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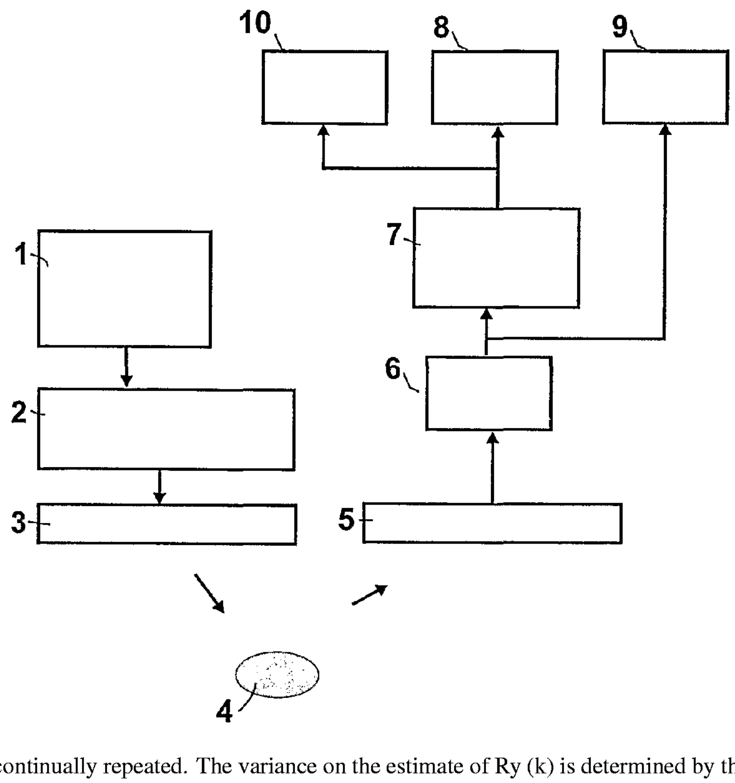
(54) **Title:** SPECTRAL VELOCITY ESTIMATION USING AUTOCORRELATION FUNCTIONS FOR SPARSE DATA SETS



(57) **Abstract:** The distribution of velocities of blood or tissue is displayed using ultrasound scanners by finding the power spectrum of the received signal. This is currently done by making a Fourier transform of the received signal and then showing spectra in an M-mode display. It is desired to show a B-mode image for orientation, and data for this has to acquired interleaved with the flow data. The power spectrum can be calculated from the Fourier transform of the autocorrelation function Ry (k), where its span of lags k is given by the number of emission N in the data segment for velocity estimation. The lag corresponds to the difference in pulse number, so that for lag k data from emission i is correlated with i + k. The autocorrelation for lag k can be averaged over N-k pairs of emissions. It is possible to calculate Ry (k) for a sparse set of emissions, as long as all combinations of emissions cover all lags in Ry (k). A sparse set of emissions inter-spaced with B-mode emissions can, therefore, be used for estimating Ry (k) The sequence ‘v B v v B! gives 2 B-mode emissions (B) for every 3 velocity emissions (v) and is denoted a 3:2 sequence. All combinations on lags are present k=‘0123..!, if the sequence is continually repeated. The variance on the estimate of Ry (k) is determined by the number of emission pairs for the value of k, and it can be lowered by averaging the RF data over the range gate. Many other sequences can be devised with this property giving 3:3, 3:4, and 5:8 or even random sequences, so that the ratio between B-mode frame rate and spectral precision can be selected.

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Spectral velocity estimation using autocorrelation functions for sparse data sets

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1 Field of the invention

The invention relates to an apparatus for determining the velocity distribution for a remotely sensed object using ultrasound and at the same time display an image of the object under investigation. The data acquisition is done by sending ultrasound pulses for velocity estimation and image acquisition inter-spaced in a deterministic or random order. The velocity distribution is determined from the autocorrelation function of the sparse signal sequence for velocity estimation, and the ultrasound image is displayed from the intervening emissions. A high velocity range and image frame rate can hereby be maintained. The corresponding audio signal can also be regenerated from the estimated spectrum.

2 Background of the invention

Medical ultrasound systems can be used for finding the blood and tissue velocity within the body [1]. This is done by emitting a pulse consisting of a number of sinusoidal oscillations, and then measure the scattered signal returned from the blood or tissue. The measurement is repeated a number of times, and data are sampled at the depth of interest in the tissue yielding one sample per pulse emission. The frequency of the received sampled signal is proportional to the velocity of the object along the ultrasound beam and is given by:

$$f_p = \frac{2|\vec{v}|\cos\theta}{c} f_0 \tag{1}$$

where $\vec{v}$ is the velocity vector, θ is the angle between the ultrasound beam and the velocity vector, c is the speed of sound, and f_0 is the emitted ultrasound frequency.

The velocity distribution for a given spatial position over time can be found by focusing the ultrasound beam at the position of interest. The received RF data is Hilbert transformed to give the in-phase and quadrature component. The data is sampled at the depth of interest to give the complex signal $y(i)$, where i is the pulse emission number. A Fourier transform is then performed on the sampled data. The power spectrum corresponds to the velocity distribution, and the short time Fourier transform displayed over time reveals the temporal variation of the velocity distribution.

The sampled data used for determining the velocity distribution has a sampling frequency of:

$$f_{prf} = \frac{c}{2d}, \tag{2}$$

where d is the depth of interrogation. The maximum frequency, that can be correctly found, is, thus, $f_{max} <= f_{prf}/2$ and the maximum unambiguous velocity is

$$v_{max} = \frac{c}{2\cos\theta} \cdot \frac{f_{prf}}{2f_0}. \tag{3}$$

The Fourier transform of the data is performed on short segments of data consisting of usually 128 or 256 samples to capture the frequency variation over time of the signal. A Hanning window is often applied on the data and the fast Fourier transform is then performed. An estimate of the power spectrum $\hat{P}_y(f)$ of the sampled complex signal $y(i)$ for a rectangular window is

$$\hat{P}_y(f) = \frac{1}{N}\left|\sum_{i=0}^{N-1} y(i)\exp(-j2\pi fi/f_{prf})\right|^2, \tag{4}$$

where i is the sample number and N is the number of samples in a segment.

The estimate has a significant variance given by

$$\mathrm{Var}[\hat{P}_y(f)] \approx P_y^2(f)\left[1 + \left(\frac{\sin 2\pi f/f_{prf}N}{N\sin 2\pi f/f_{prf}}\right)^2\right], \tag{5}$$

where $P_y(f)$ is the true power spectrum. The variance is for $f \neq 0$, thus, on the order of the estimate itself, and this is seen as speckle noise in the resulting spectral display.

Often a B-mode image should be shown at the same time for orientation and selection of the point of interest, and time must be spend on acquiring this image. This can either be done by acquiring the B-mode data interleaved with the velocity data or by acquiring a full B-mode image over a time interval. The first approach will only make every second emission useful for velocity estimation, and this will reduce the sampling frequency by a factor of 2 and reduces the maximum velocity v_{max} by a factor of 2. The second approach introduces periods, where no velocity estimation can be made, since data is not acquired, and the true velocity variation can therefore not be followed.

The components in the measured signal will lie in the audio range. For emitted frequencies f_0 of 3 to 5 MHz and velocities of 0.5 to 2 m/s at $\theta = 45^o$ gives frequencies f_p of 1 to 9 kHz, which can be perceived by the human ear. The sound of the measured signal is, thus, often played. This is a problem in the second approach, where there are gaps in the audio stream. This will easily be perceived by the human ear, and the signal cannot be used for faithful audio reproduction.

3 Summary of the invention

It is the object of the invention to overcome this deficiency and disadvantage of the known kinds of apparatus. With the invention this object is achieved by an apparatus, that acquires a sparse sequence of sampled data, and then uses an autocorrelation estimator and a Fourier transform for determine the velocity distribution. This makes is possible to keep the highest attainable velocity equal to the theoretical maximum, at the same time as a B-mode image can be acquired using part of the sparse data sequence. The invention can also be used for reconstructing the audio signal.

The limit on maximum velocity can also be exceeded by using a cross-correlation estimator to find the mean velocity and then adjust the velocity distribution according to this estimate.

In a first aspect, the invention relates to a method of estimating a velocity of tissue or of a fluid in a fluid vessel inside a person or an animal, the method comprising:

- transmitting a plurality of ultrasound pulses into the person or animal in a direction toward the vessel or tissue, the pulses being at least substantially equidistant in time,
- receiving a plurality of ultrasound pulses reflected by or emitted from the fluid or tissue,
- estimating the velocity on the basis of a first group of the received pulses,
- providing at least one image of the vessel/tissue, each image being provided on the basis of a predetermined number of a second group of the received pulses,

wherein the transmitting step comprises intermittently transmitting series of one or more pulses from the first group and one or more pulses from the second groups and wherein

- at least one series of pulse(s) from the first group comprises at least two neighbouring pulses and
- the number of consecutive pulses in each series of pulses of the second group is less than 10% of the predetermined number, and

wherein the estimating step comprises:

- calculating an autocorrelation function on information derived from each of the received pulses of the first group within a predetermined time interval, and
- estimating a velocity spectrum of the tissue/fluid as a power spectrum of the autocorrelation function.

Thus, images may be provided while providing the basis for determining the velocity of the tissue or fluid, such as in a predetermined position in the image.

Images are normally generated on the basis of pulses from a number of the series of pulses of the second group, and the velocity estimate may be provided simultaneously from the information provided by the pulses of the first group being spread over the same period of time.

In general, when estimating a velocity on the basis of a plurality of pulses, the maximum velocity determinable is determined by the time spacing of neighbouring pulses used for that determination. Requiring that at least sometimes, such as at least each predetermined period of time, two pulses of the first group are neighbouring, the maximum velocity determinable is determined by the time period between the pulses. In fact, when the velocity estimation comprises providing the estimate using an autocorrelation function based on the information of the pulses of the first group, the information at the points in time where pulses of the second group are transmitted may be reconstructed if desired.

Normally, the fluid velocity within a fluid vessel, such as an artery or a vein, will differ from the sides of the vessel to the centre thereof. Also, constrictions or the like may be interesting phenomena to investigate. The providing of the image helps the positioning of the transmitting means, so that the correct estimate of the velocity is obtained.

To sample and process the data comprised within the received pulses, the pulses transmitted into the person or animal are normally transmitted as equidistantly in time as possible. Small variations may, however, occur due to delays etc in the equipment.

When the "distance" between the series of pulses of the first group are limited (to the max 10% of the predetermined number), the autocorrelation function will be able to estimate the information at the times where the pulses of the second group were transmitted. Thus, this mathematical algorithm may provide velocity estimation or data as if all pulses transmitted were of the first group, i.e. as if no image data was provided.

Presently, the desired output is a velocity spectrum illustrating the full spectrum of the investigated moving object (tissue of fluid).

Suitably, the step of estimating the power spectrum may comprise estimating the Fourier Transform of the autocorrelation function.

Another desired output of the estimation is an audio signal relating to the velocity of the fluid/tissue. In the present manner, this may be provided by, continuously or repeatedly:

- generating, from the autocorrelation function, a filter representing the estimated velocity,
- providing an initial signal, and
- providing the audio signal by filtering the initial signal with the filter.

Thus, as the autocorrelation function is continuously calculated on differing sets of the received pulses, the sound generation is varied by keeping estimating new filters when a new autocorrelation function is calculated.

The initial signal normally is noise, such as white noise, but any signal may be used that contains frequencies in the band in which the audio signal is to be constructed.

Consequently, in the cause of varying velocity, the autocorrelation function will vary, varying the filter and consequently varying the audio signal provided to the user.

Even though the autocorrelation function is a known method, it is desired, in the presently preferred embodiment, to calculate it in a new manner. Thus, preferably, the step of calculating the autocorrelation function on a first set of pulses, the set comprising a predetermined number of pulses of the first group, comprises:

- providing a super set comprising the first set of pulses and a number, corresponding to the predetermined number, of subsequent pulses of the first group,
- for $k = 1$ to the predetermined number: multiplying the pulses of the first set with a second set of pulses from the super set, the second set comprising the predetermined number of

pulses, the second set of pulses being shifted k pulses, within the super set, in relation to the first set of pulses,

- adding the resulting values.

Thus, when calculating the autocorrelation function of a set of pulses, twice as many pulses are used, where all pulses are consecutive pulses of the first group. The first set of pulses are used in each iteration, but the other set of pulses to be multiplied on to the first set of pulses is shifted one pulse from the beginning of the super set (thus having a large overlap with the first set) toward the end thereof (having finally no overlap with the first set).

The advantage of this method may be seen in that a large overlap between the first and second sets provides good estimates for low velocities (small shift between the sets), whereas the shifted set provides a good estimate at high velocities.

Thus, performing the autocorrelation in this manner provides a final function which has a good precision at both high and low velocities.

When providing or illustrating the estimated velocity spectrum, it may be a problem that a large part of the elements of the person or animal in the path of the pulses of the first group are stationary or close to stationary. Thus, a large contribution of the velocity spectrum has a low velocity. This, however, need not be the interesting part of the spectrum. Consequently, the step of estimating the velocity spectrum may further comprise the steps of either determining a mean value of the pulses of the first group and removing the mean value from the pulses or determining a DC value of the velocity spectrum and removing the DC value from the velocity spectrum.

In this manner, the velocity spectrum is more easily visualized with a high resolution at the relevant velocity.

In general, the overall providing or transmitting of the pulses into the person or animal may be chosen in a number of manners, as long as the overall limits are honoured. Thus, every once in a while, two neighbouring pulses should be of the first group and the "pauses" between series of pulses of the first group should not be too long, such as no longer than 10% of the predetermined number, preferably no longer than 1% of the predetermined number, such as down to 1-3 pulses.

In addition, the number of pulses in each series of pulses needs not be the same. This is, however, a simple manner of providing the pulses.

In one situation, the transmitting step comprises repeating a number of the intermittent series of pulses. Thus, a number of series (intermittent series with pulses of the first group and series with pulses of the second group) may be determined and then simply repeated.

In another situation, the transmitting step comprises transmitting, as a series of pulses from the first group, a series comprising a randomly selected number of pulses from the first group. Thus, each series of pulses of the first group will have a randomly selected number of pulses.

Alternatively or in addition, the transmitting step may comprise transmitting, as a series of pulses from the second group, a series comprising a randomly selected number of pulses from the second group. Thus, also the series of pulses of the second group may be randomly selected, as long as they remain below the maximum number of pulses.

Normally, when generating ultrasound images, the pulses forming the basis of the images are provided in a number of directions, where after the image may be generated on the basis of the pulses received from those directions.

Conversely, the pulses used for determining the velocity are transmitted into and received from a single direction toward the vessel or tissue.

Thus, preferably, the step of providing the image comprises providing the image from pulses received from each of a plurality of directions inside the person or animal, and wherein the transmitting step comprises transmitting the pulses of the first group in a predetermined direction into the person or animal and each pulse of the second group into one of the plurality of directions.

Also, preferably, a pulse of the first group is transmitted during a first period of time and the step of receiving a pulse comprises receiving a part of the pulse during a period of time being at least the first period of time and being delayed a predetermined period of time from the transmission thereof.

The delay relates to the depth of the interesting tissue/vessel in the person or animal. The larger the distance between the ultrasound transducer, eg., and the vessel/tissue, the longer the delay before the reflected ultrasound pulse is received. Thus, this delay may be changed depending on the actual application.

Normally, the received pulse is sampled or received only a fraction of the time duration of the actual pulse (as defined by the time duration at launch/transmission). However, it has been found that it is advantageous to actually sample the received pulse during a period of time being at least that during which the pulse was transmitted. In this manner, it is possible to actually average the data and obtain a better precision.

It is clear from the above that the ratio between the number of pulses (within a given period of time or number of pulses) of the first and second groups may be more or less freely determined - and may actually vary

Thus, an interesting embodiment is one comprising the steps of:

- the transmitting step firstly transmitting intermittent series of pulses of the first and second groups with a first ratio of the number of pulses of the first group to the number of pulses of the second group and
- the transmitting step next transmitting intermittent series of pulses of the first and second groups with a second ratio of the number of pulses of the first group to the number of pulses of the second group, the first and second ratios being different.

The ratio of the number of pulses of the first group to the number of pulses of the second group determines the frequency of the providing of the images and the precision of the determination of the velocity spectrum or the generation of the corresponding sound.

Thus, a high ratio of pulses of the second group may be desired initially in order to obtain a high frequency of images in order to position the ultrasound transceiver correctly in relation to the tissue/vessel, where after the ration may be altered to emphasize the pulses of the first group to obtain a better velocity estimation.

In a second aspect, the invention relates to an apparatus for estimating a velocity of tissue or a fluid in a fluid vessel inside a person or an animal, the apparatus comprising:

- means for transmitting a plurality of ultrasound pulses into the person or animal in a direction toward the vessel or tissue, the pulses being at least substantially equidistant in time,
- means for receiving a plurality of ultrasound pulses reflected by or emitted from the fluid or tissue,
- means for estimating the velocity on the basis of a first group of the received pulses,
- means for providing at least one image of the vessel/tissue, each image being provided on the basis of a predetermined number of a second group of the received pulses,

wherein the transmitting means comprises means for intermittently transmitting series of one or more pulses from the first group and one or more pulses from the second groups and further being adapted to

- transmit at least one series of pulse(s) from the first group comprising at least two neighbouring pulses and
- transmit the series of pulses of the second group with a number of consecutive pulses being less than 10% of the predetermined number, and

wherein the means for estimating the velocity comprise means for:

- calculating an autocorrelation function on information derived from each of the received pulses of the first group within a predetermined time interval,
- estimating a velocity spectrum of the tissue/fluid as a power spectrum of the autocorrelation function.

Normally, the transmitting means and the receiving means are provided in one and the same ultrasound transmitter, which is a known instrument.

Suitably, the estimating means are adapted to estimate the power spectrum on the basis of a Fourier Transform of the autocorrelation function.

As mentioned above, it may be desired to also have means for providing an audio signal, the audio signal providing means comprising means for, continuously or repeatedly:

- generating, from the autocorrelation function, a filter representing the estimated velocity,
- providing an initial signal, and
- providing the audio signal by filtering the initial signal with the filter.

Thus, when a new autocorrelation is calculated (which may be when a new pulse of the first group is received or performed equidistantly in time), a new filter may be generated, whereby the provided audio signal will alter with altering autocorrelation and thereby altering velocity.

An especially interesting manner of calculating the autocorrelation is one where the means for calculating the autocorrelation function are adapted to calculate the autocorrelation function on a first set of pulses, the set comprising a predetermined number of pulses of the first set, by:

- providing a super set comprising the first group of pulses and a number, corresponding to the predetermined number, of subsequent pulses of the first group,
- for $k = 1$ to the predetermined number: multiplying the pulses of the first set with a second set of pulses from the super set, the second set comprising the predetermined number of pulses, the second set of pulses being shifted k pulses, within the super set, in relation to the first set of pulses, and
- adding the resulting values.

In order to emphasize the interesting velocity/velocities in the velocity spectrum, the estimating means may further be adapted to either determine a mean value of the pulses of the first group and remove the mean value from the pulses or determine a DC value of the velocity spectrum and remove the DC value from the velocity spectrum.

Naturally, the transmitting means may simply repeat transmitting the same series of pulses of the first group intermittently with the same series of pulses of the second group.

Alternatively, the transmitting means may be adapted to repeat a number of the intermittent series of pulses. Thus, a more complex pattern of the series may be provided while keeping the method simple.

A more complex method is seen when the transmitting means are adapted to provide:
- as a series of pulses from the first group, a series comprising a randomly selected number of pulses from the first group and/or
- as a series of pulses from the second group, a series comprising a randomly selected number of pulses from the second group.

Preferably, the means for providing the image is adapted to provide the image from pulses received from each of a plurality of directions inside the person or animal, and wherein the transmitting means are adapted to provide the pulses of the first group in a predetermined direction into the person or animal and each pulse of the second group into one of the plurality of directions.

This type of ultrasound transmitter normally comprises an array of ultrasound transmitting elements, and means for phase shifting the signal for each element in the array in order to direct the ultrasound in the direction desired. This known type of transducer is well suited for the present purpose.

An interesting reception strategy is one where the transmitting means are adapted to transmit a pulse of the first group during a first period of time and wherein the receiving means are adapted to receive a part of the pulse during a period of time being at least the first period of time and being delayed a predetermined period of time from the transmission thereof. In this manner, it is possible to actually average over a certain depth within the person or animal and thereby obtain a better precision of the velocity estimate.

Finally, an interesting embodiment is one further comprising means for receiving information relating to a ratio of the number of pulses to be provided of the first group to the number of pulses of the second group, and wherein the transmitting means are adapted to transmit intermittent series of pulses of the first and second groups with a received ratio of the number of pulses of the first group to the number of pulses of the second group.

Consequently, having positioned the transceiver correctly, the ratio receiving means may be operated in order to actually emphasize more on the pulses of the first group (transmit more pulses of the first group per unit time), whereby the frequency of image generation is lowered and the precision of the velocity estimation is increased.

4 Brief description of the drawings

A preferred embodiment of the invention will be described in detail hereinafter with reference to the accompanying drawings.

Fig. 1 shows schematically a block diagram of the main components of the system.

Fig. 2 shows a typical result from a traditional spectral estimation and the result for a velocity determination using the new approach.

5 Description of the preferred embodiment

In Fig. 1 is shown an example of a preferred embodiment of an apparatus according to the invention. This embodiment of the invention has its application within diagnostic medical ultrasound. A typical example is the determination of blood flow in peripheral vessels such as arteries in an arm, a leg, or in the carotid artery or the determination of tissue velocity.

The underlying theory and operation of the system will now be described.

5.1 Power spectrum estimation

The power spectrum of a stochastic signal is formally calculated from the Fourier transform of the autocorrelation function $R_y(k)$ as:

$$R_y(k) \leftrightarrow P_y(f). \tag{6}$$

An estimate of the autocorrelation can be calculated by:

$$\hat{R}_y(k) = \frac{1}{N-|k|} \sum_{i=0}^{N-k-1} y(i)y^*(i+k), \tag{7}$$

when data are available for a segment of N samples and * denotes complex conjugate. The estimate of the power spectrum is then calculated by applying, *e.g.*, a Hanning window on $\hat{R}_y(k)$ and then performing a Fourier transform. A trade-off between spectral resolution and spectral estimate variance can be selected by using a window shorter than $2N-1$.

5.2 Sparse data sequences

The autocorrelation function calculated by (7) is found by correlating all samples in the signal segment $y(i)$ with a time shifted version $y(i+k)$ of the signal. It is, however, possible to calculate the correlation estimate even if some of the samples in the signal are missing. The correlation is then calculated with fewer values, and this will result in an increased standard deviation of the estimate. In general the variance of the estimate is inversely proportional to the number of independent values, which here is proportional to $N-k$. Having $M(k)$ missing values will increase the variance by a factor $(N-k)/(N-k-M(k))$. Keeping $M(k)$ moderate compared to N will, thus, give a moderate increase in variance. The overall variance of the spectral estimate will be determined by the lag values with the highest variance, and therefore is should be ensured that $M(k)$ roughly has the same value for all k.

For a sparse sequence $M(k)$ will in general depend on the lag k, and it must be ensured that all lag values of $\hat{R}_y(k)$ can be calculated with a sufficient accuracy. The estimate of autocorrelation function is then:

$$\hat{R}_y(k) = \frac{1}{N-|k|-M(k)} \sum_{i=0}^{N-k-1} y(i)y^*(i+k), \tag{8}$$

where missing data in the signal are represented by a zero. This equation assumes that only a fixed segment of data is passed to the estimator.

It is also possible to use data from the next segment. The estimate of autocorrelation function is then:

$$\hat{R}_y(k) = \frac{1}{N-M(k)} \sum_{i=0}^{N-1} y(i)y^*(i+k), \tag{9}$$

since data for $2N$ samples are available It is then possible to get a more accurate estimate of higher lags in the autocorrelation function as more data are used, which improves the accuracy of the final velocity estimate.

It should also be noted that only the autocorrelation function for positive lags needs to be calculate, since negative lags can be reconstructed from

$$\hat{R}_y(k) = \hat{R}_y^*(-k). \tag{10}$$

The missing values in the sparse sequence can be used for *e.g.* B-mode emissions, so that a B-mode image can be acquired simultaneously with the velocity data. An example of a sequence is:

v v b v v b ...,

where v is a velocity emission and b is a B-mode emission. Overlapping for the different lags is illustrated by:

$k=0$	v	v	b	v	v	b	...
	v	v	b	v	v	b	...
$k=1$	v	v	b	v	v	b	...
	b	v	v	b	v	v	...
$k=2$	v	v	b	v	v	b	...
	v	b	v	v	b	v	...
$k=3$	v	v	b	v	v	b	...
	v	v	b	v	v	b	...

It can be seen that for all sequences there is overlap between velocity data, and all lags can therefore be calculated. For this sequence 66% of the time is spent on velocity data and 33% is spend on B-mode data acquisition. For imaging to a depth of 15 cm, a pulse repetition frequency of 5 kHz can be maintained, and this gives a frame rate of 15 images/sec for images consisting of 100 emissions.

The frame rate can be lowered by inserting more flow emissions between each B-mode emission, and the B-mode frame rate can therefore easily be selected. Other sequences can put more emphasize on the B-mode imaging to increase frame rate at the drawback of an increased variance of the spectral estimate. Some other sequences are:

B-mode	Flow	
40 %	60 %:	v b v v b ...
50 %	50 %:	v b v v b b ...
57 %	43 %:	v b v v b b b ...
62 %	38 %:	v b v v b b b v v b b b b ...

The interleaved emissions can also be used for color flow mapping, which also can be found from sparse sequences. A 50%-50% sequence can also be used to make two spectral estimates at the same time with full velocity range.

It is also possible to use fully random sequences, where there is no deterministic repetition of the emission sequence. The sequence could for example be determined by using a white, random signal $x(n)$ with a rectangular distribution between zero and one. The determination of whether a B-mode or flow emission should be made is determined by

$$e(n) = (x(n) > P_f), \tag{11}$$

where $e(n) = 1$ indicates a flow emission and $e(n) = 0$ indicates a B-mode emission, and P_f is the probability of flow emission. The ratio between flow and B-mode emissions is then determined by P_f and $1 - P_f$, respectively. It has to be ensured that the autocorrelation can be found for all lags as explained above. The advantage of this approach is that noise, that might be repetitive with the deterministic firing sequence, is spread out over the full spectrum, and that the time division between flow estimation and B-mode imaging can be precisely tailored to the need using P_f.

5.3 Averaging RF data

The pulse emitted for velocity estimation will in general have a number of sinusoidal oscillations to keep the bandwidth small and increase the emitted energy. The received signal is then correlated over the pulse duration, and applying a matched filter to increase the signal-to-noise ratio will increase the correlation to a duration of roughly two pulse lengths. This data can also be used in calculation of the autocorrelation as:

$$\hat{R}_y(k) = \frac{1}{(N-|k|-M(k))N_r} \sum_{j=0}^{N_r-1} \sum_{i=0}^{N-k-1} y(j+J_d,i)y^*(j+J_d,i+k), \tag{12}$$

where j is the RF sample index, J_d is the index for the depth of the range gate start, and N_r is the number of RF samples. Averaging over several RF samples will in general lower the variance of the estimated autocorrelation function and thereby of the spectral estimate.

It is also possible to use data from the next segment. The estimate of autocorrelation function is then:

$$\hat{R}_y(k) = \frac{1}{(N-M(k))N_r} \sum_{j=0}^{N_r-1} \sum_{i=0}^{N-1} y(j+J_d,i)y^*(j+J_d,i+k), \tag{13}$$

since data for $2N$ samples are available. It is then possible to get a more accurate estimate of higher lags in the autocorrelation function as more data is used, which improves the accuracy of the final velocity estimate.

5.4 Stationary echo canceling

The measured signal will often contain large signal components around low frequencies emanating from the tissue, especially near the vessel wall. These signals can be removed if they obscure the blood velocity signal and makes its spectral visualization difficult. This can be done either in the time or the frequency domain. The first approach is to take the mean value of the signals and subtract that. The mean signal as a function RF sample number j is found from

$$y_{sta}(j) = \frac{1}{(N-M(k))} \sum_{i=0}^{N-1} y(j,i), \tag{14}$$

where $y_{sta}(j)$ is the estimated stationary signal. Missing RF signals are replaced by zeros in the sum. The estimated stationary signal is then subtracted from $y(j,i)$ to remove a fully stationary component. This should be done before the autocorrelation function is calculated.

The stationary echo canceling can also be performed by fitting a first or higher order polynomial to the data using e.g. a least squares fit. The polynomial will only be fitted to the data, that has been measured. The values for the polynomial is then subtracted from this data in order to remove the stationary component.

This processing can also be performed in the frequency domain. Here frequency components around $f = 0$ Hz are set to zero in the spectrum to remove the stationary component. The cut-off frequency in the spectrum should be determined from the velocity of the tissue surrounding the blood vessel using (1).

5.5 Audio reproduction

The audio signal can be regenerated from the estimated autocorrelation function. An appropriate model for the audio signal $y(n)$ is given by

$$y(n) = h(n;n) * e(n) \tag{15}$$

where $h(n;n)$ is a time varying filter impulse response at time index n and $e(n)$ is a Gaussian, white random signal. $e(n)$ models the many random and independent red blood cells in the vessel. $h(n;n)$ models the velocity spectrum at the given time. The filter is time varying, since the velocity and thereby frequency content varies over the cardiac cycle. The autocorrelation of this is

$$R_y(k;n) = R_h(k;n) * R_e(k) = R_h(k;n) * P_e\delta(k) = P_e R_h(k;n) \leftrightarrow P_e|H(f;n)|^2, \tag{16}$$

where P_e is the power of the blood scattering signal and $H(f;n)$ is the Fourier transform of $h(n;n)$. The linear phase impulse response of the filter can then be found from

$$h_l(k;n) = \mathcal{F}^{-1}\left\{\sqrt{\mathcal{F}\{R_y(k;n)\}}\right\} = \mathcal{F}^{-1}\left\{\sqrt{P_e}|H(f;n)|\right\} \tag{17}$$

where $\mathcal{F}\{\}$ denotes Fourier transform and $\mathcal{F}^{-1}\{\}$ inverse Fourier transform. A window can be applied to the impulse response to reduce edge effects. It is also appropriate to mask out small amplitude values in the frequency domain, since this most probably is noise from the reconstruction process.

The phase of the filter is neglected and only a linear phase version is reconstructed. A minimum phase version could be reconstructed using a Hilbert transform, but this is of no consequence, since it is a stochastic signal that needs to be made. The final signal is made by convolving $h_l(k;n)$ with a Gaussian, white random signal. This will be the audio signal for a given time segment, and this signal should be added to signals from other segments properly time aligned. To avoid edge effects, a window is applied on the signal segment before addition.

5.6 Increasing the maximum velocity

The maximum velocity, that can be estimated, is restricted by (3) due to aliasing. This is really not a restriction on the maximum velocity, but on the widest spread of velocities, where the distance between the lowest and highest velocity at any given time must be less than

$$2v_{max} = \frac{c}{2\cos\theta}\frac{f_{prf}}{f_0}. \tag{18}$$

Estimating the mean velocity and adjusting the spectrum to lie around this velocity can therefore increase the maximum velocity range.

The maximum velocity can be estimated by using the cross-correlation approach developed in [2]. Two or more RF lines are then cross-correlated and the shift in time between them found. This will yield the mean velocity of the flow. The center of the spectrum is then offset to lie around this mean frequency.

The same data as for the spectral estimation can be used, if a narrow pulse is emitted. The spectrum will be widened due to the wide bandwidth of the pulse, but this can be avoided by filtering the received RF data with a narrow-band pulse before calculation of the autocorrelation function. This will narrow the bandwidth and the velocity spectrum width.

5.7 Directional focusing

Data beamformed along the flow direction as described in [3] can also be used for the flow estimation. The received data then tracks the movement of the scatterers, and a single or narrow distribution of velocities are then found. This will give a spectrum, that is narrower than for taking data out at a range of depths.

5.8 A preferred embodiment

In Fig. 1 the specific setup of the measuring apparatus itself is indicated schematically. It comprises a generator or pulser 1, an emit beam former 2, an array ultrasound emitting transducer 3 providing an emitted field, an array ultrasound receiving transducer 5 receiving a scattered field, a receive beam former 6, and a velocity estimator 7 for estimating the velocity spectrum. The estimate is the passed on to a display 8 and to a unit 10 that reconstructs the audio sound from the spectrum. The unit 9 is used for presenting the B-mode image.

The pulser 1 generates a pulsed voltage signal with four sinusoidal oscillations at a frequency of 2 MHz in each pulse, that is fed to the emit beam former 2. The emit beam former 2 splits up the signal from the pulser into a plurality of signals, which are being fed to the respective elements of the emitting transducer array 3. The emit beam former 2 is capable of individually attenuating and delaying the signals to each of the elements of the transducer array 3.

In the preferred embodiment the same array transducer is used for both emitting and receiving the pulsed ultrasound field. It consists of 64 elements with an element width of 0.43 mm and a spacing between neighboring elements of 0.05 mm. The height of the elements is 5 mm.

The emitted field from the transducer is scattered by the blood in the blood vessel 4 and part of the scattered field is received by the array transducer, and the signals from the individual elements are passed on to the receive beam former. The signals from the elements are individually scaled in amplitude and individually delayed and are thereafter summed to yield a single output signal from the receive beam former focused at the depth of interest. A Hilbert transformation is the performed on the data to yield the in-phase and quadrature component for $y(i)$.

This processing is repeated for a number of emissions. Two emissions are done for flow estimation and one emission can be used for making a B-mode image. A sequence with 128 emissions are made and the emissions for flow estimation are collected in the velocity estimator processor 7. The autocorrelation function of this data is calculated at the depth of interest using

$$\hat{R}_y(k) = \frac{1}{(N-M(k))N_r} \sum_{j=0}^{N_r-1} \sum_{i=0}^{N-1} y(j+J_d,i)y^*(j+J_d,i+k) \tag{19}$$

Transducer center frequency	f_0	2 MHz
Pulse cycles	M	4
Speed of sound	c	1540 m/s
Pitch of transducer element	w	0.435 mm
Height of transducer element	h_e	5 mm
Kerf	k_e	0.05 mm
Number of active elements	N	64
RF lines for estimation	N	128
RF samples for estimation	N_r	20
Sampling frequency	f_s	10 MHz
Pulse repetition frequency	f_{prf}	5 kHz
Radius of vessel	R	2 mm
Distance to vessel center	Z_{ves}	60 mm
Angle between beam and flow	θ	55°

Table 1: Standard parameters for transducer and femoral flow simulation.

for positive values of k. The values for negative lags are calculated by

$$\hat{R}_y(k) = \hat{R}_y^*(-k). \tag{20}$$

The power spectrum of the data is then calculated by

$$\hat{P}_y(f) = \sum_{k=-N}^{N} \hat{R}_y(k) \exp(-j2\pi f k T_{prf}). \tag{21}$$

The process is repeated continuously and the spectra are displayed as a gray scale image as a function of time and frequency or velocity.

An example of results from application of the method is shown in Fig. 2. Simulated data from flow in the femoral artery has been used with the parameters shown in Table 1. The RF data was obtained from simulating the flow in the femoral artery by using the Field II program [4] and the Womersley-Evans method for pulsed flow. One heart cycle of pulsatile flow was simulated and the received RF signal from the array focused at the vessel was found. The data were then processed using the traditional approach using a Hanning window on 128 samples segments. The result is shown on the top in Fig. 2. A sparse sequence was then used in the new approach, where every third received signal was replaced by zeros (v v B sequence). The autocorrelation estimate was calculated using (12) and the parameters in Table 1. A Hanning window covering 75% of the autocorrelation function was multiplied onto it and the power spectrum found. This is shown in the bottom in Fig. 2. It can be seen that a more smooth spectrum can be found although 33% of the data is missing.

6 Claims

1. A method of estimating a velocity of tissue or fluid in a vessel inside a person or an animal, the method comprising:

- transmitting a plurality of ultrasound pulses into the person or animal in a direction toward the vessel or tissue, the pulses being at least substantially equidistant in time,
- receiving a plurality of ultrasound pulses reflected by or emitted from the fluid or tissue,
- estimating the velocity on the basis of a first group of the received pulses,
- providing at least one image of the vessel/tissue, each image being provided on the basis of a predetermined number of a second group of the received pulses,

wherein the transmitting step comprises intermittently transmitting series of one or more pulses from the first group and one or more pulses from the second groups and wherein

- at least one series of pulse(s) from the first group comprises at least two neighbouring pulses and
- the number of consecutive pulses in each series of pulses of the second group is less than 10% of the predetermined number and

wherein the step of estimating the velocity comprises:

- calculating an autocorrelation function on information derived from each of the received pulses of the first group within a predetermined time interval, and
- estimating the velocity spectrum of the tissue/fluid as a power spectrum of the autocorrelation function.

2. A method according to claim 1, wherein the step of estimating the power spectrum comprises estimating the Fourier Transform of the autocorrelation function.

3. A method according to claim 1 or 2, further comprising a step of providing an audio signal related to the velocity by, continuously or repeatedly:

- generating, from the autocorrelation function, a filter representing the estimated velocity,
- providing an initial signal, and
- providing the audio signal by filtering the initial signal with the filter.

4. A method according to any of the preceding claims, wherein the step of calculating the autocorrelation function on a first set of pulses, the set comprising a predetermined number of pulses of the first group, comprises:

- providing a super set comprising the first set of pulses and a number, corresponding to the predetermined number, of subsequent pulses of the first group,
- for $k = 1$ to the predetermined number: multiplying the pulses of the first set with a second set of pulses from the super set, the second set comprising the predetermined number of pulses, the second set of pulses being shifted k pulses, within the super set, in relation to the first set of pulses,
- adding the resulting values.

5. A method according to any of the preceding claims, wherein the step of estimating the velocity spectrum further comprises the steps of:
- determining a mean value of the pulses of the first group and
- removing the mean value from the pulses.

6. A method according to any of the preceding claims, wherein the step of estimating the velocity spectrum further comprises the steps of:
- determining a DC value of the velocity spectrum and
- removing the DC value from the velocity spectrum.

7. A method according to any of the preceding claims, wherein the step of estimating the velocity spectrum further comprises the steps of:
- fitting a first or higher order polynomial to the received data and
- subtracting the values of the polynomial from the data.

8. A method according to any of the preceding claims, wherein the transmitting step comprises repeating a number of the intermittent series of pulses.

9. A method according to any of the preceding claims, wherein the transmitting step comprises transmitting, as a series of pulses from the first group, a series comprising a randomly selected number of pulses.

10. A method according to any of the preceding claims, wherein the transmitting step comprises transmitting, as a series of pulses from the second group, a series comprising a randomly selected number of pulses.

11. A method according to any of the preceding claims, wherein the step of providing the image comprises providing the image from pulses received from each of a plurality of directions inside the person or animal, and wherein the transmitting step comprises transmitting the pulses of the first group in a predetermined direction into the person or animal and each pulse of the second group into one of the plurality of directions.

12. A method according to any of the preceding claims, wherein a pulse of the first group is transmitted during a first period of time and wherein the step of receiving a pulse comprises receiving a part of the pulse during a period of time being at least the first period of time and being delayed a predetermined period of time from the transmission thereof.

13. A method according to any of the preceding claims, the method comprising the steps of:
- the transmitting step firstly transmitting intermittent series of pulses of the first and second groups with a first ratio of the number of pulses of the first group to the number of pulses of the second group and
- the transmitting step next transmitting intermittent series of pulses of the first and second groups with a second ratio of the number of pulses of the first group to the number of pulses of the second group, the first and second ratios being different.

14. An apparatus for estimating a velocity of tissue or fluid in a vessel inside a person or an animal, the apparatus comprising:
- means for transmitting a plurality of ultrasound pulses into the person or animal in a

direction toward the vessel or tissue, the pulses being at least substantially equidistant in time,

- means for receiving a plurality of ultrasound pulses reflected by or emitted from the fluid or tissue,
- means for estimating the velocity on the basis of a first group of the received pulses,
- means for providing at least one image of the vessel/tissue, each image being provided on the basis of a predetermined number of a second group of the received pulses,

wherein the transmitting means comprises means for intermittently transmitting series of one or more pulses from the first group and one or more pulses from the second groups and further being adapted to

- transmit at least one series of pulse(s) from the first group comprising at least two neighbouring pulses and
- transmit the series of pulses of the second group with a number of consecutive pulses being less than 10% of the predetermined number and

wherein the means for estimating the velocity comprise means for:

- calculating an autocorrelation function on information derived from each of the received pulses of the first group within a predetermined time interval, and
- estimating a power spectrum of the autocorrelation function corresponding to the velocity spectrum of the tissue/fluid.

15. An apparatus according to claim 14, wherein the estimating means are adapted to estimate the power spectrum on the basis of a Fourier Transform of the autocorrelation function.

16. An apparatus according to claim 14 or 15, further comprising means for providing an audio signal, the audio signal providing means comprising means for, continuously or repeatedly:

- generating, from the autocorrelation function, a filter representing the estimated velocity,
- providing an initial signal, and
- providing the audio signal by filtering the initial signal with the filter.

17. An apparatus according to any of claims 14-16, wherein the means for calculating the autocorrelation function are adapted to calculate the autocorrelation function on a first set of pulses, the set comprising a predetermined number of pulses of the first group, by:

- providing a super set comprising the first set of pulses and a number, corresponding to the predetermined number, of subsequent pulses of the first group,
- for $k = 1$ to the predetermined number: multiplying the pulses of the first set with a second set of pulses from the super set, the second set comprising the predetermined number of pulses, the second set of pulses being shifted k pulses, within the super set, in relation to the first set of pulses, and
- adding the resulting values.

18. An apparatus according to any of claims 14-17, wherein the estimating means further are adapted to:

- determining a mean value of the pulses of the first group and

- removing the mean value from the pulses.

19. An apparatus according to any of claims 14-18, wherein the estimating means further are adapted to:
  - determining a DC value of the velocity spectrum and
  - removing the DC value from the velocity spectrum.

20. An apparatus according to any of the preceding claims, wherein the step of estimating the velocity spectrum further comprises the steps of:
  - fitting a first or higher order polynomial to the received data and
  - subtracting the values of the polynomial from the data.

21. An apparatus according to any of claims 14-20, wherein the transmitting means are adapted to repeat a number of the intermittent series of pulses.

22. An apparatus according to any of claims 14-21, wherein the transmitting means are adapted to provide, as a series of pulses from the first group, a series comprising a randomly selected number of pulses.

23. An apparatus according to any of claims 14-22, wherein the means for providing the pulses is adapted to provide, as a series of pulses from the second group, a series comprising a randomly selected number of pulses.

24. An apparatus according to any of claims 14-23, wherein the means for providing the image is adapted to provide the image from pulses received from each of a plurality of directions inside the person or animal, and wherein the transmitting means are adapted to provide the pulses of the first group in a predetermined direction into the person or animal and each pulse of the second group into one of the plurality of directions.

25. An apparatus according to any of claims 14-24, wherein the transmitting means are adapted to transmit a pulse of the first group during a first period of time and wherein the receiving means are adapted to receive a part of the pulse during a period of time being at least the first period of time and being delayed a predetermined period of time from the transmission thereof.

26. An apparatus according to any of claims 14-25, further comprising means for receiving information relating to a ratio of the number of pulses to be provided of the first group to the number of pulses of the second group, and wherein the transmitting means are adapted to transmit intermittent series of pulses of the first and second groups with a received ratio of the number of pulses of the first group to the number of pulses of the second group.

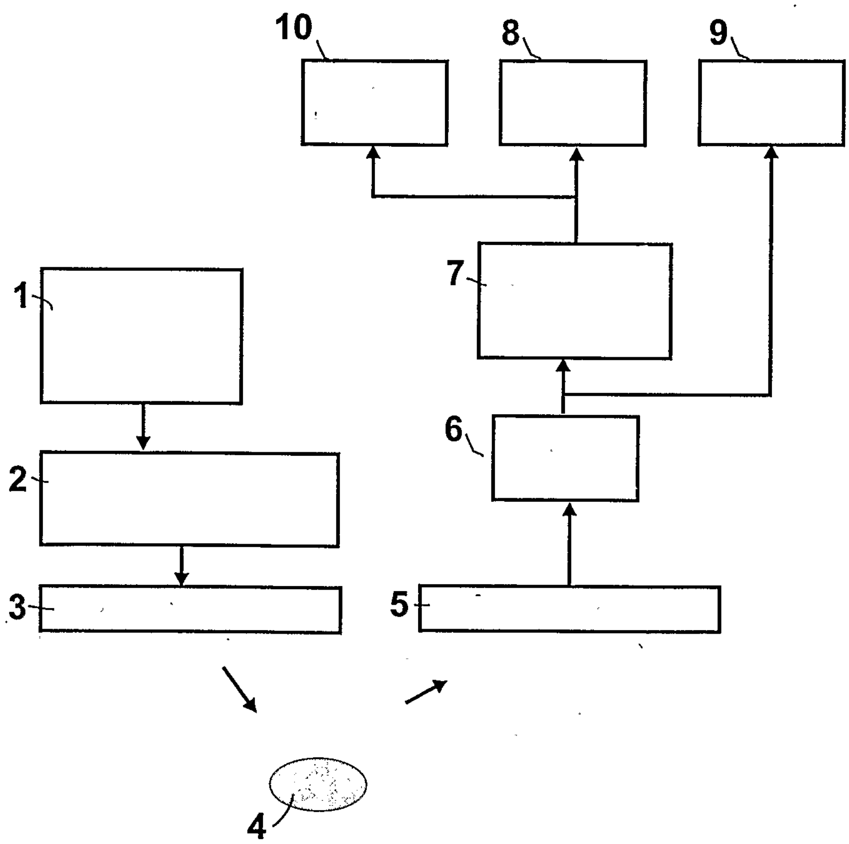

Fig. 1

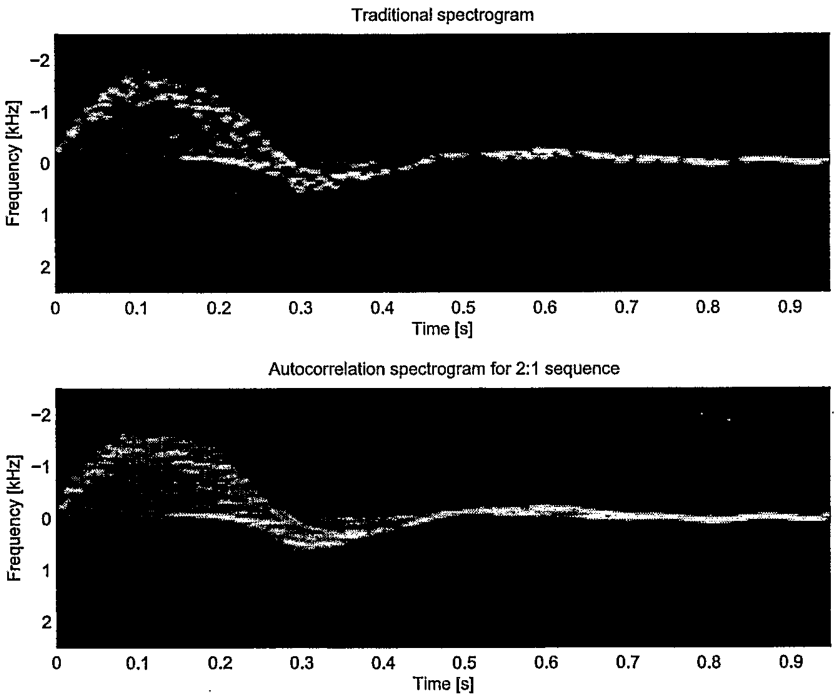
Traditional spectrogram
Frequency [kHz]
-2
-1
0
1
2
0
0.1
0.2
0.3
0.4
0.5
0.6
0.7
0.8
0.9
Time [s]
Autocorrelation spectrogram for 2:1 sequence
Frequency [kHz]
-2
-1
0
1
2
0
0.1
0.2
0.3
0.4
0.5
0.6
0.7
0.8
0.9
Time [s]

Fig. 2

INTERNATIONAL SEARCH REPORT

International Application No
PCT/DK2005/000392

A. CLASSIFICATION OF SUBJECT MATTER
IPC 7 G01S15/89 G01S15/58 A61B8/06

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)
IPC 7 G01S A61B

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, PAJ

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category °	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	US 6 423 006 B1 (BANJANIN ZORAN) 23 July 2002 (2002-07-23) the whole document	1-24
A	KIRKHORN J ET AL INSTITUTE OF ELECTRICAL AND ELECTRONICS ENGINEERS: "A new technique for improved spatial resolution in high frame rate color doppler imaging" 2003 IEEE ULTRASONICS SYMPOSIUM PROCEEDINGS. HONOLULU, HAWAII, OCT. 5, vol. VOL. 1 OF 2, 5 October 2003 (2003-10-05), pages 1947-1950, XP010701093 ISBN: 0-7803-7922-5 the whole document	1-24

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[X] Further documents are listed in the continuation of box C.

[X] Patent family members are listed in annex.

° Special categories of cited documents :

"A" document defining the general state of the art which is not considered to be of particular relevance
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"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)
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"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
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"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 July 2005	05/08/2005

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, Tx. 31 651 epo nl, Fax: (+31–70) 340–3016	Reuss, T

C.(Continuation) DOCUMENTS CONSIDERED TO BE RELEVANT

Category °	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
A	US 5 876 341 A (WANG ET AL) 2 March 1999 (1999-03-02) the whole document -----	1-24
A	US 5 501 223 A (WASHBURN ET AL) 26 March 1996 (1996-03-26) the whole document -----	1-24
A	BASOGLU C ET AL: "Computing requirements of modern medical diagnostic ultrasound machines" PARALLEL COMPUTING, ELSEVIER PUBLISHERS, AMSTERDAM, NL, vol. 24, no. 9-10, September 1998 (1998-09), pages 1407-1431, XP004148103 ISSN: 0167-8191 page 1412 - page 1421; figure 1; table 1 page 1428 -----	1-24

INTERNATIONAL SEARCH REPORT

Information on patent family members

International Application No
PCT/DK2005/000392

Patent document cited in search report		Publication date	Patent family member(s)	Publication date
US 6423006	B1	23-07-2002	NONE	
US 5876341	A	02-03-1999	NONE	
US 5501223	A	26-03-1996	NONE	