shift
the sinusoid frequency to a higher frequency. A 60
Hz signal can be simply modulated up to 180 Hz, for
instance by multiplying the frequency of the
synthesized sinusoid by three.

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While successful for tissue Doppler, this mean
frequency shifting approach is inadequate for Doppler
audio of blood flow. As mentioned above, the blood
cells in a vein or artery are disassociated and will
move in the blood flow substantially independent of
each other. The flow velocity at the center of a
blood vessels will be greater than the rate along the
vessel walls due to friction at the vessel walls
which is absent in the center of the vessel. Blood
cells can also move in many different directions due
to turbulence caused at obstructions and valves.
Consequently, blood flow is not characterized by a
single velocity as tissue is, but by a multitude of
simultaneous velocities. It is the spectrum of audio
frequencies corresponding to these velocities which
is produced by Doppler audio, which to a trained ear
is rich in harmonics and subtle timbre. Trained
sonographers rely upon this richness of the Doppler
audio sound to guide them in probe and sample volume
placement. Simply shifting the mean frequency of the
Doppler signal to a higher frequency cannot reproduce
the full spectrum of Doppler shift frequencies
arising from blood flow and will produce a monotonic,
artificial sound that is unfamiliar to the trained
sonographer. Consequently a different approach must
be taken to the problem of improving the reproduction
of low frequency venous flow audio Doppler sounds.

    Diagnosis and use of Doppler audio requires
considerable experience, as the nuances of the
complex Doppler sounds can be very subtle and are
contextually comprehended only by trained
sonographers. This challenge is made more difficult
by the fact that the Doppler demodulation frequency
plays a role in the Doppler audio sound.   While
Doppler harmonic imaging (used primarily for tissue

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PW Doppler and contrast imaging) uses a demodulation
frequency twice that of the transmitted frequency so
as to detect the second harmonic of the transmit
frequency, conventionally the Doppler demodulation

5   frequency is set equal to the transmit frequency for
optimal blood-flow detection in the absence of
contrast agents. A probe which transmits and
receives at 3 MHz (a 3 MHz probe) will produce a
different range of Doppler audio than a 5 MHz probe

10   for instance, and the sonographer will select a
different frequency probe for different exams
depending upon factors such as the depth of the blood
vessel to be examined. It would be desirable if this
difference in Doppler frequency bands could be

15   eliminated so that the Doppler audio would be
reproduced at the same range of audio frequencies
regardless of the frequency of the probe.

    In accordance with the principles of the present
invention, a diagnostic ultrasound system is

20   described which produces blood flow Doppler audio at
an audio frequency which is not the Doppler shift
frequency. In one implementation the sonographer is
provided with a user control that enables adjustment
of the pitch of the Doppler audio signal. The

25   control preferably enables the Doppler audio sound to
be shifted in pitch by fractions of an octave or by
an octave or more. This is done, not by simply
shifting the mean frequency of the Doppler audio
band, but by stretching or dilating the entire range

30   of frequencies within the audio band so that the
nuances of the blood flow tones are accurately
preserved and reproduced. While the sounds of the
Doppler audio are shifted in pitch, visual display of
the Doppler blood flow velocities is done at the

35   unshifted Doppler velocity frequencies. The user

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control can cause the Doppler audio signal to be
produced at a higher octave during venous flow exams
with a small speaker ultrasound system, for instance.
In another implementation, pitch control is used to
reproduce audio Doppler in a constant frequency
spectrum, thereby eliminating the difference in
Doppler audio due to the use of different frequency
probes.

In the drawings:

FIGURE 1 illustrates in block diagram form an
ultrasonic diagnostic imaging system constructed in
accordance with the principles of the present
invention.

FIGURE 2 illustrates an ultrasound system
display screen showing a spectral Doppler display and
a 2D image of the sample volume and flow cursor
locations.

FIGURES 3a-3e illustrate the shifting of the
Doppler audio spectrum in accordance with the
principles of the present invention.

FIGURE 4 illustrates a technique for shifting
the pitch of a blood flow Doppler audio which may be
implemented in a sampled data system.

Referring first to FIGURE 1, an ultrasound
system constructed in accordance with the principles
of the present invention is shown in block diagram
form. The array transducer 12 of an ultrasound probe
10 transmits ultrasonic waves and receives ultrasonic
echo signals in response to the transmitted signals.
The elements of the array transducer 12 are shown
transmitting ultrasound beams over a sector-shaped
field of view 100 and along a Doppler beam axis 22.
The transmission of ultrasound by the array
transducer is controlled by a transmit beamformer 62
which controls parameters such as the frequency of

transmission and the timing of transmission by the
individual elements of the array. The elements of
the transducer array convert the received ultrasound
signals to electrical signals which are transmitted
5   by way of a transmit/receive (T/R) switch 26 to a
receive beamformer 64. The receive beamformer 64
forms coherent echo signal samples S(t) from the
signals received from the transducer elements. The
transmit and receive beamformers are synchronized and
10   controlled by a beamformer controller 60.

The coherent echo signals are generally received
along a sequence of beam directions and the echo
signals undergo quadrature detection by a quadrature
bandpass (QBP) filter 28. A typical QBP filter is
15   described in US Pat. 6,050,942 (Rust et al.) The QBP
filter 28 produces quadrature I and Q components for
each echo signal. These components may be amplitude
detected to form grayscale image data by a B mode
processor 30 using the expression $\sqrt{I^2+Q^2}$. The I,Q
20   component pairs are also stored in an ensemble memory
32 in temporally different samples from the same
image field location for Doppler processing. The
ensemble memory facilitates a transform from "fast
time," which is a function of the r.f. sampling rate,
25   to "slow time," which is a function of the pulse
repetition interval (PRI) at which each sample volume
in the image field is interrogated. The data
ensembles are coupled to a wall filter 37 which
removes undesired Doppler shift components. For
30   blood flow imaging the wall filter 37 removes low
frequency components from slow-moving tissue, and for
tissue motion imaging the higher frequency components
of flowing blood are removed. The instantaneous
frequencies at different points in the image field
35   may be color-coded in correspondence with the

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frequencies (which correspond to velocities) and
displayed in a color Doppler display by a colorflow
Doppler processor 38. The color Doppler display
generally overlays a grayscale B mode image for
5   structural orientation of the color-coded motion or
flow. The colorflow Doppler display will give the
clinician a view of the instantaneous flow or motion
over the full image field, dynamically shown in real
time.

10   The ultrasound system of FIGURE 1 is also
capable of continuous wave (CW) Doppler measurements.
In CW Doppler a Doppler signal is transmitted
continuously from one aperture of the array
transducer 12 and echoes are continuously received by
15   another aperture of the transducer array. The echoes
received are those from the overlap of the transmit
and receive beams. The received signals s(t) are
mixed with sine and cosine functions of the transmit
frequency and low pass filtered to remove undesired
20   mixed components, generally the sum frequencies. The
difference frequencies are then image processed and
displayed spectrally in the same manner as a pulsed
wave (PW) spectral Doppler display.

         In accordance with the principles of the present
25   invention the received echo signals S(t) are phase
demodulated to the Doppler shift band by a gated
Doppler demodulator 34. The gating demodulates
echoes returned from a sample volume location which
may be provided by the Doppler beam vector and sample
30   volume gating signal discussed below. The
demodulated Doppler signals I_0, Q_0 are referenced to
the transmit Doppler frequency f_0 and are of the form
$I(t)+jQ(t)$, sometimes referred to as the "analytic
signal." The demodulated Doppler signals are
35   filtered by a wall filter 35 to remove tissue

components and pass only blood flow components. The
wall filtered blood flow Doppler signals are of the
form I_0', Q_0'. These blood flow signals are then
processed for spectral display by a spectral Doppler

5   processor 36. For spectral Doppler display the
spectral Doppler processor will produce a spectrum of
frequency values which correspond to the range of
frequencies of blood flow that exist at the time of
the measurement. Each time-sequential spectrum is

10   displayed as a spectral line in a (generally
scrolling or sweeping) spectral display by the
spectral Doppler processor 36 as shown in FIGURES 2
and 3. The spectral Doppler display will give the
clinician a detailed quantification of the flow or

15   motion components at a specific sample volume in the
image field.
              The grayscale (B mode) image data, the spectral
Doppler data, the CW Doppler data, and the colorflow
data are all coupled to an image processor 40 for the

20   production of one or more images in the desired image
format(s) on a display 24.
              In accordance with the principles of the present
invention the ultrasound system of FIGURE 1 also
produces an audio Doppler signal from a loudspeaker

25   44. Conventionally the audio Doppler signal is at
Doppler-shifted frequencies based on the transmitted
Doppler frequency f_0 and produced as an audio signal
since the Doppler-shifted frequencies are in the
human audible spectrum, typically 100Hz to 10kHz.

30   When the Doppler audio signal is formed digitally, it
is converted to an analog signal by a digital to
analog converter (DAC) 42, amplified, and applied to
the loudspeaker 44. The audio signals may also be
separated as a function of the flow direction with

35   respect to the transducer array by a forward/reverse

separator 54, which will reproduce signals from flow
toward the transducer array through one speaker and
flow away from the transducer array through a second
speaker. The flow direction is conveniently given by
the sign of the Doppler shift. In addition, the
ultrasound system of FIGURE 1 enables the user to
control the pitch of the audio Doppler signal for
better fidelity. For example, the loudspeaker 44 may
have poor reproduction fidelity of low frequency
Doppler signals from low velocity venous blood flow.
In that case, the user would increase the pitch of
the reproduced Doppler sounds to a frequency range
which is more clearly reproduced by the loudspeaker.

In the example of FIGURE 1 the ultrasound system
has a control panel 20 by which the user can control
a number of the Doppler features of the ultrasound
system. It will be appreciated that the control
panel can be constructed in hardware or as softkeys
on a display screen or a combination of the two. A
user control can be manipulated by the user to steer
a Doppler beam vector 22 over the image field. The
vector 22 is graphically produced as a line over the
ultrasound image and is the indicator to the
ultrasound system of the direction that the Doppler
beam is to be transmitted for spectral Doppler
interrogation. The user can also manipulate a
control of the control panel to position a sample
volume graphic 16 (see FIGURE 2) at the depth along
the Doppler beam where spectral Doppler data is to be
acquired. The Doppler beam vector and the sample
volume location are coupled to the beamformer
controller 60 to inform the controller of the
direction in which the Doppler beam is to be
transmitted and the depth along the beam where
spectral Doppler measurements are to be made. Gating

of the Doppler demodulator 34 may be set by the same timing signals. In addition, the user can manipulate a control of the control panel 20 to indicate the direction of blood flow in a blood vessel with a flow

5   cursor. This cursor setting is used by the ultrasound system to correct the Doppler frequency estimate, as the Doppler equation used to estimate the Doppler frequency is angle-dependent upon the angle between the Doppler beam direction and the

10   direction of the blood flow. The beamformer controller also controls Doppler transmission to be at a nominal Doppler transmit frequency of the transducer array, f_0. This information is typically supplied to the ultrasound system by a memory device

15   in the probe when the probe is connected to the ultrasound system as described in US Pat. 4,868,476 (Respaut). In Doppler operation the probe 10 will transmit the Doppler beam at this nominal transmit frequency and the Doppler shift or frequency will be

20   offset from this nominal frequency.

In the example of FIGURE 1, the nominal Doppler transmit frequency f_0, the Doppler beam vector setting, and the flow cursor setting are coupled to a pitch controller 52. In addition the pitch

25   controller receives a pitch parameter m which is set by the user from the control panel. The pitch controller is then able to calculate the angle between the Doppler beam vector and the flow cursor and provide this angle in the form of cosΦ to the

30   spectral Doppler processor 36, which uses this term for angle correction of the Doppler estimates. The pitch controller 52 also uses the data it receives to calculate a pitch scaling factor K for pitch control. The factor K and the Doppler frequency f_D from the

35   Doppler frequency estimator 34 are applied to a phase

vocoder 50 for control of the pitch of the Doppler audio signal. Phase vocoders have been used in the past for speech synthesis and music editing. However in the ultrasound system of FIGURE 1 the phase

5   vocoder 50 is used for Doppler audio by producing pitch-shifted signal components of the form I_S, Q_S, which are used by the forward/reverse separator 54 and the DAC 42 to drive the loudspeaker(s) 44 with a Doppler audio sound different from that which is

10   based on the transmitted Doppler frequency f_o.

A conventional Doppler ultrasonic imaging system produces a broadband audio signal whose frequency spectrum is related to the velocities of moving scatterers within a region of interest in the body

15   through the Doppler equation. That is, the intensity of the audio signal at each audio frequency is proportional to the sum of the intensities of the acquired ultrasonic echoes from all scatterers moving at a velocity v, where f_D and v are related through

20   the Doppler equation as:

$$f_D = \frac{2vf_0 \cos\phi}{c}$$

where f_0 is the ultrasonic demodulation center frequency (conventionally equal to the transmit

25   frequency,) Φ is the Doppler angle, the angle between the Doppler beam direction and the direction of blood flow, and c is the speed of sound. Normally, once the Doppler frequency is estimated, the Doppler equation is used to calculate the blood flow

30   velocity. However, in an implementation of the present invention, the frequencies of all components of the Doppler audio signal are scaled such that the intensity of the frequency-scaled audio signal at each audio frequency, $f_{D'}$, is now proportional to the

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sum of the intensities of the acquired ultrasonic
echoes from all scatterers moving at a velocity v,
where $f_{D'}$ and v are related through the following
equation:

$$f_{D'} = K \frac{2vf_0 \cos\phi}{c}$$

where k typically is in the range of 1.0 <= k <= 4.

In the ultrasound system of FIGURE 1, the phase
vocoder 50 uses the quadrature Doppler components
I_0', Q_0' used for spectral Doppler display and the
factor K supplied by the pitch controller 52 to
produce a new signal for Doppler audio reproduction
in accordance with the equation:

$$f_{D'} = K \frac{2vf_0 \cos\phi}{c}$$

The factor K in the equation will shift the pitch of
the f_D Doppler frequency band to produce a Doppler
audio signal $f_{D'}$ with a different pitch set by the
pitch scaling factor K. The pitch scaling factor is
set by the user's adjustment of the variable term m,
where $K = f(m)$. In a constructed embodiment the term
m can be sequenced over a range of values with each
discrete value producing a one-third octave shift of
the Doppler audio sound. Six values are used so that
the Doppler audio sound can be increased in pitch by
up to two octaves. Adjustment of the user control
for Doppler audio pitch change will not affect the
Doppler shift frequency values used for the visual
spectral and colorflow Doppler displays, which
produce their visual information using the unaltered
f_D Doppler frequency.

The pitch controller 52 and the phase vocoder 50
can be used in other implementations to provide other
benefits. For instance, as mentioned above, since
the Doppler equation contains the term f_0, the nominal
transmit Doppler frequency, the Doppler sound will

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depend upon the frequency of the particular probe used for the exam. A 3MHz probe will produce a lower frequency sound than will a 5MHz probe. The sonographer may have a discerning ear for 5MHz Doppler audio, for example, and may want the Doppler sound to be referenced to 5MHz regardless of the probe or Doppler transmit frequency which is used for the exam. This can be done by having the pitch controller 52 set K equal to:

$$K = \frac{5\text{MHz}}{f_0}$$

when the factor K is calculated in this manner, the Doppler equation with the K factor becomes:

$$f_{D'} = \frac{2v(5\text{MHz})\cos\phi}{c}$$

The factor K thus causes probe-dependent frequency term f_0 to be eliminated and the frequency $f_{D'}$ is always a function of a fixed 5MHz. Thus, the Doppler audio will always sound like that of a 5MHz probe. Doppler audio with a consistent pitch is produced for Doppler probes of different Doppler frequencies.

In another implementation the factor K may be calculated by the pitch controller 52 to produce Doppler audio sound which is invariant with changes in the Doppler angle. This may be done by calculating K to be:

$$K = \frac{1}{\cos\phi}$$

When this K factor is used in the Doppler equation the Doppler angle is removed from the calculation of the Doppler audio frequency $f_{D'}$.

It will also be appreciated that the two concepts above could be merged to make the Doppler sound invariant to both changes in the Doppler angle and the frequency of the probe used. Embodiments of the present invention can shift the pitch of the

audio Doppler signal, without any alteration of the transmitted ultrasonic frequency or the Doppler demodulation frequency.

A typical Doppler display presented on the display 24 for simultaneous colorflow and spectral Doppler interrogation is shown in FIGURE 2. In this display the upper sector image 100 contains the B mode structural display overlaid with color information showing the blood flow. The transducer array is positioned at the apex 14 to acquire the image. The blood flow through the vessel passing through the center of the sector image 100 may continually change from red to blue with the pulsatile change in the blood flow velocity, for example. The user can position a Doppler beam direction line 22 over the image 100 with a user control to intersect the blood vessel at a point where a spectral Doppler measurement is to be made. The sample volume graphic 16 is then moved up and down the line 22 with a user control until it is positioned over the blood vessel where the spectral Doppler measurement is to be made. The tilt of the flow cursor 18 is then adjusted by a user control to indicate the direction of blood flow, and the angle between the Doppler beam line 22 and the flow cursor 18 is the Doppler angle for angle correction. The resultant spectral Doppler display 120 of the blood flow at the sample volume 16 is shown at the bottom of the display screen.

The concept of shifting the pitch of the Doppler audio spectrum may be appreciated by referring to FIGURES 3a-3e which show a sequence of spectral lines of a portion 70 of a spectral Doppler display. Each spectral line of the display such as spectral line 70 is comprises of a sequence of data points along the

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line. The position of a data point on the line
represents a velocity value as indicated by the ±V
scale to the left of the spectral lines, and each
data point has a magnitude which is a function of the
5   content of the frequency bin which provided that
velocity. This spectral information may be
represented in both frequency and magnitude by the
spectral band 80 drawn to the left of the spectral
display in FIGURE 3b. As the curve 80 illustrating
10   the envelope of the spectral band indicates, the
spectrum has a particular shape determined by the
range of velocities of blood cells and by the
presence of predominate velocities which are
represented by increased amplitude of the curve (to
15   the left in the drawings).

A simple way to change the Doppler sound for a
higher frequency response loudspeaker is simply to
shift the frequency of the spectrum 80 to a higher
frequency band 80' as shown in FIGURE 3c. The sound
20   will now be reproduced at the higher frequencies of
the shifted band 80'. But this simple frequency
shift will not reproduce the timbre of the sound.
The higher frequency sound will be metallic and
inharmonic and will sound dissonant to the ear of the
25   trained sonographer. To prevent this an
implementation of the present invention will shift
the pitch of the sound, changing the spectrum 80 of
FIGURE 3b to the pitch-shifted spectrum 82 of FIGURE
3d. In this new spectrum the frequency components
30   have been stretched or dilated, and harmonic
relationships retained. It is seen that the shape of
the spectral envelope is preserved but stretched.
Thus, a 100Hz component is scaled to 200Hz, a 200Hz
component is scaled to 400Hz, a 300Hz component is
35   scaled to 600Hz, and so on, which preserves the

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harmonic content and the timbre of the sound. This
effect is shown in a normalized logarithmic frequency
scale in FIGURE 3e, where the conventional audio
spectrum Sa is shifted by one octave to a pitch-

5   shifted spectrum Sa'. It is seen that the spectrum
appears unaltered in shape, but is shifted up in
pitch by one octave. The pitch-shifted Doppler audio
will appear to the sonographer to be the same Doppler
sound as before, but at a higher pitch. And when the

10   higher pitch is better aligned with the passband of
the loudspeaker, a clearer and more distinct Doppler
audio sound is produced by the ultrasound system.

       There are a number of techniques for shifting
the pitch of the Doppler audio signal such as time

15   domain harmonic scaling, wavelet processing, and use
of a phase vocoder. The phase vocoder approach is
preferred as phase vocoders are well understood in
other applications and a phase vocoder can be
implemented with overlapping short-time FFT

20   processing, which is a common approach for Doppler
spectral signal processing. The phase vocoder, which
can be implemented in hardware or algorithmically by
software processing, models the audio signal as a set
of narrowband tones, one for each FFT frequency bin.

25   The change of magnitude of an FFT bin between
successive overlapping FFTs is interpreted as a
gradual change of amplitude of the narrowband tone
over that short span of time. The change of phase of
an FFT bin between successive overlapping FFTs is

30   interpreted as a precise frequency adjustment of the
narrowband tone over that short span of time, a
perturbation from the center frequency of the FFT
bin. Each frequency-adjusted tone is then used to
interpolate the corresponding FFT bin in time with

35   intermediate magnitude and phase values, producing

more overlapping FFTs for the same original time
span. These overlapping values are processed by
inverse-FFT processing and overlapped-added (with the
original overlap factor) in the usual way of short-
time FFT reconstruction, producing more samples than
in the original sampled signal. If replayed at the
original sample rate, this would produce time-
stretched audio at the original pitch. But if
replayed at a faster sample rate to match the
original time span, the pitch is scaled up by the FFT
interpolation factor. The pitch or time can be
scaled down by decimating instead of interpolating
the FFTs.

    Phase vocoder processing is illustrated by
FIGURE 4 as one possible implementation of pitch
shifting. In this example the frequencies will be
doubled, shifting the pitch up one octave. The small
vertical lines 90 represent a sequence of Doppler
audio samples before pitch shifting, with time
progressing in the horizontal direction and the
spacing between the samples representing the sample
rate. The sequence of samples can be segmented in
overlapping windows W as shown by the brackets above
and below the sequence of samples 90. In this
illustration each window contains eight samples and
is 50% overlapped with adjacent windows, although
windows with greater or fewer samples and more or
less overlap can be used.

    Each segmented window of time samples is
multiplied by a smoothly tapered window function,
then converted to frequency domain with an FFT (Fast
Fourier Transform) in step 902. The smoothly tapered
window function is not shown in the figure, but is a
standard step in FFT processing to reduce time
discontinuity and frequency smearing caused by the

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FFT treating the time segment as a periodic function.
The result of the FFT of successive windows is a time
sequence of frequency domain complex spectra,
represented by the vertical sets of small horizontal
lines 96 in the figure. Each sample (called a "bin")
of each spectrum is a complex number, whose magnitude
and phase correspond to a windowed sinusoid at the
center frequency of the bin. Considering the same
frequency bin in two successive complex spectra, the
difference in phase over that increment of time can
be interpreted as a slight frequency offset from the
center frequency of a bin, since frequency is the
time derivative of phase. In other words, the center
frequency (phase rate) of the bin produces a
deterministic large change of phase over the time
interval between spectra, and the actual phase
difference between the spectra is used as a slight
adjustment to this phase rate. Phase calculations
are modulo 2π radians (360 degrees).

Using the slightly adjusted frequencies and the
magnitudes for each bin in successive spectra,
additional complex spectra 98 are interpolated as
shown in the figure and indicated by step 904. In
this example, the number of spectra over a span of
time is doubled, corresponding to an audio frequency
scaling of two, which is a one octave increase in
pitch. The complex spectra are then converted back
to the time domain with an inverse FFT, producing
segmented windows W' of time samples 92, which are
added together with the same overlap fraction (50% in
this example) as in the earlier window and FFT
processing step 902. This is illustrated in the
figure with the small vertical lines 92 and brackets
W'. Since the interpolation between spectra produced
more spectra per unit of time, after the inverse FFT

and overlap add there are more samples per unit of time. The frequency content of the data is equivalent to the original audio signal in terms of samples, but the sample rate now higher (doubled in this example). So when the data are converted to a continuous analog signal with the new sample rate and applied to a loudspeaker, all of the frequencies are scaled by the ratio of new to original sample rates.

The frequency scale factor is not constrained to an integer, because the complex spectra can be resampled to any rate. For the time samples to align in the overlap-add step 906 following the inverse FFT, the scaling factor should be a ratio of integers, where the denominator is the number of samples in the window. However, even this constraint can be easily overcome by multiplying each interpolated frequency spectrum with a linear phase ramp corresponding to its fractional sample time shift. So there is essentially no constraint on the quantization of frequency scale factor.

The primary trade-off in the phase vocoder processing is the FFT size. Many small FFTs can better follow temporal transients, but only have a few frequency bins, which can be audible as distinct tones. Few large FFTs can better estimate a continuous frequency spectrum, but have a slow temporal evolution that can be audible as phase smearing. The artifacts become more noticeable or objectionable as the pitch scale factor increases. Thus, the preferred embodiment uses medium FFT sizes corresponding to about 20-30 msec.

If the sample sequence 94 of FIGURE 4 were played at the original sample rate of FIGURE 4 through the DAC 42 and loudspeaker 44, the audio signal would be a time-stretched version of the audio

signal of the sequence 90 of FIGURE 4 at the same
pitch as the FIGURE 4 sequence. But when the
sequence 94 is played at a higher sample rate so that
each window at the bottom of FIGURE 4 is played in
5   the same time interval as a window at the top of
FIGURE 4, the pitch is scaled up. In the illustrated
example where the number of samples in each window is
doubled, each window of samples in FIGURE 4 is then
played at twice the sample rate of FIGURE 4 and the
10   pitch is scaled up by a full octave. If the FFT
processing is used to decimate the sample sequence
instead of upsampling it, the pitch would be scaled
down instead of up.

It is seen in FIGURE 4 that there is a ratio of
15   the number of samples 90 in the windows W of the
original sequence to the number of samples 94 in the
windows of the final sequence at the bottom of FIGURE
4. This ratio can be expressed as a ratio of
integers. In this example the ratio is 8:16 or 1:2.
20   The user control of the control panel which adjusts
the pitch of the Doppler audio can provide one of
these integers, m.

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WHAT IS CLAIMED IS:

    1.   An ultrasonic diagnostic imaging system
which produces a Doppler audio signal of blood flow
5   comprising:
        an ultrasound probe, operating at an ultrasonic
Doppler transmit frequency f_0, which acquires Doppler
ultrasound signals referenced to the Doppler transmit
frequency from a location of blood flow;
10   a Doppler demodulator, responsive to the Doppler
ultrasound signals, which produces Doppler shift
signals from the velocity of blood flow in an audio
frequency band;
        a Doppler information display, responsive to
15 Doppler shift signals, which displays blood flow
velocity information based upon the Doppler shift;
        an audio Doppler system, responsive to the
Doppler shift signals, which produces pitch-shifted
audio Doppler without changing the displayed blood
20 flow velocity.

    2.   The ultrasonic diagnostic imaging system of
Claim 1, wherein the audio Doppler system is
operative to produces pitch-shifted audio Doppler
25 without change in the ultrasonic transmit frequency f_0
or the displayed blood flow velocity.

    3.   The ultrasonic diagnostic imaging system of
Claim 2, further comprising a user control, coupled
30 to the audio Doppler system, by which the user
controls the pitch shift of the audio Doppler.

    4.   The ultrasonic diagnostic imaging system of
Claim 3, wherein the user control is operable to
35 control the scaling of the frequency band for audio

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Doppler.

5.   The ultrasonic diagnostic imaging system of Claim 3, wherein the audio Doppler system is responsive to the user control to shift the pitch of the Doppler-shifted signals by a fractional or integer number of octaves.

6.   The ultrasonic diagnostic imaging system of Claim 2, wherein the audio Doppler system further comprises a phase vocoder which shifts the pitch of the Doppler-shifted signals.

7.   The ultrasonic diagnostic imaging system of Claim 3, wherein the audio Doppler system further comprises a pitch controller, responsive to the user control, which produces a pitch shift factor K for control of the audio pitch shift.

8.   The ultrasonic diagnostic imaging system of Claim 1, wherein the audio Doppler system further comprises a loudspeaker exhibiting a given frequency response; and
      wherein the shift in pitch is determined by the frequency response of the loudspeaker.

9.   The ultrasonic diagnostic imaging system of Claim 1, wherein the Doppler demodulator is operative to produce demodulated Doppler signals in a band exhibiting a demodulation reference frequency, which band is determined by a Doppler equation,
      wherein the pitch-shifted audio Doppler band is different from the band determined by the Doppler equation and the demodulation reference frequency.

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10. The ultrasonic diagnostic imaging system of Claim 9, wherein the Doppler equation is of the form

$$f_D = \frac{2vf_0 \cos\phi}{c}.$$

5   11. The ultrasonic diagnostic imaging system of Claim 1, wherein the imaging system is operable with a selected one of a plurality of ultrasound probes, each exhibiting a different nominal Doppler transmit frequency, and each of which is operable to acquire

10   Doppler ultrasound signals from a location of blood flow;
       wherein the Doppler demodulator is further operable to detect Doppler shift signals in a frequency band which is a function of the nominal

15   Doppler transmit frequency of a selected ultrasound probe; and
       wherein the audio Doppler system is responsive to the detected Doppler shift signals to produce audio Doppler related to a common nominal frequency

20   for probes of different nominal transmit frequencies.

       12. The ultrasonic diagnostic imaging system of Claim 11, wherein each of the ultrasound probes exhibits a different nominal Doppler frequency f_0;

25   wherein the Doppler demodulator is operable to detect Doppler signals by a Doppler equation which is a function of f_0 of a selected probe; and
       wherein the audio Doppler system produces audio Doppler related to a selected f_0 frequency for probes

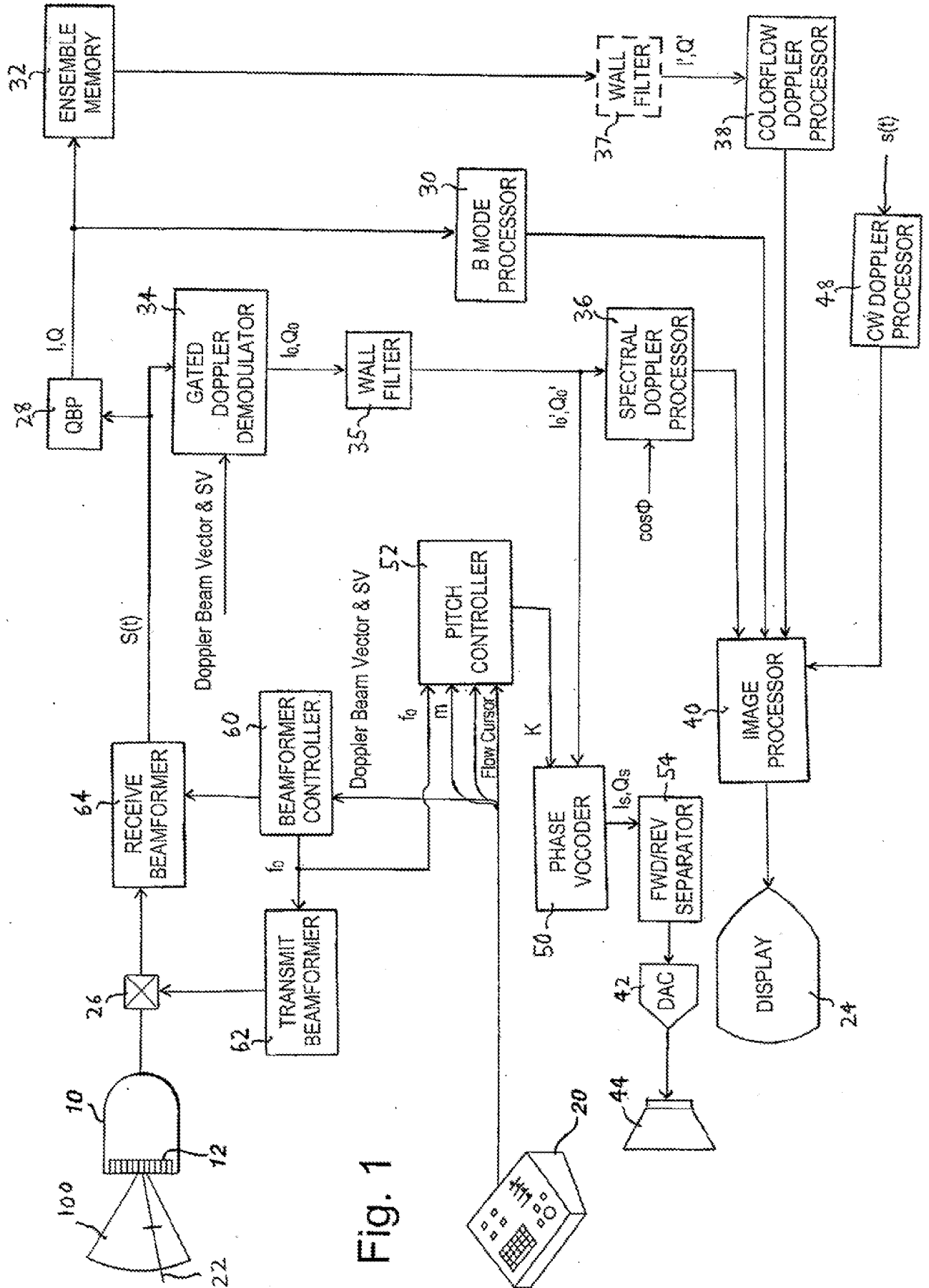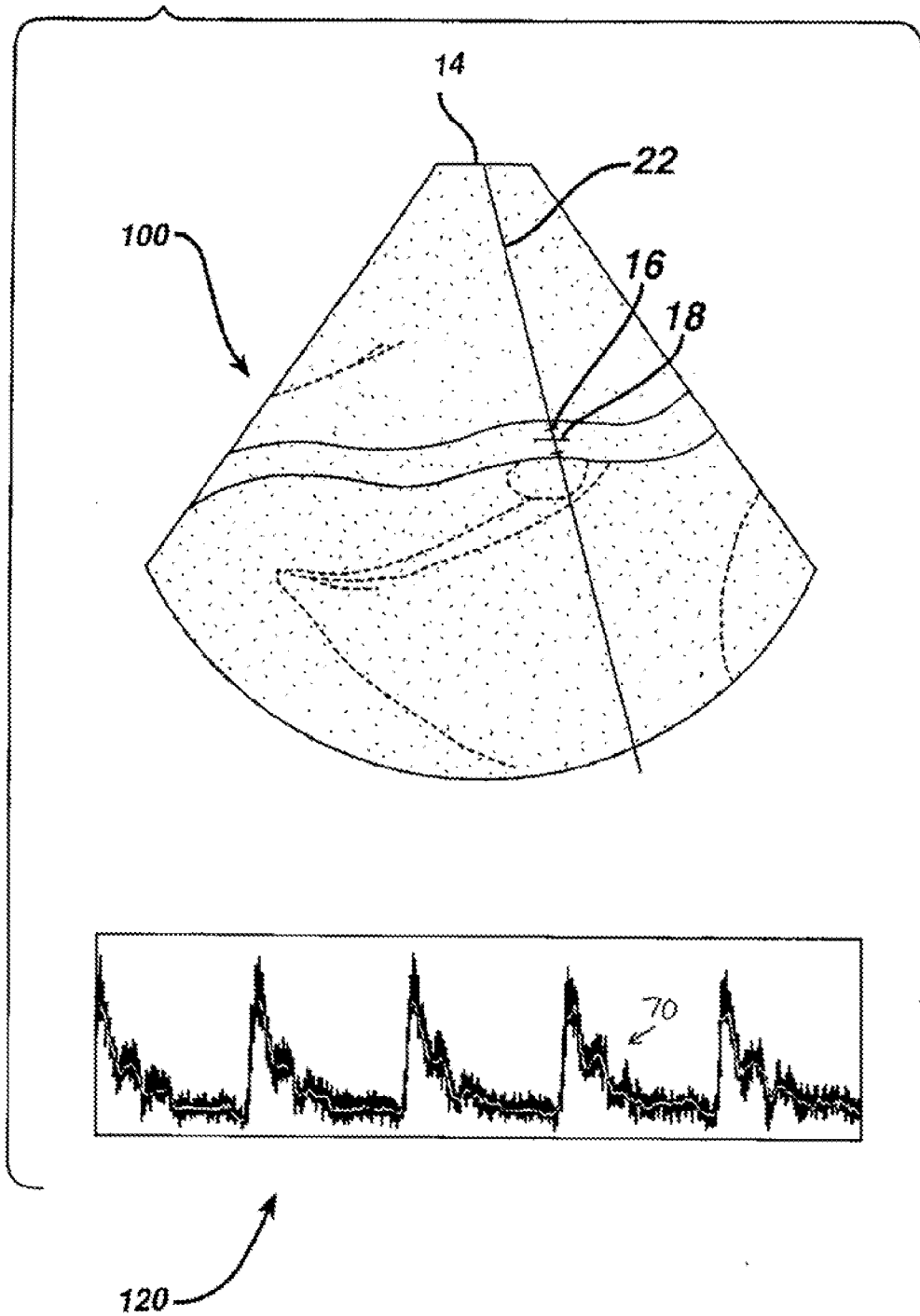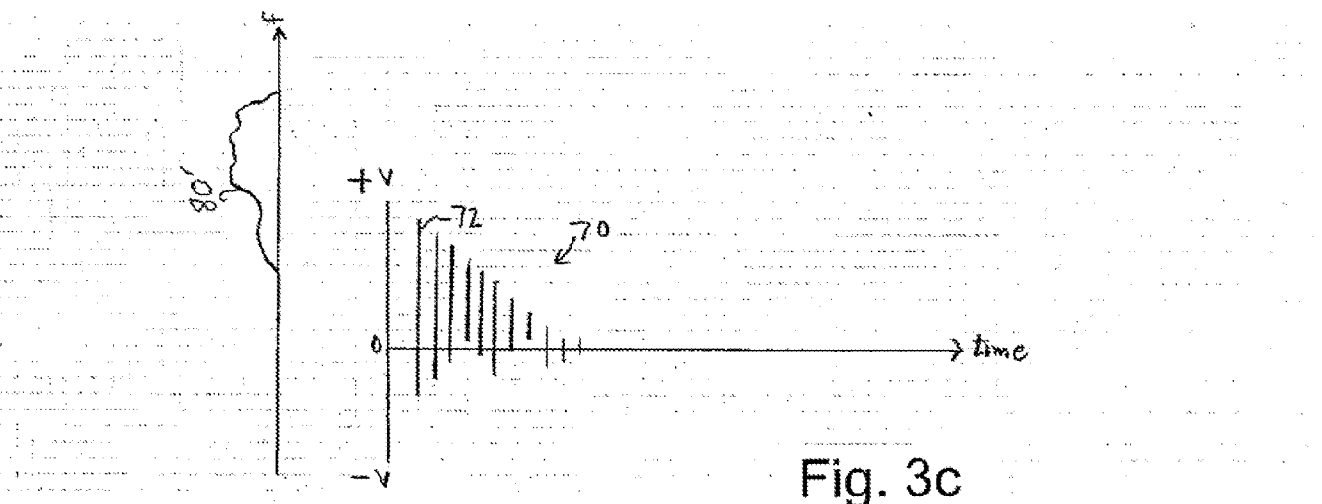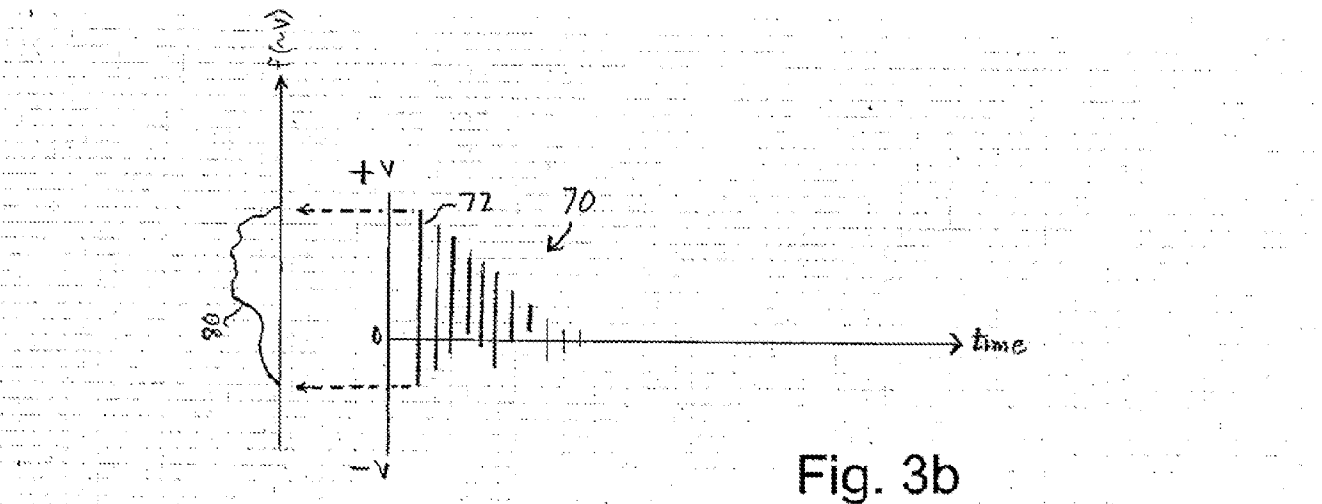30   of different nominal Doppler frequencies.

Fig. 1

Fig. 2
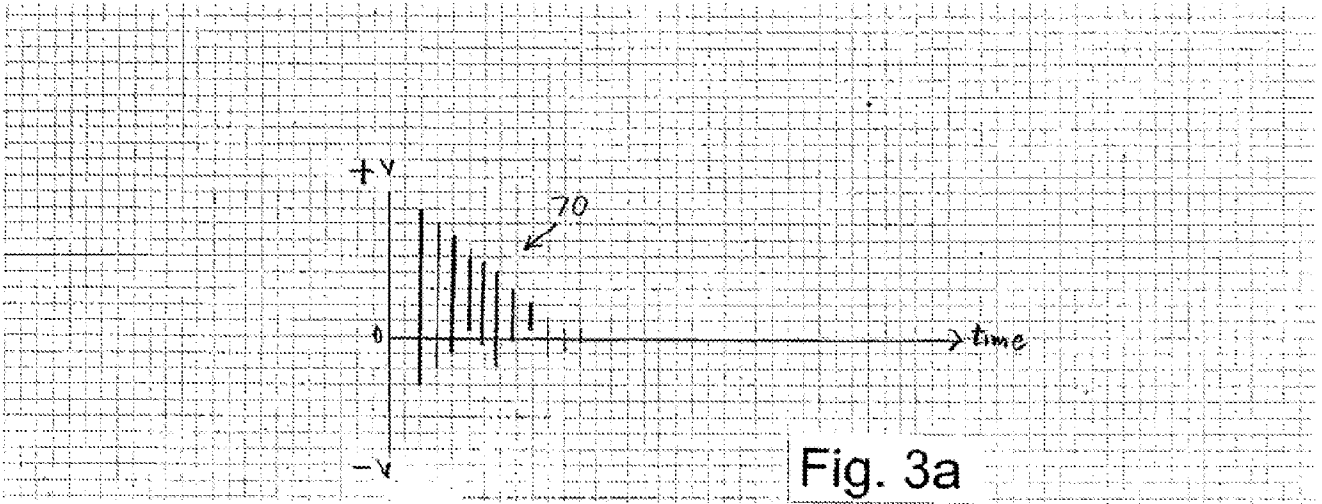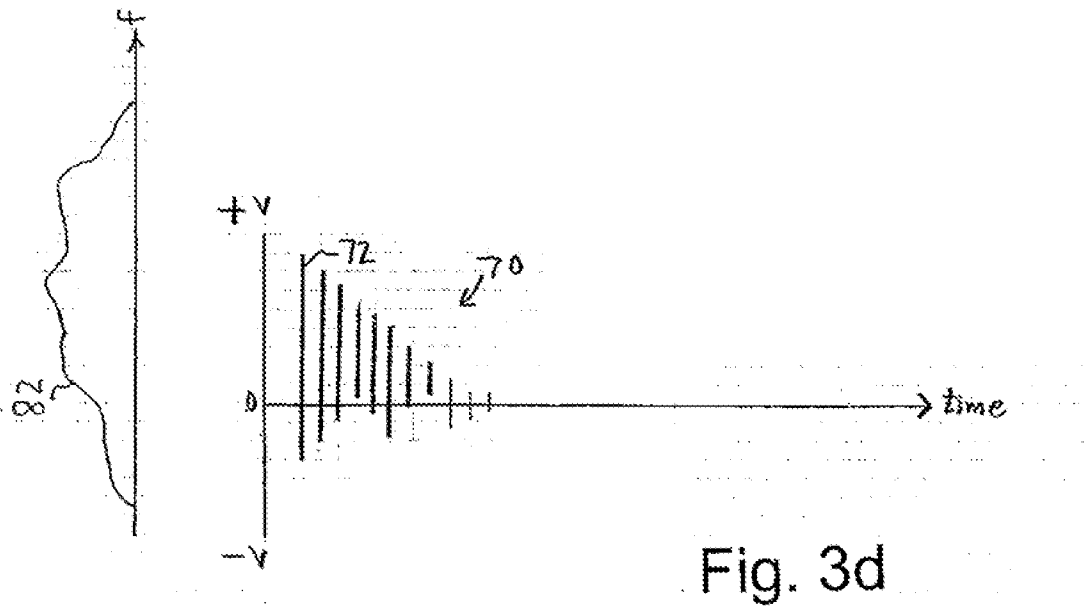
Fig. 3a
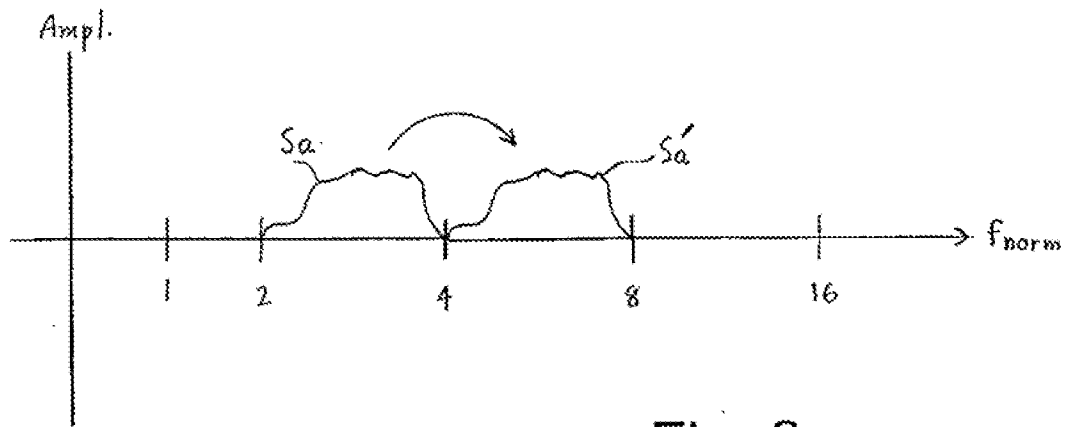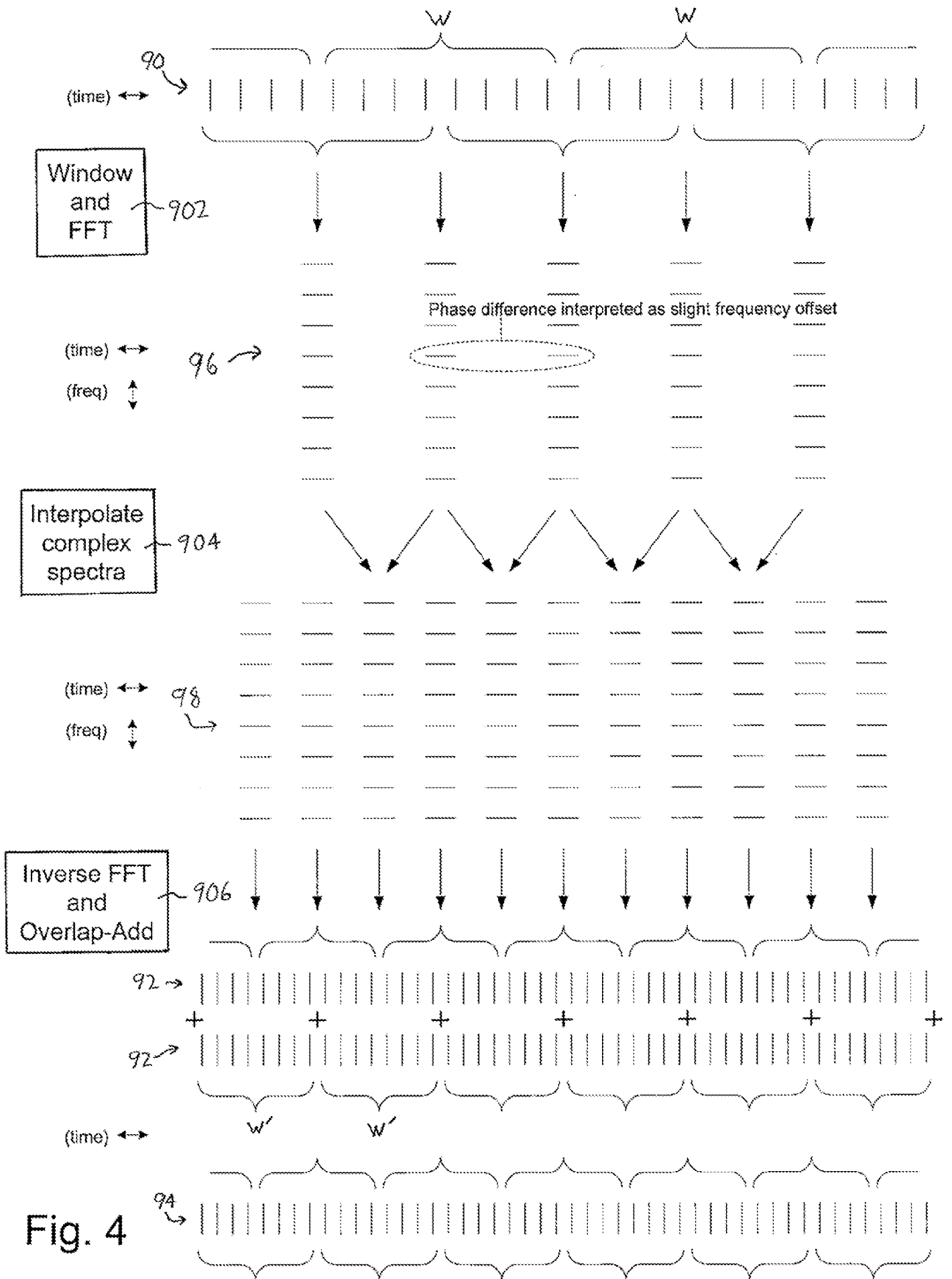
Fig. 3b

Fig. 3c

Fig. 3d

Fig. 3e

W   W

90
(time) ←→

┌──────────┐
│ Window   │
│ and   │──902
│ FFT   │
└──────────┘

Phase difference interpreted as slight frequency offset

(time) ←→   96 →

(freq) ↕

┌──────────┐
│Interpolate│
│ complex │──904
│ spectra │
└──────────┘

(time) ←→   98 →

(freq) ↕

┌──────────┐
│Inverse FFT│
│ and   │──906
│Overlap-Add│
└──────────┘

92 →

+   +   +   +   +   +   +

92 →

(time) ←→

W'   W'

94 →

Fig. 4

INTERNATIONAL SEARCH REPORT

A. CLASSIFICATION OF SUBJECT MATTER

INV. A61B8/06   A61B8/08   G01S15/89
ADD.

According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

A61B G01S

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

Electronic data base consulted during the international search (name of data base and, where practical, 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	WO 2007/023438 A2 (KONINKL PHILIPS ELECTRONICS NV [NL]; CLARK DAVID W [US]) 1 March 2007 (2007-03-01) the whole document	1-12

☐ 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 document 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	Date of mailing of the international search report
29 June 2010	06/07/2010

Name and mailing address of the ISA/	Authorized officer
European Patent Office, P.B. 5818 Patentlaan 2 NL – 2280 HV Rijswijk Tel. (+31-70) 340-2040, Fax: (+31-70) 340-3016	Willig, Hendrik

Form PCT/ISA/210 (second sheet) (April 2005)

International application No
PCT/IB2010/051712

Patent document cited in search report		Publication date	Patent family member(s)		Publication date
WO 2007023438	A2	01-03-2007	EP	1922563 A2	21-05-2008
			US	2009131791 A1	21-05-2009