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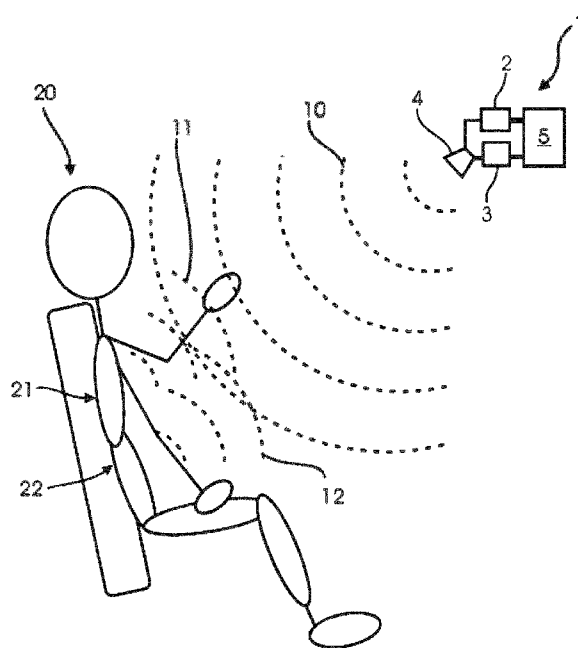


Fig. 1

(57) Abstract: A method for breathing monitoring using a radar sensor system (1), the radar sensor system (1) comprising a transmitter (2), a receiver (3) and processing device (5). In order to provide reliable and efficient means for breathing monitoring, the method comprises: the transmitter (2) simultaneously irradiating a chest region (21) and an abdominal region (22) of a person (20) with radar radiation (10) having a carrier frequency; the receiver (3) generating a receiver signal from reflected radiation (11, 12) from the chest region and the abdominal region; the processing device (5) generating an in-phase signal and a quadrature signal based on the receiver signal, and determining, from the in-phase signal and the quadrature signal, information characterizing a chest motion and an abdominal motion, at least relative to each other.

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RADAR SENSOR SYSTEM FOR BREATHING MONITORING AND CORRESPONDING METHOD

Technical field

[0001] The present invention relates method and a system for breathing monitoring.

Background of the Invention

[0002] Various methods for monitoring breathing of a person are known in the art. Such monitoring may be performed in order to assess the fitness of an athlete, to monitor sleeping behavior or to identify breathing anomalies. Some of these methods are contact methods, which e.g. require the person to wear a mask and/or stretch resistance bands. Apart from these, there are non-contact methods.

[0003] In order to detect certain breathing disorders, it is sufficient to monitor chest displacement as a function of time – to identify any changes in the breathing frequency or amplitude or sudden stops in breathing (apnea). However, in some cases, it is necessary to simultaneously monitor chest and abdominal displacement, since a significant degree of asynchrony between abdominal and pulmonary motion provides an indication of certain diseases such as bronchopulmonary dysplasia, obstructive sleep apnea, upper or lower airway obstruction, chronic lung disease in prematurely born infants, certain neuromuscular diseases and in general abnormalities in the thoraco-abdominal motion, which cannot be diagnosed by looking at the chest displacement signal alone.

[0004] Hence, there is a need to quantify and measure the degree of chest-abdomen asynchrony. In this regard, it is especially challenging to conduct experiments with young infants and children in order to monitor their breathing, and current state of the art methods, such as plethysmography, apart from suffering from a flaw of being contact measurements (and therefore potentially disturbing the infant), often lead to large uncertainties in the phase lag ϕ between the chest and abdominal motion. Some medical literature also states that the outcome of the plethysmography measurement might depend on the type of the mask used. It has been proposed to use Doppler radar information data to monitor breathing patterns, e.g. in Zakrzewski et al, "Noncontact Respiration Monitoring

During Sleep with Microwave Doppler Radar”, IEEE Sensors Journal, Vol. 15, No. 10, 2015. In this case however, only the chest displacement was analyzed. There have been other attempts to measure the chest-abdomen asynchrony using radar Doppler measurements. Such an approach is described e.g. in Gu et al, “Assessment of Human Respiration Patterns via Noncontact Sensing Using Doppler Multi-Radar System”, Sensors 2015, 15, 6383-6398. However, this approach relies on using two different radar units to monitor the chest and abdominal displacements separately, which may lead to synchronization problems and relatively high costs and does not provide phase coherent signals.

Object of the invention

[0005] It is an object of the present invention to provide reliable and efficient means for breathing monitoring. The object is achieved by a method according to claim 1 and a system according to claim 14.

General Description of the Invention

[0006] The present invention provides a method for breathing monitoring using a radar sensor system, the radar sensor system comprising a transmitter, a receiver and a processing device.

[0007] The method comprises:

- the transmitter simultaneously irradiating a chest region and an abdominal region of a person with radar radiation having a carrier frequency;
- the receiver generating a receiver signal from reflected radiation from the chest region and the abdominal region;
- the processing device generating an in-phase signal and a quadrature signal based on the receiver signal, and determining, from the in-phase signal and the quadrature signal, information characterizing a chest motion and an abdominal motion, at least relative to each other.

[0008] Although the transmitter and the receiver are at least partially distinct, separate components, an antenna of the radar sensor system may be used by the transmitter (as a transmitting antenna) as well as by the receiver (as a receiving antenna). The processing device comprises hardware components, although some of its functionalities may be realized by software. The processing device

may also comprise any kind of volatile or non-volatile memory device. Normally, the processing device is connected to the receiver by a wired connection. It may also be connected to the transmitter in order to monitor and/or control the operation of the transmitter.

[0009] According to the inventive method, the transmitter simultaneously irradiates a chest region and an abdominal region of a person with radar radiation having a carrier frequency. As implied by the term "radar", the radiation is an electromagnetic wave. The carrier frequency of this electromagnetic wave is preferably between 1 GHz and 300 GHz. Preferably, the radiation is a continuous wave of the following form:

$$u_t = A_t e^{-i(2\pi f t)}$$

where f is the carrier frequency and A_t is the amplitude of the transmitted signal. Alternatively, a frequency modulated function may be used. This radiation from the transmitter simultaneously irradiates a chest region of a person as well as an abdominal region of this person. In order to achieve simultaneous irradiation, the transmitter may use an antenna with a relatively broad main lobe. Normally, the skin of the person is irradiated, i.e. the torso of the person should be exposed. However, if the person wears clothing that is skintight and/or (largely) transparent for the radiation, the inventive method will work as well. The person may be an adult or, in particular, a child. Since the inventive method is a non-contact method, it can be used for any kind of person, especially for children, without creating any disturbance or discomfort. It is understood that while some transmission through the person's body or absorption by the body may occur, a major part of the radiation is reflected by the chest portion and the abdomen portion, respectively.

[0010] The receiver generates a receiver signal from reflected radiation from the chest region and the abdominal region. The reflected radiation is in general a superposition of radiation reflected by the chest region and radiation reflected by the abdominal region. It is preferred that one and the same antenna is used by the transmitter to transmit radiation and by the receiver to receive reflected radiation.

[0011] In general, the time-dependent distances of the chest portion and the abdomen portion, respectively, from the antenna, $r_1(t)$ and $r_2(t)$, have the following form:

$$r_1(t) = R_1 + x_1(t),$$

$$r_2(t) = R_2 + x_2(t), \quad \text{Eq. (1)}$$

where $R_{1/2}$ are the constant average distances and $x_{1/2}(t)$ are the time-dependent, oscillating components. The method can be applied to an arbitrary periodic breathing pattern. As an example, in the case of a sinusoidal breathing pattern one can write

$$x_1(t) = x_1^0 \sin(2\pi f_{b1} t + \phi)$$

$$x_2(t) = x_2^0 \sin(2\pi f_{b2} t), \quad \text{Eq. (2)}$$

wherein the frequencies of the chest and abdomen motion are f_{b1} and f_{b2} (which are usually the same), the projections of the amplitudes of the chest and abdomen motion along the radial direction are x_1^0 , and x_2^0 , respectively, and ϕ is the phase lag.

[0012] Regardless of the explicit form of $r_1(t)$ and $r_2(t)$, the reflected radiation, which results from a superposition, at the location of the antenna is:

$$u_r = A_1 e^{-i(2\pi f t + \frac{4\pi}{\lambda} r_1(t))} + A_2 e^{-i(2\pi f t + \frac{4\pi}{\lambda} r_2(t))}, \quad \text{Eq. (3)}$$

where $\lambda = \frac{c}{f}$ is the wavelength corresponding to the carrier frequency f , while A_1 and A_2 denote the power amplitudes, which can be regarded as independent of $x_{1/2}(t)$ in the limit where $x_{1/2}^0 \ll R_{1/2}$ (which is always the case for any practical set-up, since the normal chest/abdominal displacement is in the order of a few millimeters) and depend only on the average distances $R_{1/2}$, on the antenna gain pattern and the radar cross sections of the abdomen and the chest, respectively. Eq.(3) holds when the receiver and the transmitter use one and the same antenna (or two antennas with negligible spacing). It should be noted that although the calculations become more complicated if two spaced-apart antennas are used, the inventive method is still feasible.

[0013] The processing device generates an in-phase signal and a quadrature signal based on the receiver signal. As implied by the terms, the quadrature signal has a phase shift of 90° (or $\frac{\pi}{2}$) with respect to the in-phase signal. The wording "based on" includes the possibility that one of the in-phase signal and the quadrature signal is identical to the receiver signal. However, normally, the

processing device performs some signal processing, e.g. mixing and/or filtering, to obtain the in-phase signal from the receiver signal. The quadrature signal is obtained by mixing the 90-shifted transmitted signal with the received signal.

[0014] Furthermore, the processing device determines, from the in-phase signal and the quadrature signal, information characterizing a chest motion and an abdominal motion, at least relative to each other. In this context, the term "chest/abdominal motion" also encompasses the chest/abdominal displacement as a function of time. In other words, the processing device uses the in-phase signal and the quadrature signal obtained from a single receiver signal, which is based on a superposition of reflections by the chest region and the abdominal region to obtain information that are either characteristic of the chest motion and the abdominal motion or characteristic of a relation of these two motions. As will be shown in the following, there are some embodiments where it is possible to monitor both the chest motion and the abdominal motion, even though a single receiver signal is used. In other embodiments, information on a relation of the chest motion and the abdominal motion is determined.

[0015] The inventive method is advantageous since a single transmitter and a single receiver can be used to determine information not only relating to the chest region (or the abdominal region), but to both the chest region and the abdominal region. Therefore, this information may be the basis for diagnosing breathing anomalies which cannot be discovered when monitoring the chest motion alone. However, since a single transmitter and a single receiver is used (usually together with a single antenna), there are no synchronization problems and the design of the radar sensor system employed in the method is relatively simple and cheap.

[0016] According to a preferred embodiment, the information determined by the processing device represents a phase lag between the chest motion and the abdominal motion. "Represent" means that the information can either be the phase lag itself or some quantity that is directly related to the phase lag (e.g. proportional to the phase lag). As already mentioned, the phase lag (or phase difference) between chest motion and abdominal motion provides an indication of certain diseases such as bronchopulmonary dysplasia, obstructive sleep apnea, certain neuromuscular diseases and other abnormalities in the thoraco-abdominal motion.

[0017] As mentioned above, the in-phase signal is normally obtained by preprocessing the receiver signal. Normally, generating the in-phase signal comprises at least partially removing the carrier frequency from the receiver signal. This can be performed by mixing the receiver signal (with the transmitted signal) and subsequently applying a low pass filter. For instance, mixing and passing through a low pass filter may convert the signal of Eq.(3) to the following form:

$$Y = B_1 e^{-i\left(\frac{4\pi}{\lambda} r_1(t)\right)} + B_2 e^{-i\left(\frac{4\pi}{\lambda} r_2(t)\right)}, \quad \text{Eq. (4)}$$

The real part and the imaginary part of Eq.(4) can represent the in-phase signal and the quadrature signal, respectively.

[0018] Preferably, the processing device generates I-Q plane data by relating the in-phase signal to the quadrature signal and determines the phase lag by comparing the I-Q plane data with reference data. In other words, the processing device generates data that correspond to a plot in the I-Q plane, where one axis (I-axis) represents the value of the in-phase signal and the other axis (Q-axis) represents the quadrature signal. The I-Q plane data for different times together form a graph or figure, the shape of which depends on the phase lag. By comparing the data obtained from the in-phase signal and the quadrature signal with reference data representing a known phase lag, the present phase lag can be determined at least approximately. This comparison can be done e.g. by performing a curve fitting procedure. It is understood that in order to perform the comparison, at least a sample of the in-phase signal and the quadrature signal need to be recorded in a memory device, which can be part of the processing device or to which the processing device can be connected. Also, the reference data need to be available from the same memory device or another memory device (e.g. a read-only memory).

[0019] Apart from the phase lag, the shape of the graph depends on the difference of the above-mentioned (constant) average distances $R_{1/2}$, the power amplitudes B_1 , B_2 (which in turn depend on the radar cross sections of the chest region and the abdominal region, the average distances $R_{1/2}$ and the antenna gain pattern) and the chest and abdomen amplitudes $x_{1/2}^0$. For a successful identification of the reference data corresponding to the present phase lag, these

parameters should be known to the processing device. Therefore, it is preferred that the processing device uses an average chest distance from the radar sensor system, an average abdomen distance from the radar sensor system, individual power amplitudes received from the chest and abdomen region and chest and abdomen displacement amplitudes to determine the information representing the phase lag. These average distances may be measured and input to the radar sensor system by a user. Alternatively, the radar sensor system could comprise a distance meter, which could be based e.g. on radar, lidar or ultrasound and which is connected to the processing device. During the initial setup for the measurement, a user could point the distance meter at the chest portion and subsequently at the abdomen portion so that the distances can be directly measured and made available to the processing device. The respective distance from the radar sensor system may in particular be a distance from the antenna used by the receiver (and, usually, by the transmitter). The power amplitudes received from the chest and abdominal regions, $B_{1/2}$ can, for example, be pre-determined in a calibration step where first the abdominal region is shielded and the strength of the reflected signal from the chest region only is measured, and then vice-versa - i.e. chest region is shielded and then the signal from the abdominal region only is measured. The chest and abdomen amplitudes, $x_{1/2}^0$ can be pre-determined in a similar way, by performing a series of measurements by e.g. radar, lidar or ultrasound, and subtracting the average distance from the maximum distance to obtain the displacement amplitude of the respective body part (i.e. chest or abdomen).

[0020] It is preferred that the carrier frequency is less than 3.5 GHz, preferably less than 2.5 GHz. In this regime, it can be assumed that the amplitude of the chest motion and the abdominal motion, respectively, are considerably smaller than the wavelength. For 2.4 GHz, the wavelength is approximately 125 mm, which is an order of magnitude larger than the normal chest/abdominal displacement, which is in the order of a few millimeters. In such a case, it is possible to expand equation (4) to a linear order in $\frac{x_0^i}{\lambda}$, which leads to the following simplification:

$$Y = Y_0 - iB_1 e^{-i\frac{4\pi}{\lambda}R_1} \frac{4\pi}{\lambda} x_1(t) - iB_2 e^{-i\frac{4\pi}{\lambda}R_2} \frac{4\pi}{\lambda} x_2(t), \quad \text{Eq. (5)}$$

with Y_0 a constant, and $x_1(t)$ and $x_2(t)$ the time-varying parts of the chest and abdominal displacements, respectively. Apart from the constant Y_0 , Eq.(5) has two time-varying contributions, which are proportional to the chest displacement and the abdominal displacement, respectively.

[0021] In general, the proportionality factors in Eq.(5) are not purely real or imaginary. However, Eq.(5) represents a linear equation system which may be written in the following form:

$$I(t) = I_c + \alpha x_1(t) + \beta x_2(t) \quad \text{and} \quad Q(t) = Q_c + \gamma x_1(t) + \delta x_2(t)$$

where α , β , γ , δ are known constants which depend on $R_{1/2}$ and $B_{1/2}$, and I_c and Q_c which are constant. If $R_{1/2}$ and $B_{1/2}$ are known, the above linear system of equations for I and Q can be solved to obtain $x_{1/2}(t)$ as linear combinations of I and Q . In a preferred embodiment, the processing device determines the a time-dependent chest displacement and a time-dependent abdominal displacement from the in-phase signal and the quadrature signal. One can then plot $x_1(t)$ against $x_2(t)$ in a Lissajous figure plot to determine the phase lag from the location of the zero points and the dimension of the Lissajous figure. Preferably, the processing device relates the time-dependent chest displacement to the time-dependent abdominal displacement to generate data corresponding to a Lissajous figure and determines the phase lag from the Lissajous figure.

[0022] Under special conditions, solving the linear equation system becomes trivial. If one of the proportional factors in Eq.(5) is real while the other is imaginary, the two displacements can be directly observed by monitoring the in-phase signal and the quadrature signal, respectively. In other words, the following equations have to be fulfilled for this simplification:

$$e^{-i\frac{4\pi}{\lambda}R_1} = \pm i \quad \text{and} \quad e^{-i\frac{4\pi}{\lambda}R_2} = \pm 1 \quad \text{or} \quad R_1 \leftrightarrow R_2$$

[0023] These conditions are fulfilled according to a preferred embodiment, wherein the radar sensor system and the person are positioned so that one of the average chest distance R_1 and average abdomen distance R_2 is equal to

$$(2n + 1) \frac{\lambda}{8}$$

and the other is equal to

$$(2m+2)\frac{\lambda}{8}$$

where m, n are integers. Whether or not these conditions are fulfilled can be determined using a distance meter. It should be noted that in the above-mentioned frequency regime, the wavelength is large enough so that these conditions can be fulfilled and verified with relative ease.

[0024] For instance, if $R_1 = (2n+1)\frac{\lambda}{8}$ and $R_2 = (2m+2)\frac{\lambda}{8}$, the in-phase and quadrature signals reduce to

$$I(t) = I_c \pm B_1 \frac{4\pi}{\lambda} x_1(t),$$

$$Q(t) = Q_c \pm B_2 \frac{4\pi}{\lambda} x_2(t), \quad \text{Eq. (6)}$$

where I_c, Q_c are constant offsets (i.e. not time-varying), given by $I_c = \text{Re}(Y_0)$ and $Q_c = \text{Im}(Y_0)$.

[0025] In other words, in this configuration the in-phase signal represents the chest displacement as a function of time only, while the quadrature signal represents the abdominal displacement only. Moreover, there is a simple linear proportionality in the regime of large wavelength. In other words, this embodiment makes it possible not only to determine some relation of the chest motion and the abdominal motion but also to monitor each motion individually, although the method is based on a single receiver signal.

[0026] By subtracting the constant offset and assuming sinusoidal displacements, one can rewrite the in-phase and quadrature signal in the following form:

$$I - I_c = I_0 f(f_{b1}, t, \phi),$$

$$Q - Q_c = Q_0 f(f_{b2}, t, 0), \quad \text{Eq. (7)}$$

where f is an arbitrary periodic function of time with frequency f_b and phase offset ϕ , $I_0 = B_1 \frac{4\pi}{\lambda} x_1^0$ and $Q_0 = B_2 \frac{4\pi}{\lambda} x_2^0$ are the amplitudes of the in-phase and quadrature signals, which can be read-off from the recorded in-phase and quadrature signals, respectively. In the case of sinusoidal breathing $f = \sin(2\pi f_{b1} t + \phi)$. From above, one can see that the ratio of the amplitudes

$\frac{I_0}{Q_0} \propto \frac{x_1^0}{x_2^0}$ depends on the ratio of the chest displacement amplitude and the abdominal displacement amplitude. Thus, the processing device can determine the ratio of the amplitudes of the chest motion and the abdominal motion, if the ratio of the power amplitudes B_1/B_2 is known. This can be determined by measuring the ratio of the radar cross sections of the abdomen and the chest, respectively. The ratio of the chest and abdomen amplitudes can be displayed to a user. One can use this information to deduce if a subject is undergoing pulmonary breathing, characterized by $\frac{x_1^0}{x_2^0} > 1$ or diaphragmatic breathing, characterized by $\frac{x_1^0}{x_2^0} < 1$.

[0027] Further, if one analyses

$$\begin{aligned}\check{I}(t) &= \frac{I - I_c}{I_0} = \sin(2\pi f_{b1}t + \phi), \\ \check{Q}(t) &= \frac{Q - Q_c}{Q_0} = \sin(2\pi f_{b2}t), \quad \text{Eq. (8)}\end{aligned}$$

and plots $\check{I}$ vs $\check{Q}$, one will obtain a so-called Lissajous figure. In a preferred embodiment, the processing device relates the in-phase signal to the quadrature signal, optionally after normalizing each signal and removing a constant offset, to generate I-Q plane data corresponding to a Lissajous figure and determines the phase lag from dimensions of the Lissajous figure. It is a well-known procedure to deduce the phase lag of two sinusoidal oscillations from a Lissajous figure. Normally, one can assume $f_{b1} = f_{b2}$. In this case, depending on the phase lag, the Lissajous figure corresponds to a circle, an ellipse or a line and the phase lag can be determined from the quotient of the total width of the Lissajous figure and the distance of its zero points. In the case of different $f_{b1} \neq f_{b2}$, the phase lag can also be deduced from the Lissajous figure, although the procedure may be more complicated. Of course, the phase lag can also be determined without normalizing (i.e. dividing by I_0 or Q_0 , respectively) and removing the constant offset I_c, Q_c .

[0028] The previously discussed embodiments are based on the time evolution of the in-phase signal and the quadrature signal as such. According to another embodiment, the information representing the phase lag is determined from a cross-correlation function of the in-phase signal and the quadrature signal in the

limit where the wavelength is much larger than the typical breathing amplitude. The advantage of this embodiment is that it works extremely well even in the presence of random noise, since the noise effects cancel out in the cross correlation function (the integral over noise averages out to zero).

[0029] The cross correlation between the in-phase signal and the quadrature signal can be defined as

$$C(\tau) = \lim_{T \rightarrow \infty} \frac{1}{2T} \int_{-T}^T \check{I}(t) \check{Q}(t + \tau) dt, \quad \text{Eq.(9)}$$

[0030] In general, the in-phase and quadrature signals are linear combinations (or linear superpositions) of $x_1(t)$ and $x_2(t)$. The cross-correlation function, which relates in-phase and quadrature signals, can then be calculated and from the locations of its extrema and/or its zero points, it is possible to extract the phase lag between the chest and the abdomen. While this is applicable for any value of R_1 and R_2 , in the specific case where the average distances $R_{1/2}$ are chosen such that the in-phase signal records the chest displacement only and the quadrature signal the abdomen displacement only, the cross correlation function reduces to the following Eq.(10) and the analysis becomes somewhat simpler. One can again assume $f_{b1} = f_{b2}$ (abdomen and chest frequency are the same). Using trigonometric identities, one can show for a specific example of sinusoidal breathing that

$$C(\tau) = \cos(\phi - 2\pi f_{b1} \tau) \quad \text{Eq.(10)}$$

[0031] This corresponds to a cosine function with frequency f_{b1} and a phase shift ϕ . Therefore, by looking at the position of the extrema in the cross-correlation function represented by Eq.(10), one can read off the values of the breathing frequency f_{b1} and of the phase lag ϕ . Alternatively, one could look at the position of the zero points or a combination of extrema and zero points. Generally speaking, the phase lag may preferably be determined from the position of at least two points of the cross-correlation function, each of which is an extremum or a zero point. While Eq.(10) holds for sinusoidal breathing, a similar analysis can be performed for any periodic in-phase and quadrature signals, and the phase shift ϕ can be extracted from the cross-correlation function in a similar way.

[0032] The great advantage of evaluating the cross-correlation function is that if the in-phase signal and the quadrature signal contain a large amount of random (i.e. uncorrelated) noise, the contribution of this noise to the cross-correlation function averages out to zero. Therefore, the cross-correlation function will be a smooth curve even if the original signals are impaired by noise.

[0033] The present invention also provides a radar sensor system for breathing monitoring, comprising:

- a transmitter for irradiating a chest region and an abdominal region of a person with radar radiation having a carrier frequency;
- a receiver for generating a receiver signal from reflected radiation from the chest region and the abdominal region;
- a processing device adapted for generating an in-phase signal and a quadrature signal based on the receiver signal, and determining, from the in-phase signal and the quadrature signal, information characterizing a chest motion and an abdominal motion, at least relative to each other.

[0034] These terms have been described above with respect to the inventive method and will not be explained again. Preferred embodiments of the radar sensor system correspond to those of the inventive breathing monitoring method described above.

Brief Description of the Drawings

[0035] Preferred embodiments of the invention will now be described, by way of example, with reference to the accompanying drawings, in which:

Fig. 1 shows a schematic representation of a first embodiment of a radar sensor system and a person;

Fig. 2 is a flow chart illustrating a first embodiment of a method for breathing monitoring;

Fig. 3 shows an example of I-Q plane diagrams for different phase lags;

Fig. 4 is a flow chart illustrating a second embodiment of a method for breathing monitoring;

Fig. 5 shows an example of an I-Q plane diagram with a Lissajous figure;

Fig. 6 is a flow chart illustrating a third embodiment of a method for breathing monitoring;

Fig. 7a shows an example of the time evolution of an in-phase signal and a quadrature signal; and

Fig. 7b shows the cross-correlation function of the signals from the fig. 7a.

Description of Preferred Embodiments

[0036] Fig. 1 schematically shows a setup for a first embodiment of a method for breathing monitoring according to the present invention.

[0037] A first embodiment of a radar sensor system 1 for breathing monitoring, which is used in this method, comprises a transmitter 2 and a receiver 3, which are connected to a single antenna 4. In other words, the transmitter 2 uses the antenna 4 to transmit radar radiation 10 and the receiver 3 uses the same antenna 4 to receive radar radiation 11, 12. Both the transmitter 2 and the receiver 3 are connected to a processing device 5, which comprises hardware components, although some of its functionalities may be realized by software. It also comprises a memory device.

[0038] As shown in fig. 1, a person 20 is positioned opposite the radar sensor system 1. The transmitter 2 simultaneously irradiates a chest region 21 and an abdominal region 22 with a transmitted signal 10 of radar radiation having a carrier frequency of e.g. 10 GHz. The transmitted signal 10 is a continuous wave of the following form:

$$u_t = A_t e^{-i(2\pi f t)}$$

where f is the carrier frequency and A_t is the amplitude of the transmitted signal 10. Alternatively, a frequency-modulated continuous wave could be used. The transmitted signal 10 is reflected by the chest region 21 and the abdominal region 22, whereby a first reflected signal 11 and a second reflected signal 12 are generated. These reflected signals 11, 12 are received by the receiver 3 via the antenna 4.

[0039] The receiver 3 generates a receiver signal from the reflected signals 11, 12 from the chest region 21 and the abdominal region 22. Since both regions 21, 22 undergo an oscillating breathing motion, the time-dependent distances of the

chest portion and the abdomen portion, respectively, from the antenna, $r_1(t)$ and $r_2(t)$, have the following form:

$$r_1(t) = R_1 + x_1(t),$$

$$r_2(t) = R_2 + x_2(t),$$

where $R_{1/2}$ are the constant average distances and $x_{1/2}(t)$ are the time-dependent, oscillating components. The method is applicable to arbitrary periodic breathing motion; however, in order to demonstrate the procedure we consider a particular example of a sinusoidal breathing, where

$$x_1(t) = x_1^0 \sin(2\pi f_{b1} t + \phi)$$

$$x_2(t) = x_2^0 \sin(2\pi f_{b2} t), \quad \text{Eq. (2)}$$

wherein the (usually identical) frequencies of the chest and abdomen motion are f_{b1} and f_{b2} , the amplitudes of the chest and abdomen motion projected along the radial direction are x_1^0 , and x_2^0 , respectively, and the phase lag is ϕ .

[0040] The reflected radiation, which results from a superposition, at the location of the antenna 4 is:

$$u_r = A_1 e^{-i(2\pi f t + \frac{4\pi}{\lambda} r_1(t))} + A_2 e^{-i(2\pi f t + \frac{4\pi}{\lambda} r_2(t))}, \quad \text{Eq. (3)}$$

where $\lambda = \frac{c}{f}$ is the wavelength corresponding to the carrier frequency f , while A_1 and A_2 denote the power amplitudes, which can be regarded as independent of $x_{1/2}(t)$ in the limit where $x_{1/2}^0 \ll R_{1/2}$ and depend only on the average distances $R_{1/2}$, on the antenna gain pattern and the radar cross sections of the abdomen and the chest respectively. For subsequent evaluation, the average distances $R_{1/2}$ may be measured e.g. by a user and input to the processing device 5.

[0041] Fig. 2 is a flowchart illustrating the first embodiment of the inventive method. After transmitting and receiving radiation, the receiver 3 generates a receiver signal which may be proportional to the reflected radiation at the location of the antenna 4. This receiver signal is then provided to the processing device 5, which generates an in-phase signal and a quadrature signal. The in-phase signal may be generated from the receiver signal by mixing and applying a bandpass filter in order to remove the carrier frequency f (and similar high-frequency

components). The resulting signal may have to the following form, with the real part and the imaginary part representing the in-phase signal and the quadrature signal, respectively:

$$Y = B_1 e^{-i\left(\frac{4\pi}{\lambda} r_1(t)\right)} + B_2 e^{-i\left(\frac{4\pi}{\lambda} r_2(t)\right)}, \quad Eq. (4)$$

[0042] For a certain time period, the processing device 5 generates data for the in-phase signal and the quadrature signal and records them in the above-mentioned memory device. This process is represented by a loop in fig. 2. When enough data has been sampled and recorded, the loop terminates and the processing device 5 relates the in-phase signal and the quadrature signal to generate I-Q plane data. These are a series of data points in the I-Q plane, each representing a given point in time. The overall distribution of the data points will correspond to a graph, the shape of which depends on the phase lag ϕ between the chest motion and the abdominal motion. Apart from the phase lag, the shape of the graph depends on the average distances $R_{1/2}$, the power amplitudes $B_{1/2}$, and the chest and abdomen displacement amplitudes $x_{1/2}^0$, wherefore they need to be provided to the processing device 5 for a successful identification of the phase lag.

[0043] In a subsequent step, the processing device 5 determines the phase lag by comparing the I-Q plane data with reference data. The reference data may be available from the above-mentioned memory device. These data may correspond to one of the graphs shown in fig. 3, it which represents a different phase lag. The comparison may include the processing device 5 performing a curve fitting procedure in order to find the most similar graph. Once the most similar graph has been identified, the corresponding phase lag is output to a user. The phase lag may e.g. provide an indication of certain diseases such as bronchopulmonary dysplasia, obstructive sleep apnea and other abnormalities in the thoraco-abdominal motion.

[0044] Fig. 4 is a flowchart of a second embodiment of a method for breathing monitoring according to the invention. The setup is generally as described with reference to fig. 1. However, the carrier frequency is reduced to 2.4 GHz, wherefore the amplitudes of the chest motion and the abdominal motion, respectively, are considerably smaller than the wavelength. Thus, it is possible to

expand equation (4) to a linear order in $\frac{x_0^i}{\lambda}$, which leads to the following simplification:

$$Y = Y_0 - iB_1 e^{-i\frac{4\pi}{\lambda}R_1} \frac{4\pi}{\lambda} x_1(t) - iB_2 e^{-i\frac{4\pi}{\lambda}R_2} \frac{4\pi}{\lambda} x_2(t), \quad \text{Eq. (5)}$$

with Y_0 being a constant. In general Eq.(5) leads to two equations, or rather a linear equation system, for the in-phase and quadrature signal. In general, it is possible to solve this equation system to obtain $x_1(t)$ and $x_2(t)$. Solving the equation system becomes trivial, though, if the relative position of the antenna 4 with respect to the chest portion 21 and the abdominal portion 22 is chosen so that

$$R_1 = (2n + 1) \frac{\lambda}{8} \text{ and } R_2 = (2m + 2) \frac{\lambda}{8}$$

where m, n are integers. Whether or not these conditions are fulfilled can be determined using a distance meter.

[0045] Under this condition, the in-phase and quadrature signals reduce to

$$\begin{aligned} I(t) &= I_c \pm B_1 \frac{4\pi}{\lambda} x_1(t), \\ Q(t) &= Q_c \pm B_2 \frac{4\pi}{\lambda} x_2(t), \quad \text{Eq. (6)} \end{aligned}$$

where I_c, Q_c are constant offsets. In other words, the in-phase signal represents the chest displacement as a function of time only, while the quadrature signal represents the abdominal displacement only.

[0046] The sampling loop in fig. 4 does not differ from the one shown in fig. 2 and therefore will not be explained again. After enough data has been sampled and recorded, the loop ends and the in-phase signal and the quadrature signal are related to generate I-Q plane data. From these data, assuming sinusoidal motions, a constant offset can be identified and removed, i.e. subtracted:

$$\begin{aligned} I - I_c &= I_0 \sin(2\pi f_{b1} t + \phi), \\ Q - Q_c &= Q_0 \sin(2\pi f_{b2} t), \quad \text{Eq. (7)} \end{aligned}$$

where $I_0 = B_1 \frac{4\pi}{\lambda} x_1^0$ and $Q_0 = B_2 \frac{4\pi}{\lambda} x_2^0$ are the amplitudes of the in-phase and quadrature signals. From these amplitudes, with the ratio of B_1 and B_2 known (from a previously measurement of the radar cross-sections of the chest and the

abdomen), the processing device 5 determines the ratio $\frac{x_1^0}{x_2^0}$ of the chest motion and the abdominal motion. This ratio may be output to a user. One can use this information to deduce if a subject is undergoing pulmonary breathing, characterized by $\frac{x_1^0}{x_2^0} > 1$, or diaphragmatic breathing, characterized by $\frac{x_1^0}{x_2^0} < 1$. It should be noted that the calculation of the amplitude ratio is optional and that the method could continue without this step.

[0047] In a subsequent step, the data of the in-phase signal and the quadrature signal are divided by the corresponding amplitude in order to normalize them:

$$\begin{aligned}\check{I}(t) &= \frac{I - I_c}{I_0} = \sin(2\pi f_{b1}t + \phi), \\ \check{Q}(t) &= \frac{Q - Q_c}{Q_0} = \sin(2\pi f_{b2}t), \quad Eq. (8)\end{aligned}$$

[0048] The corresponding plot of $\check{I}$ vs $\check{Q}$ corresponds to a Lissajous figure, which are shown exemplary in fig. 5. In the next two steps, the processing device 5 determines the phase lag from dimensions of the Lissajous figure. One relevant dimension is the total width a of the Lissajous figure while another relevant dimension is the distance b of its zero points. The phase lag ϕ can be determined according to $\phi = \pm \sin^{-1}\left(\frac{b}{a}\right)$ when the top of the ellipse lies in the first quadrant, or by $\phi = \pm \left(\pi - \sin^{-1}\left(\frac{b}{a}\right)\right)$ when the top of the ellipse lies in the second quadrant.

[0049] As indicated by the dashed arrows in Fig.4, some of the steps of the method may be skipped if a user is only interested in the amplitude ratio. In such a case, the constant offset can be removed without relating the in-phase signal and the quadrature signal. After outputting the amplitude ratio, the method may end.

[0050] It should be noted, though, that even if the simplifications of Eq.(6) do not apply, it is possible to obtain $x_1(t)$ and $x_2(t)$ from the linear equation system for $I(t)$ and $Q(t)$ derived from Eq.(5). One can then plot $x_1(t)$ against $x_2(t)$ in a Lissajous figure plot to determine the phase lag from the location of the zeros and the dimension of the Lissajous figure.

[0051] Fig. 6 is a flowchart of a third embodiment of a method for breathing monitoring according to the invention. The setup corresponds to the 2nd

embodiment and therefore will not be explained again. Again, the loop for data sampling and recording does not differ from fig. 2 and 4. However, the in-phase signal and the quadrature signal are recorded for a long enough time T , so that $Tf_{b1} \gg 1$ (corresponding to an observation time of e.g. 10 sec or longer).

[0052] After exiting the loop, the processing device 5 removes an offset and normalizes the in-phase signal and the quadrature signal. The corresponding signals may e.g. have the form shown in fig. 7A. In this case, both signals show a significant amount of noise. Subsequently, the processing device 5 calculates the cross-correlation function of the in-phase signal:

$$C(\tau) = \lim_{T \rightarrow \infty} \frac{1}{2T} \int_{-T}^T \check{I}(t) \check{Q}(t + \tau) dt, \text{ Eq.(9)}$$

[0053] This has the advantage that (random) noise effects cancel out in the cross correlation function. The method works for any kind of breathing pattern. However, for sake of simplicity, we assume a sinusoidal breathing pattern, where in the above-mentioned limit $Tf_{b1} \gg 1$, the correlation function can be approximated as

$$C(\tau) = \cos(\phi - 2\pi f_{b1} \tau) \quad \text{Eq.(10)}$$

[0054] This corresponds to a cosine function with frequency f_{b1} and a phase shift ϕ , which is shown by way of example in fig. 7b. In order to determine the phase lag ϕ , the breathing frequency f_{b1} has to be identified. Therefore, the processing device 5 identifies at least two (characteristic) points of the cross-correlation function, each one of which is an extremum or a zero point. E.g., if the first maximum and the first minimum may be determined, the frequency f_{b1} can be calculated from this information. In the next step, the processing device 5 uses the position of the first maximum at t^* to calculate the phase lag, which is given by $\phi = 2\pi f_b t^*$. The phase lag is then output to a user.

Claims

1. A method for breathing monitoring using a radar sensor system (1), the radar sensor system (1) comprising a transmitter (2), a receiver (3) and processing device (5), the method comprising:
 - 5 the transmitter (2) simultaneously irradiating a chest region (21) and an abdominal region (22) of a person (20) with radar radiation (10) having a carrier frequency;
 - the receiver (3) generating a receiver signal from reflected radiation (11, 12) from the chest region and the abdominal region;
 - 10 the processing device (5) generating an in-phase signal and a quadrature signal based on the receiver signal, and determining, from the in-phase signal and the quadrature signal, information characterizing a chest motion and an abdominal motion, at least relative to each other.
2. The method of claim 1, wherein the transmitter (2) and the receiver (3) use the
15 same antenna (4).
3. The method of claim 1 or 2, wherein the information represents a phase lag between the chest motion and the abdominal motion.
4. The method of any of the preceding claims, wherein generating the in-phase signal comprises at least partially removing the carrier frequency from the
20 receiver signal.
5. The method of any of the preceding claims, wherein the processing device (5) generates I-Q plane data by relating the in-phase signal to the quadrature signal and determines the phase lag by comparing the I-Q plane data with reference data.
- 25 6. The method of any of the preceding claims, wherein the processing device (5) uses an average chest distance from the radar sensor system and an average abdomen distance from the radar sensor system (1), individual power amplitudes received from the chest (21) and abdomen region (22) and chest and abdomen displacement amplitudes to determine the information
30 representing the phase lag.

7. The method of any of the preceding claims, wherein the carrier frequency is less than 3.5 GHz, preferably less than 3 GHz, more preferably less than 2.5 GHz.
8. The method of any of the preceding claims, wherein the processing device (5) determines a time-dependent chest displacement and a time-dependent abdominal displacement from the in-phase signal and the quadrature signal.
9. The method of any of the preceding claims, wherein, the processing device (5) relates the time-dependent chest displacement to the time-dependent abdominal displacement to generate data corresponding to a Lissajous figure and determines the phase lag from the Lissajous figure.
10. The method of any of the preceding claims, wherein the radar sensor system (1) and the person (20) are positioned so that one of the average chest distance and average abdomen distance is equal to

$$(2n + 1) \frac{\lambda}{8}$$

and the other is equal to

$$(2m + 2) \frac{\lambda}{8}$$

where λ is the wavelength corresponding to the carrier frequency and where m, n are integers

11. The method of any of the preceding claims, wherein the processing device (5) relates the in-phase signal to the quadrature signal, optionally after normalizing each signal and removing a constant offset, to generate I-Q plane data corresponding to a Lissajous figure and determines the phase lag from dimensions of the Lissajous figure.
12. The method of any of the preceding claims, wherein the information representing the phase lag is determined from a cross-correlation function of the in-phase signal and the quadrature signal.
13. The method of any of the preceding claims, wherein the phase lag is determined from the position of at least two points of the cross-correlation function, each of which is an extremum or a zero point.

14. A radar sensor system (1) for breathing monitoring, comprising:

a transmitter (2) for simultaneously irradiating a chest region (21) and an abdominal region (22) of a person (20) with radar radiation having a carrier frequency;

5 a receiver (3) for generating a receiver signal from reflected radiation from the chest region (21) and the abdominal region (22);

a processing device (5) adapted for generating an in-phase signal and a quadrature signal based on the receiver signal, and determining, from the in-phase signal and the quadrature signal, information characterizing a chest
10 motion and an abdominal motion, at least relative to each other.

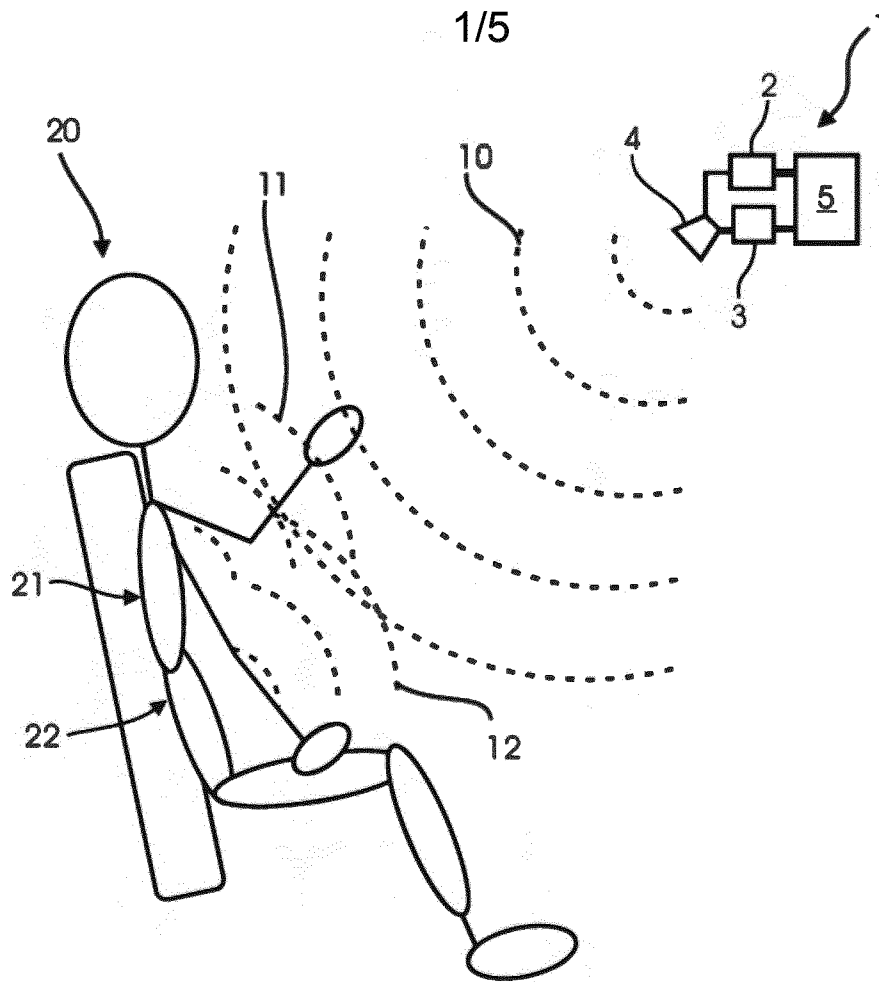


Fig. 1

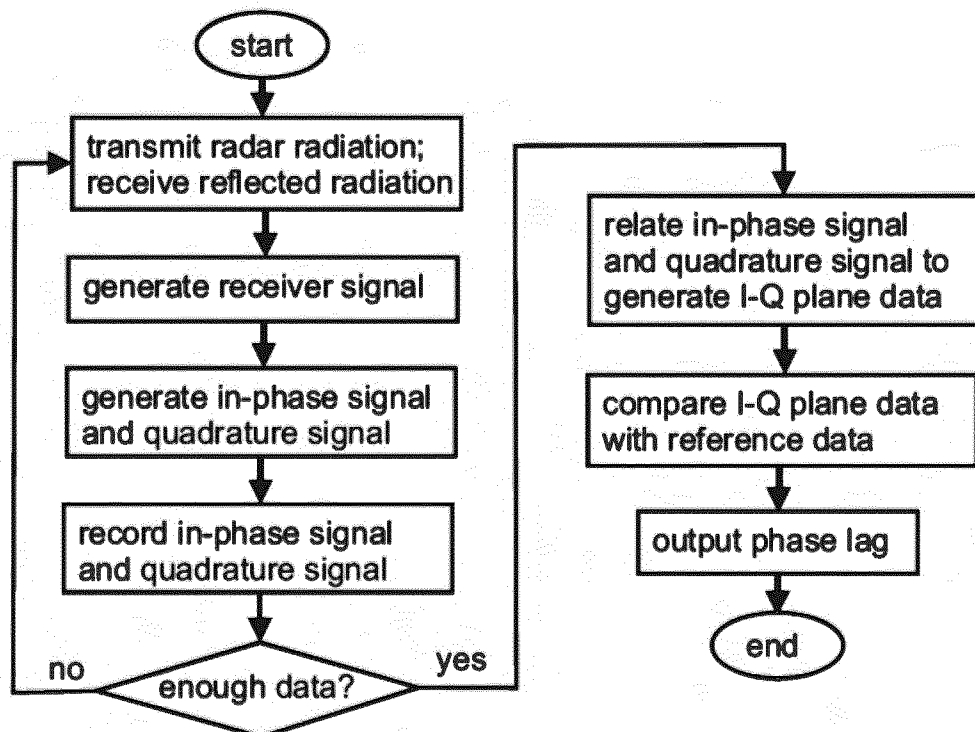


Fig. 2

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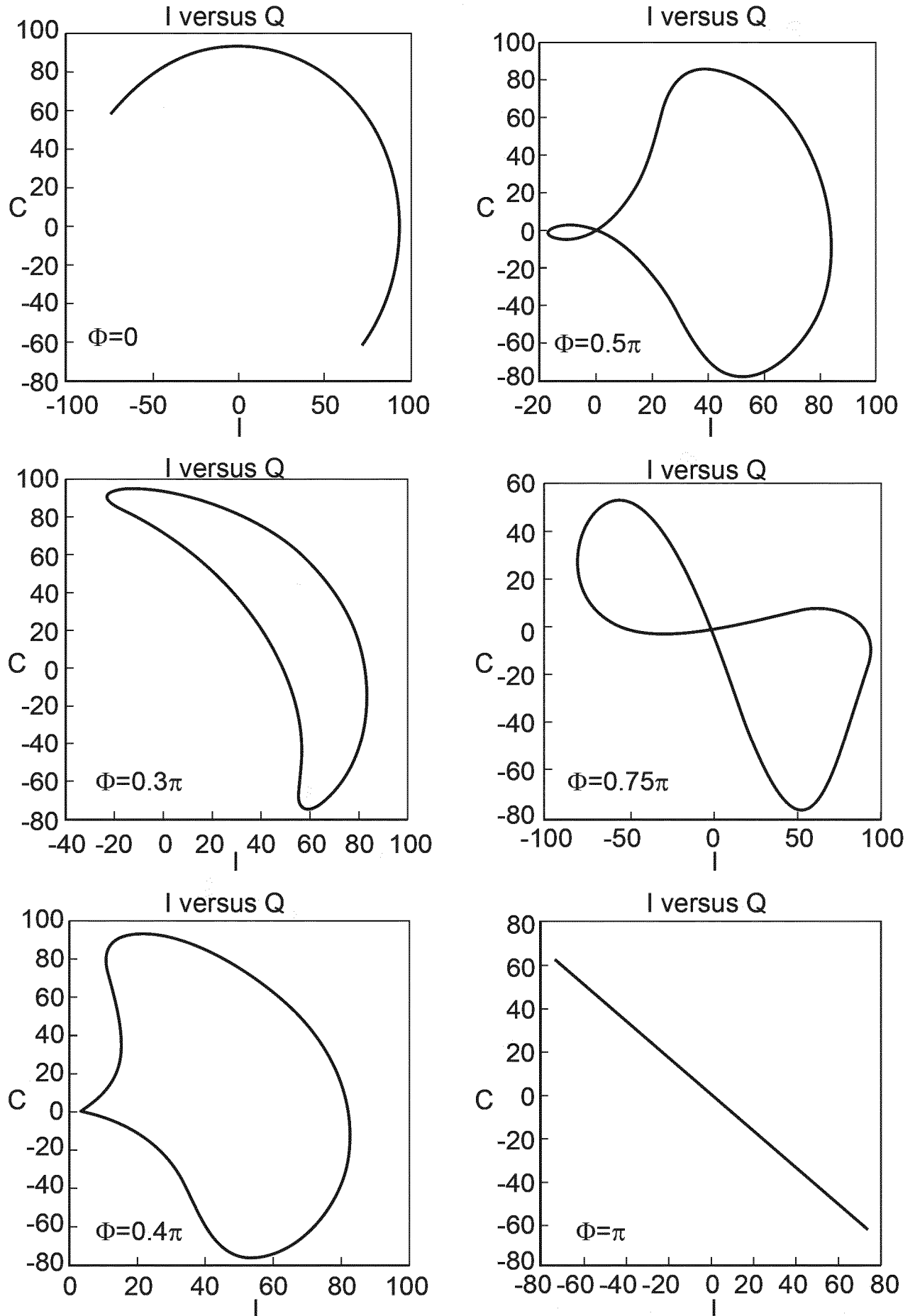


Fig. 3

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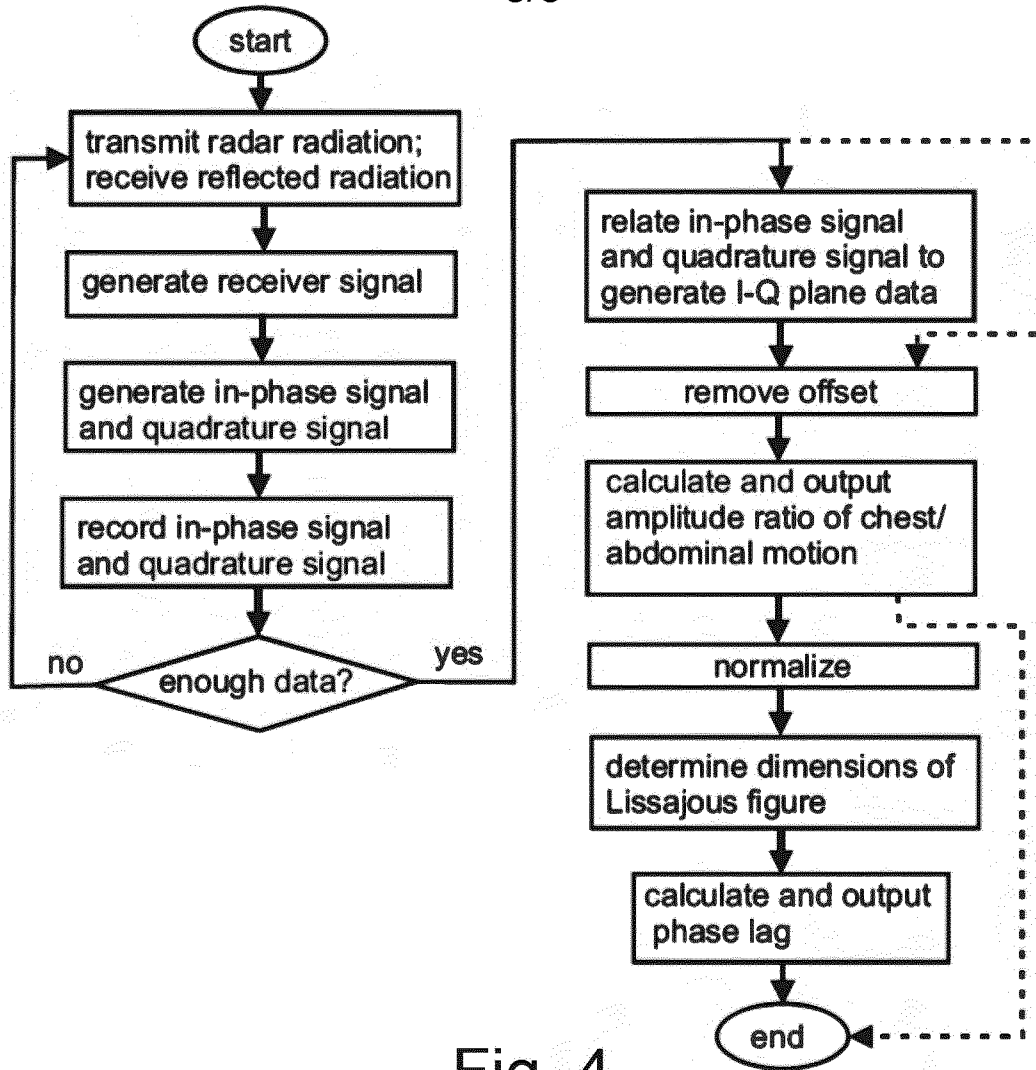


Fig. 4

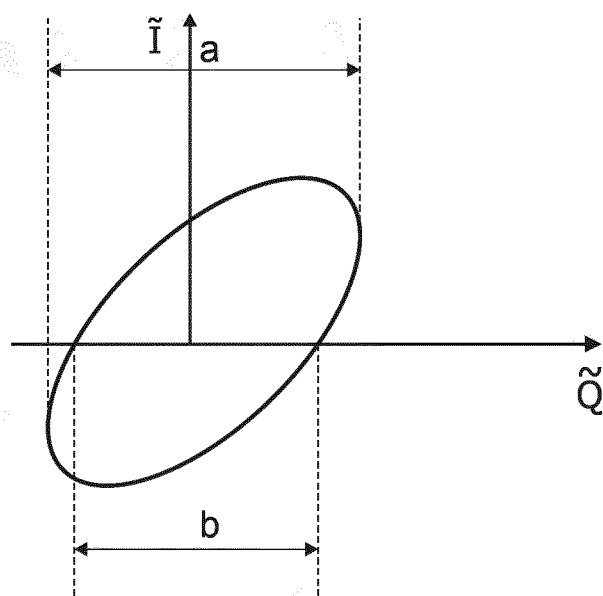


Fig. 5

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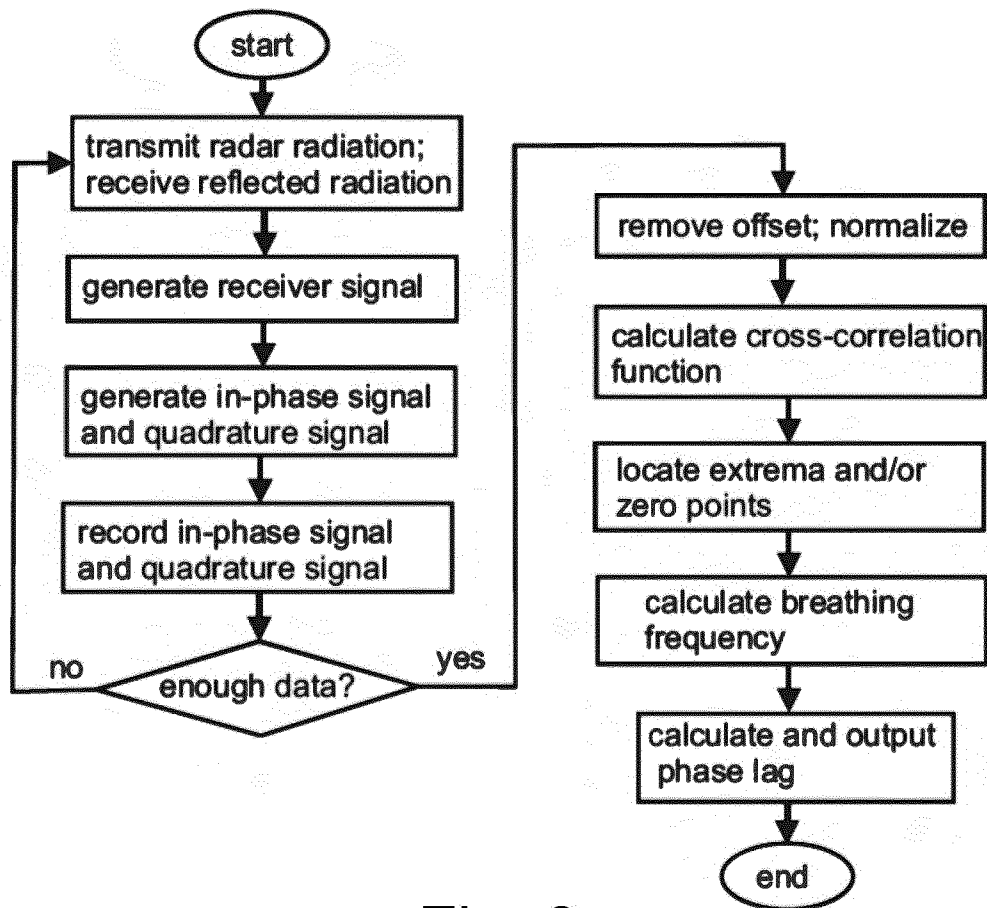


Fig. 6

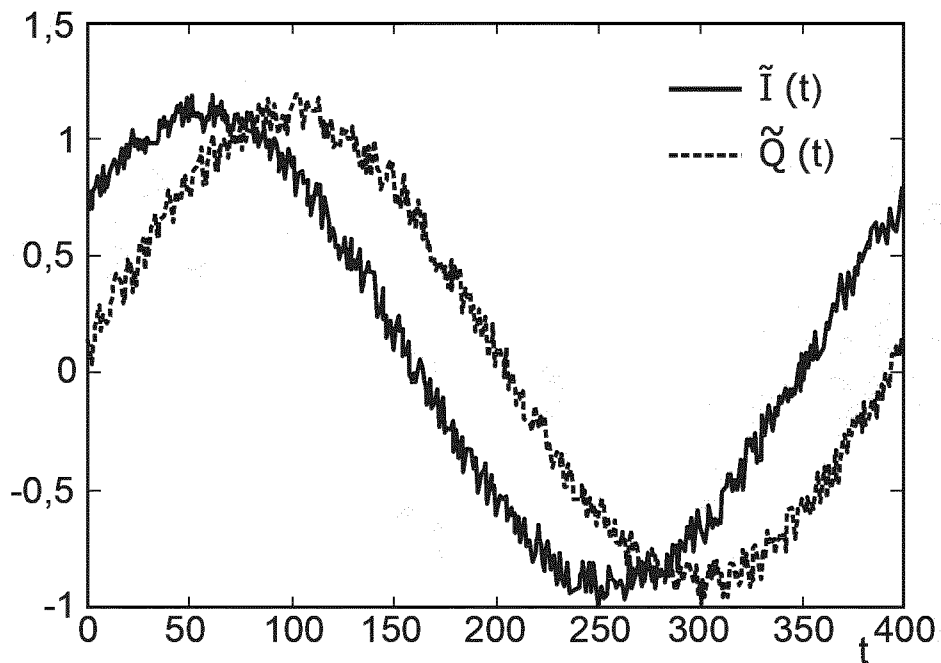


Fig. 7a

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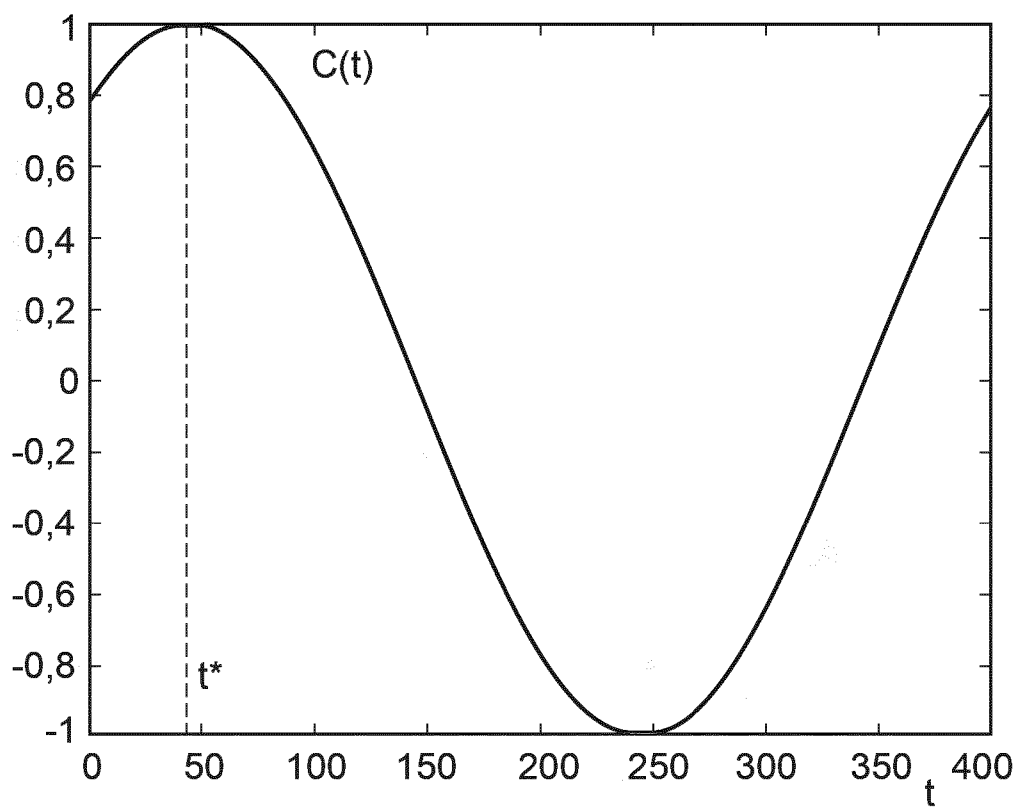


Fig. 7b

INTERNATIONAL SEARCH REPORT

International application No
PCT/EP2018/058322

A. CLASSIFICATION OF SUBJECT MATTER INV. G01S7/35 G01S13/88 G01S7/40 A61B5/0205 A61B5/05 A61B5/08 A61B5/113 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) 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 practicable, search terms used) EPO-Internal, INSPEC, WPI Data								
C. DOCUMENTS CONSIDERED TO BE RELEVANT <table border="1" style="width: 100%; border-collapse: collapse;"> <thead> <tr> <th style="width: 10%; padding: 5px;">Category*</th> <th style="width: 70%; padding: 5px;">Citation of document, with indication, where appropriate, of the relevant passages</th> <th style="width: 20%; padding: 5px;">Relevant to claim No.</th> </tr> </thead> <tbody> <tr> <td style="text-align: center; vertical-align: top; padding: 5px;">X</td> <td style="padding: 5px;"> US 2010/152600 A1 (DROITCOUR AMY [US] ET AL) 17 June 2010 (2010-06-17) abstract; figure 1A paragraphs [0031], [0036], [0044], [0045], [0056] paragraphs [0192], [0246], [0247]; figure 7 paragraphs [0274] - [0277], [0289] - [0291], [0300] paragraphs [0438] - [0453], [0456]; figures 11,12 paragraph [0528] paragraphs [0604], [0645]; figure 17 paragraphs [0673] - [0674]; figure 36A ----- -/-- </td> <td style="text-align: center; vertical-align: top; padding: 5px;">1-14</td> </tr> </tbody> </table>			Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.	X	US 2010/152600 A1 (DROITCOUR AMY [US] ET AL) 17 June 2010 (2010-06-17) abstract; figure 1A paragraphs [0031], [0036], [0044], [0045], [0056] paragraphs [0192], [0246], [0247]; figure 7 paragraphs [0274] - [0277], [0289] - [0291], [0300] paragraphs [0438] - [0453], [0456]; figures 11,12 paragraph [0528] paragraphs [0604], [0645]; figure 17 paragraphs [0673] - [0674]; figure 36A ----- -/--	1-14
Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.						
X	US 2010/152600 A1 (DROITCOUR AMY [US] ET AL) 17 June 2010 (2010-06-17) abstract; figure 1A paragraphs [0031], [0036], [0044], [0045], [0056] paragraphs [0192], [0246], [0247]; figure 7 paragraphs [0274] - [0277], [0289] - [0291], [0300] paragraphs [0438] - [0453], [0456]; figures 11,12 paragraph [0528] paragraphs [0604], [0645]; figure 17 paragraphs [0673] - [0674]; figure 36A ----- -/--	1-14						
☒ Further documents are listed in the continuation of Box C. ☒ See patent family annex.								
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Date of the actual completion of the international search 8 May 2018	Date of mailing of the international search report 18/05/2018							
Name and mailing address of the ISA/ European Patent Office, P.B. 5818 Patentlaan 2 NL - 2280 HV Rijswijk Tel. (+31-70) 340-2040, Fax: (+31-70) 340-3016	Authorized officer van Norel, Jan							

INTERNATIONAL SEARCH REPORT

International application No

PCT/EP2018/058322

C(Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT		
Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	CHANGZHAN GU ET AL: "Antenna array technology for radar respiration measurement in motion-adaptive lung cancer radiotherapy", BIOMEDICAL WIRELESS TECHNOLOGIES, NETWORKS, AND SENSING SYSTEMS (BIOWIRELESS), 2012 IEEE TOPICAL CONFERENCE ON, IEEE, 15 January 2012 (2012-01-15), pages 21-24, XP032153322, DOI: 10.1109/BIOWIRELESS.2012.6172731 ISBN: 978-1-4577-1135-0 abstract paragraph [00II]; figure 1 -----	1-5,7,8, 14

INTERNATIONAL SEARCH REPORT

Information on patent family members

International application No

PCT/EP2018/058322

Patent document cited in search report	Publication date	Patent family member(s)	Publication date
US 2010152600 A1	17-06-2010	US 2010152600 A1	17-06-2010
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		US 2010249633 A1	30-09-2010
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