(12) INTERNATIONAL APPLICATION PUBLISHED UNDER THE PATENT COOPERATION TREATY (PCT)

(19) World Intellectual Property Organization
International Bureau

(43) International Publication Date
05 October 2023 (05.10.2023)

WIPO | PCT

(10) International Publication Number
WO 2023/187323 A1

(51) International Patent Classification:
A61B 5/0507 (2021.01) *A61B 5/08* (2006.01)
A61B 5/00 (2006.01) *A61B 5/11* (2006.01)

(21) International Application Number: PCT/GB2023/050697

(22) International Filing Date: 20 March 2023 (20.03.2023)

(25) Filing Language: English

(26) Publication Language: English

(30) Priority Data:
2204372.3 28 March 2022 (28.03.2022) GB

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(81) Designated States *(unless otherwise indicated, for every kind of national protection available)*: AE, AG, AL, AM, AO, AT, AU, AZ, BA, BB, BG, BH, BN, BR, BW, BY, BZ, CA, CH, CL, CN, CO, CR, CU, CV, CZ, DE, DJ, DK, DM, DO, DZ, EC, EE, EG, ES, FI, GB, GD, GE, GH, GM, GT, HN, HR, HU, ID, IL, IN, IQ, IR, IS, IT, JM, JO, JP, KE, KG, KH, KN, KP, KR, KW, KZ, LA, LC, LK, LR, LS, LU, LY, MA, MD, MG, MK, MN, MU, MW, MX, MY, MZ, NA, NG, NI, NO, NZ, OM, PA, PE, PG, PH, PL, PT, QA, RO, RS, RU, RW, SA, SC, SD, SE, SG, SK, SL, ST, SV, SY, TH, TJ, TM, TN, TR, TT, TZ, UA, UG, US, UZ, VC, VN, WS, ZA, ZM, ZW.

(84) Designated States *(unless otherwise indicated, for every kind of regional protection available)*: ARIPO (BW, CV, GH, GM, KE, LR, LS, MW, MZ, NA, RW, SC, SD, SL, ST, SZ, TZ, UG, ZM, ZW), Eurasian (AM, AZ, BY, KG, KZ, RU, TJ, TM), European (AL, AT, BE, BG, CH, CY, CZ, DE, DK, EE, ES, FI, FR, GB, GR, HR, HU, IE, IS, IT, LT, LU, LV, MC, ME, MK, MT, NL, NO, PL, PT, RO, RS, SE, SI, SK, SM, TR), OAPI (BF, BJ, CF, CG, CI, CM, GA, GN, GQ, GW, KM, ML, MR, NE, SN, TD, TG).

(54) **Title:** A VITAL SIGNS MONITORING SYSTEM

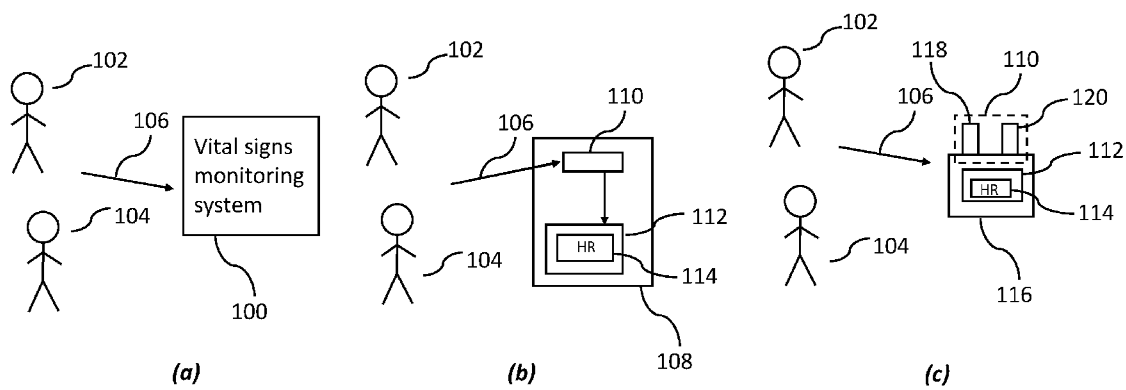

Figure 1

(57) **Abstract:** A vital signs monitoring system configured to estimate the heart rates of two or more subjects using a plurality of subcarriers of a radio frequency (RF) signal.

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WO 2023/187323 A1

Published:

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

A VITAL SIGNS MONITORING SYSTEM

The present disclosure relates to a vital signs monitoring system. In particular, the present disclosure relates to a vital signs monitoring system for estimating the heart rates of two or more subjects.

BACKGROUND

Vital signs, such as heart and respiration rate, are important markers of personal health and well being.

Heart disease is one of the biggest killers worldwide. The UK alone has 7.6 million patients suffering from heart disease with more than 30,000 out of hospital heart attacks and a survival rate of 1 in 10. Moreover, an estimated total of 12 million people in UK suffer from lung disease. The UK has one of the highest incidences of death rates due to cardiovascular and lung disease at 27.9% and 20.1%, respectively.

The COVID-19 pandemic has further exacerbated the statistics related to lung diseases and there is a requirement for remote monitoring of COVID-19 patients due to impending acute respiratory distress syndrome. Therefore, continuous remote monitoring of heart and respiration rates is an urgent requirement of healthcare services to fulfil the demands of future smart homes and health care centres.

Traditional respiration and heart rate monitors require wearable breathing belts and heart rate sensors. Pulse oximeters have been used for continuous measurement of heart rates in medical internet of things monitoring systems. Furthermore, polysomnographs and implantable electrocardiogram devices have been proposed for vital signs monitoring. Camera based systems which can detect heart rates from processing of images using computer vision techniques have been presented before.

Such wearable or implantable systems are intrusive and can only measure the vital signs of a single individual. Camera-based systems may be used to measure the heart rates of multiple subjects (for example as presented in US20160345832A1) but they are not ubiquitous and suffer from ambient light issues.

A system for measurement of multiple heart rates has been proposed in US20200300972A1, however, the system is limited to 60GHz WiFi with lower frequencies being presented as unsuitable. Furthermore, as acknowledged in US20200300972A1 there are substantial challenges in using 60GHz WiFi for multi-subject heart rate detection including attenuation issues and reflection issues.

SUMMARY

It is desirable to provide a system for enabling the detection of the heart rates of two or more individuals, that is not subject to the aforementioned shortcomings of wearable, implantable, camera-based systems or 60GHz WiFi systems.

According to a first aspect of the disclosure there is provided a vital signs monitoring system configured to estimate the heart rates of two or more subjects using a plurality of subcarriers of a radio frequency (RF) signal.

Optionally, the RF signal is pre-processed prior to estimation of the heart rates to filter out portions of the RF signal unrelated to heart rates and/or to pass portions of the RF signal related to heart rates.

Optionally, the RF signal is pre-processed prior to estimation of the heart rates to filter out frequencies unrelated to heart rates and/or to pass frequencies related to heart rates.

Optionally, the RF signal is pre-processed prior to estimation of the heart rates to filter out frequencies related to respiration rates.

Optionally, the RF signal is a WiFi signal or a 5G signal.

Optionally, the RF signal is a 6G signal.

Optionally, the WiFi signal is a 2.4GHz, 5GHz or 6GHz WiFi signal.

Optionally, vital signs monitoring system comprises a receiver configured to receive the RF signal after its interaction with the two or more subjects, and an estimation system comprising a vital signs determination unit configured to estimate the heart rates of the two or more subjects using plurality of subcarriers of the RF signal received by the receiver.

Optionally, the receiver comprises a first antenna and a second antenna, and each of the first and second antennas are configured to receive the RF signal.

Optionally, the first and/or second antennas are omnidirectional receiving antennas.

Optionally, the estimation system comprises i) a first phase signal circuit configured to determine a first phase signal for each of the subcarriers of the RF signal received at the first antenna, and ii) a second phase signal circuit configured to determine a second phase signal for each of the subcarriers of the RF signal received at the second antenna, wherein the vital signs determination unit is configured to estimate the heart rates using the first and second phase signals.

Optionally, the RF signal comprises 52 subcarriers.

Optionally, the RF signal, is a 5G signal, where the 5G signal is a 5G NR signal.

Optionally, the 5G NR signal comprises 3,300 subcarriers.

Optionally, the estimation system comprises a subtraction unit configured to determine a phase difference signal for each subcarrier by subtracting the first phase signal from the second phase signal for each subcarrier, wherein the vital signs determination unit is configured to estimate the heart rates using the first and second phase signals by using the phase difference signal for each subcarrier to estimate the heart rates.

Optionally, the estimation system comprises a data processing module for processing the phase difference signals prior to estimation of the heart rates.

Optionally the data processing module comprises a data calibration Hampel filter and/or a wavelet interval denoising circuit.

Optionally, the estimation system comprises a first pre-processing circuit configured to pre-process the RF signal prior to estimation of the heart rates by filtering out portions of the phase difference signals of each of the subcarriers that are unrelated to heart rates.

Optionally, the first pre-processing circuit is configured to pre-process the RF signal prior to estimation of the heart rates by filtering out portions of the phase difference signals of each of the subcarriers comprising frequencies unrelated to heart rates.

Optionally, the first pre-processing circuit is configured to pre-process the RF signal prior to estimation of the heart rates by filtering out portions of the phase difference signals of each of the subcarriers comprising frequencies related to respiration rates.

Optionally, the first pre-processing circuit comprises a band stop filter configured to filter out frequencies unrelated to heart rates and/or a band pass filter configured to pass frequencies related to heart rates.

Optionally, the first pre-processing circuit comprises a power spectral density determination unit configured to process the phase difference signals to determine a power spectral density of each of the subcarriers.

Optionally, the vital sign determination unit is configured to, for each of the subcarriers, identify n peaks in power spectral density, where n is an integer that is equal to the number of subjects.

Optionally, the vital signs determination unit comprises a frequency selection unit that is configured to, for each of the subcarriers, select the n frequencies having the highest power, thereby identifying the n peaks in power spectral density of each subcarrier.

Optionally, the vital signs determination unit comprises an aggregate unit configured to aggregate the power spectral density peaks from the n frequencies selected by the frequency selection unit of each of the subcarriers.

Optionally, the vital signs monitoring system comprises a peak determination unit configured to determine the maximum n peaks from the aggregated frequencies as provided by the aggregate unit.

Optionally, each subcarrier is accumulated at the receiver over a time period.

Optionally, the vital signs monitoring systems is configured to estimate respiration rates of the two or more subjects using the plurality of subcarriers of the RF signal.

Optionally, the RF signal is pre-processed prior to estimation of the respiration rates to filter out portions of the RF signal unrelated to respiration rates and/or to pass portions of the RF signal related to respiration rates.

Optionally, the RF signal is pre-processed prior to estimation of the respiration rates to filter out frequencies unrelated to the respiration rates and/or to pass frequencies related to respiration rates.

Optionally, the RF signal is pre-processed prior to estimation of the respiration rates to filter out frequencies related to heart rates.

Optionally, the vital signs determination unit is configured to estimate the respiration rate of the two or more subjects using the plurality of subcarriers of the RF signal received by the receiver.

Optionally, the receiver comprises a first antenna and a second antenna, each of the first and second antennas are configured to receive the RF signal, and the estimation system comprises i) a first phase signal circuit configured to determine a first phase signal for each of the subcarriers of the RF signal received at the first antenna, and ii) a second phase signal circuit configured to determine a second phase signal for each of the subcarriers of the RF signal received at the second antenna, wherein the vital signs determination unit is configured to estimate the respiration rates using the first and second phase signals to estimate the respiration rates.

Optionally, the estimation system comprises a subtraction unit configured to determine a phase difference signal for each subcarrier by subtracting the first phase signal from the second phase signal for each subcarrier, wherein the vital signs determination unit is configured to estimate the respiration rates using the first and second phase signals using the phase difference signal for each subcarrier to estimate the respiration rates.

Optionally, the estimation system comprises a data processing module for processing the phase difference signals prior to estimation of the respiration rates and/or the heart rates.

Optionally the data processing module comprises a data calibration Hampel filter and/or a wavelet interval denoising circuit.

Optionally, the estimation system comprises a second pre-processing circuit configured to pre-process the RF signal prior to estimation of the respiration rates by filtering out portions of the phase difference signals of each of the subcarriers that are unrelated to the respiration rates.

Optionally, the second pre-processing circuit is configured to pre-process the RF signal prior to estimation of the respiration rates by filtering out portions of the phase difference signals of each of the subcarriers comprising frequencies unrelated to the respiration rates.

Optionally, the second pre-processing circuit is configured to pre-process the RF signal prior to estimation of the heart rates by filtering out portions of the phase difference signals of each of the subcarriers comprising frequencies related to respiration rates.

Optionally, the second pre-processing circuit comprises a second band pass filter configured to pass frequencies related to the respiration rates.

Optionally, the second pre-processing circuit comprises a second power spectral density determination unit configured to process the phase difference signals to determine a power spectral density of each of the subcarriers.

Optionally, the vital signs determination unit configured to, for each of the subcarriers, identify n peaks in power spectral density, where n is an integer that is equal to the number of subjects.

Optionally, the vital signs determination unit comprises a frequency selection unit that is configured to, for each of the subcarriers, select the n frequencies having the highest power, thereby identifying the n peaks in power spectral density of each subcarrier.

Optionally, the vital signs determination unit comprises an aggregate unit configured to aggregate the power spectral density peaks from the n frequencies selected by the frequency selection unit of each of the subcarriers.

Optionally, the vital signs monitoring system comprises a peak determination unit configured to determine the maximum n peaks from the aggregated frequencies as provided by the aggregate unit.

Optionally, the vital signs monitoring system comprises a transmitter configured to transmit the RF signal.

Optionally, the transmitter comprises a transmitting antenna.

Optionally, the transmitting antenna comprises a directed transmitting antenna.

According to a second aspect of the disclosure there is provided a method of estimating the heart rate of two or more people using the vital signs monitoring system of the first aspect comprising receiving an RF signal, and estimating the two or more heart rates using the plurality of subcarriers of the RF signal.

BRIEF DESCRIPTION OF THE DRAWINGS

The disclosure is described in further detail below by way of example only and with reference to the accompanying drawings in which:

Figure 1(a) is a schematic of a vital signs monitoring system in accordance with a first embodiment of the present disclosure, Figure 1(b) is a schematic of a vital signs monitoring system in accordance with a second embodiment of the present disclosure, Figure 1(c) is a schematic of a vital signs monitoring system in accordance with a third embodiment of the present disclosure;

Figure 2(a) is a graph of an example of an RF signal, Figure 2(b) is an alternative schematic of the vital signs monitoring system as illustrated in Figure 1(c), Figure 2(c) is a schematic of a specific embodiment of the estimation system in accordance with a fourth embodiment of the present disclosure, Figure 2(d) is a schematic of a specific embodiment of the estimation system in accordance with a fifth embodiment of the present disclosure, Figure 2(e) is a schematic of a specific embodiment of the estimation system in accordance with a sixth embodiment of the present disclosure, Figure 2(f) is a schematic of a specific embodiment of the estimation system in accordance with a seventh embodiment of the present disclosure, Figure 2(g) is a schematic of a specific implementation of the phase signal circuit, Figure 2(h) is a schematic of a specific implementation of the vital signs monitoring system as shown in Figure 1(c), Figure 2(i) is a schematic of a specific implementation of pre-processing circuit;

Figure 3(a) is a schematic of a vital signs monitoring system in accordance with an eighth embodiment of the present disclosure, Figure 3(b) is a schematic of a vital signs monitoring system in accordance with a ninth embodiment of the present disclosure, Figure 3(c) is a schematic of a vital signs monitoring system in accordance with a tenth embodiment of present disclosure, Figure 3(d) is an alternative schematic of the vital signs monitoring system of Figure 3(c), Figure 3(e) is a schematic of a specific implementation of the estimation system in accordance with an eleventh embodiment of the present disclosure, Figure 3(f) is a schematic of a specific implementation of the estimation system in accordance with a twelfth embodiment of the present disclosure, Figure 3(g) is a schematic of a specific implementation of the estimation system in accordance with a thirteenth embodiment of the present disclosure, Figure 3(h) is a schematic of a specific embodiment of the estimation system in accordance with a fourteenth embodiment of the present disclosure;

Figure 4(a) is a schematic of a specific embodiment of the estimation system in accordance with a fifteenth embodiment of the present disclosure, Figure 4(b) is a schematic of a specific embodiment of the estimation system in accordance with a sixteenth embodiment of the present disclosure, Figure 4(c) is a schematic of a specific embodiment of the estimation system in accordance with a seventeenth embodiment of the present disclosure;

Figure 5(a) is a schematic of a specific embodiment of the estimation system in accordance with an eighteenth embodiment of the present disclosure, Figure 5(b) is a schematic of a specific embodiment of the estimation system in accordance with a nineteenth embodiment of the present disclosure;

Figure 6(a) is a schematic of a vital signs monitoring system in accordance with a twentieth embodiment of the present disclosure; Figure 6(b) is a schematic illustrating the operation of the frequency selection unit, Figure 6(c) is a schematic of an example embodiment of the aggregate unit, Figure 6(d) is a schematic of the peak determination unit;

Figure 7(a) is a graph showing the phase difference signals for five subcarriers, Figure 7(b) is a graph showing the CSI signal;

Figure 8(a) is a schematic of an experimental setup that was used to test the system, Figure 8(b) is a photograph of the room set up (front view), Figure 8(c) is a photograph of grounded truth sensors, and Figure 8(d) is a photograph of the room set up (right view);

Figure 9(a) is a graph showing the power spectral density of the subcarriers and the algorithm estimation which is used to determine the heart rate of the three subjects (labelled A, B and C), for a first experiment; Figure 9(b) is a graph showing the power spectral density of the subcarriers and the algorithm estimation which is used to determine the heart rate of the three subjects (labelled A, B and C), for a second experiment; Figure 9(c) is a graph showing the power spectral density of the subcarriers and the algorithm estimation which is used to determine the breathing rate of the three subjects (labelled A, B and C), for a third experiment; Figure 9(d) is a graph showing the power spectral density of the subcarriers and the algorithm estimation which is used to determine the breathing rate of the three subjects (labelled A, B and C), for a fourth experiment;

Figure 10(a) is a graph showing the average accuracy estimates for the heart rate and breathing rate experiments, Figure 10(b) is a graph showing how accuracies for heart rate estimation vary with the size of the window, Figure 10(c) is another graph showing how accuracies for heart rate estimation vary with the size of the window;

Figure 11 is a schematic of a practical implementation of an alternative implementation of the system of Figure 6(a) in use;

Figure 12 shows graphs of nine subcarriers;

Figure 13 is a graph of the power spectrum versus frequency for a plurality of subcarriers as provided to the aggregate unit;

Figure 14 is a graph of power spectrum versus frequency showing the sum of the peak values per frequency as provided as an output by a first block of the aggregate unit;

Figure 15 is a graph of power spectrum versus frequency showing a processed trace as processed by a second block of the aggregate unit; and

Figure 16 is a graph of the trace of Figure 15 showing estimated heart rate frequencies as determined using the peak determination unit.

DETAILED DESCRIPTION

Contactless WiFi sensing for monitoring vital signs has gained traction as it is non-intrusive and is independent from ambient light and environmental conditions.

A WiFi signal comprises several subcarriers. Known contactless vital sign monitoring systems using subcarrier selection methods select the most relevant subcarriers for detection of vital signs. Most of these systems use the variance in the subcarrier amplitude or the phase difference, such as mean absolute variance, to select the subcarriers with the highest variance.

CardioFi uses a frequency-based methods to select subcarriers based on the existence of a stable frequency overtime for the detection of the heart rate of a single subject [Khamis, A., Chou, C. T., Kusy, B. & Hu, W. CardioFi: Enabling heart rate monitoring on unmodified COTS WiFi devices. In Proceedings of the 15th EAI International Conference on Mobile and Ubiquitous Systems: Computing, Networking and Services, 97–106 (2018)].

Recently proposed techniques by Zeng et al. leverage signal energy ratio obtained by the ratio of respiration energy to the total energy of channel state information (CSI) signals to select subcarriers for multi-subject respiration rate detection [Zeng, Y. et al. A multi-person respiration monitoring system using COTS WiFi devices. In Adjunct Proceedings of the 2020 ACM International Joint Conference on Pervasive and Ubiquitous Computing and Proceedings of the 2020 ACM International Symposium on Wearable Computers, 195–198 (2020); Zeng, Y. et al. Multisense: Enabling multi-person respiration sensing with commodity WiFi. Proc. ACM on Interactive, Mobile, Wearable Ubiquitous Technol. 4, 1–29 (2020)].

Other known techniques for contactless measurements of vital signs include PhaseBeat [Wang, X., Yang, C. & Mao, S. Phasebeat: Exploiting CSI phase data for vital sign monitoring with commodity WiFi devices. In 2017 IEEE 37th International Conference on Distributed Computing Systems (ICDCS), 1230–1239 (IEEE, 2017); Wang, X., Yang, C. & Mao, S. On CSI-based vital sign monitoring using commodity WiFi. ACM Transactions on Comput. for Healthc. 1, 1–27 (2020)], Liu et. al. [Liu, J. et al. Tracking vital signs during sleep leveraging off-the-shelf WiFi. In Proceedings of the 16th ACM International Symposium on Mobile Ad Hoc Networking and Computing, 267–276 (2015)], ViMo [Wang, F., Zhang, F., Wu, C., Wang, B. & Liu, K. R. Vimo: Multi-person vital sign monitoring using commodity millimeter wave radio. IEEE Internet Things J. (2020)], Wital [Gu, Y., Zhang, X., Yan, H., Liu, Z. & Ren, F. Wital: WiFi-based real-time vital signs monitoring during sleep. TechRxiv (2021)], Wang et. al. [Wang, F., Zeng, X., Wu, C., Wang, B. & Liu, K. R. Radio frequency based heart rate variability monitoring. In ICASSP 2021-2021 IEEE International Conference on Acoustics, Speech and Signal Processing (ICASSP), 8007–8011 (IEEE, 2021)], and Ali et. al [Ali, K., Alloulah, M., Kawsar, F. & Liu, A. X. On goodness of WiFi based monitoring of sleep vital signs in the wild. IEEE Transactions on Mob. Comput. (2021)].

However, the above-mentioned systems cannot provide multi-subject heart rate monitoring. Currently, multi-subject monitoring of vital signs is limited to respiration rates only due to higher signal strength of chest movements. The heart beats produce subtle chest movements and detection of heart rates has remained limited to single subjects. The detection and measurement of multi-subject heart rates has remained elusive due to their weak signal strength.

Heart beats generate weaker signals than signals generated by respiration. Furthermore, multiple heart rates, due to their low energy, are difficult to ascertain reliably in a set of subcarriers. Furthermore, different heart rates for multiple subjects can exist in different or same subset of carriers.

Selecting subcarriers with the highest variance may be suitable for capturing a single heart rate with the highest signal to noise ratio when compared with other heart rate signals but it will obfuscate weaker heart signals that may exist in other subcarriers.

Moreover, the frequency-based subcarrier selection method that uses frequency stability is not suitable for signals with multiple heart rate frequencies, since the presence of multiple heart rates in the same frequency band will result in higher variance in frequencies and the proposed techniques, such as CardioFi, will reject such subcarriers.

CardioFi presented heart rate detection using CSI signal phase difference values with omnidirectional antennas and a commodity WiFi device. Furthermore, the authors proposed a frequency spectrum based method for the selection of subcarriers for detection of single subject heart rate. The method used the inverse of variance of peak frequencies obtained at different time windows in the signal to select subcarriers with the least variance in frequency, resulting in better estimates for a single subject heart rate. However, as discussed previously, the method can potentially obfuscate second or third subject heart rates that may be registered as second highest or third highest frequency peaks in the signal.

Figure 1(a) is a schematic of a vital signs monitoring system 100 configured to estimate the heart rates of two subjects 102, 104 using a plurality of subcarriers of a radio frequency (RF) signal 106 and in accordance with a first embodiment of the present disclosure. The RF signal 106 comprises the plurality of subcarriers.

The system 100, and the other systems disclosed herein, can provide non-intrusive continuous vital signs monitoring from a distance without a requirement for a wearable device. Known systems do not provide contactless sensing of heart rate for multiple subjects and commonly rely on wearable devices for measurement of heart rates.

The RF signal 106 may be provided by a transmitter. The transmitter may comprise a transmitting antenna. The transmitting antenna may comprise a directed transmitting antenna.

In the present example, there are two subjects 102, 104, with the vital signs monitoring system 100 being configured to estimate each of their respective heart rates. It will be appreciated that in further embodiments, the system 100 may be configured to estimate the heart rates of more than two subjects 102, 104. For clarity of description, the subsequent examples may include illustrations of only two subjects, however it will be appreciated that the systems described herein may be used to estimate the heart rates of more than two subjects in accordance with the understanding of the skilled person.

The RF signal 106 may be pre-processed prior to estimation of the heart rates to filter out portions of the RF signal 106 that are unrelated to the heart rates. Specifically, the RF signal 106 may be pre-processed prior to estimation of the heart rates to filter out frequencies that are unrelated to heart rates. For example, the RF signal 106 may be pre-processed prior to estimation of the heart rates to filter out frequencies that are related to respiration rates.

The RF signal 106 may be, for example, a WiFi signal or a 5G signal. The 5G spectrum is next generation technology that will be ubiquitous in future smart homes and health care centres, and will transform the landscape for healthcare and remote monitoring of public health. Next generation healthcare monitoring and services is a prime domain for 5G enabled communication.

Advanced RF technologies and RF sensing in particular are next generation technologies with substantial growth prospects. The coming 5G revolution is expected to be one of the biggest changes on horizon and will be ubiquitous in smart homes for internet connectivity and all kinds of communication devices.

In a preferred embodiment of the current disclosure, the system 100 (and other systems as disclosed herein) may use 5G RF sensing for healthcare monitoring. This can allow continuous contactless monitoring of multi-subject vital signs, in particular heart rate in smart homes; adult care centers; and hospitals.

The systems described herein can allow non-invasive monitoring without the use of any wearable devices with their associated battery charging issues. The systems disclosed herein may use the 5G spectrum which will be ubiquitous in future homes for general internet connectivity.

Currently healthcare companies provide heart and breathing rate monitoring using systems that require contact with a patient. The systems as disclosed herein can provide a non-intrusive contactless sensing system which can provide continuous real-time monitoring of vital signs, thereby aiding in patient comfort, and accuracy of measurements.

The 5G spectrum has not been used before for multiple subject vital signs monitoring, since most existing systems are based on WiFi sensing. Furthermore, to the best of our knowledge, embodiments of the present disclosure present the first example of systems using the 5G spectrum for monitoring vital signs.

WiFi sensing been proposed for single subject breathing rate measurements utilising both amplitude and phase variation of Channel State Information (CSI) from WiFi signals [Zeng, Y., Wu, D., Gao, R., Gu, T. & Zhang, D. Fullbreathe: Full human respiration detection exploiting complementarity of CSI phase and amplitude of WiFi signals. Proc. ACM on Interactive, Mobile, Wearable Ubiquitous Technol. 2, 1–19 (2018); Zhang, D., Hu, Y., Chen, Y. & Zeng, B. Breathtrack: Tracking indoor human breath status via commodity WiFi. IEEE Internet Things J. 6, 3899–3911 (2019); Zeng, Y. et al. Farsense: Pushing the range limit of WiFi-based respiration sensing with CSI ratio of two antennas. Proc. ACM on Interactive, Mobile, Wearable Ubiquitous Technol. 3, 1–26 (2019); Wang, X., Yang, C. & Mao, S. Resbeat: Resilient breathing beats monitoring with realtime bimodal CSI data. In GLOBECOM 2017-2017 IEEE Global Communications Conference, 1–6 (IEEE, 2017)].

Multi-subject respiration rate monitoring utilising CSI WiFi signals have also been proposed [Zeng, Y. et al. A multi-person respiration monitoring system using cots WiFi devices. In Adjunct Proceedings of the 2020 ACM International Joint Conference on Pervasive and Ubiquitous Computing and Proceedings of the 2020 ACM International Symposium on Wearable Computers, 195–198 (2020); Zeng, Y. et al. Multisense: Enabling multi-person respiration sensing with commodity WiFi. Proc. ACM on Interactive, Mobile, Wearable Ubiquitous Technol. 4, 1–29 (2020)].

A number of proposed methods utilise CSI amplitude and phase information for heart rate monitoring apart from breathing rate measurements for vital signs monitoring [Wang, X., Yang, C. & Mao, S. Phasebeat: Exploiting CSI phase data for vital sign monitoring with commodity WiFi devices. In 2017 IEEE 37th International Conference on Distributed Computing Systems (ICDCS), 1230–1239 (IEEE, 2017); Khamis, A., Chou, C. T., Kusy, B. & Hu, W. Cardiofi: Enabling heart rate monitoring on unmodified COTS WiFi devices. In Proceedings of the 15th EAI International Conference on Mobile and Ubiquitous Systems: Computing, Networking and Services, 97–106 (2018); Liu, J. et al. Tracking vital signs during sleep leveraging off-the-shelf WiFi. In Proceedings of the 16th ACM International Symposium on Mobile Ad Hoc Networking and Computing, 267–276 (2015); Lee, S., Park, Y.-D., Suh, Y.-J. &

Jeon, S. Design and implementation of monitoring system for breathing and heart rate pattern using WiFi signals. In 2018 15th IEEE Annual Consumer Communications & Networking Conference (CCNC), 1–7 (IEEE, 2018)].

Wang et al. [Wang, F., Zeng, X., Wu, C., Wang, B. & Liu, K. R. Radio frequency based heart rate variability monitoring. In ICASSP 2021-2021 IEEE International Conference on Acoustics, Speech and Signal Processing (ICASSP), 8007–8011 (IEEE, 2021)] have also proposed Radio Frequency (RF) for measuring heart rate variability. However, no known systems use an RF signal for multi-subject heart rate monitoring.

Figure 1(b) is a schematic of a vital signs monitoring system 108 corresponding to a specific implementation of the vital signs monitoring system 100 of Figure 1(a), and in accordance with a second embodiment of the present disclosure. The vital signs monitoring system 108 comprises a receiver 110 configured to receive the RF signal 106 after its interaction with the two or more subjects 102, 104, and an estimation system 112 comprising a vital signs determination unit 114 configured to estimate the heart rates of the two or more subjects 102, 104 using the plurality of subcarriers of the RF signal 106 received by the receiver 110.

In the present illustration, the label “HR” is shown within the vital signs determination unit 114 to illustrate that the heart rates of the two subjects 102, 104 is determined using this component.

It will be appreciated that the heart rate information, comprising at least one of the estimated heart rates, may be stored in a memory element within the vital signs determination unit 114 and/or provided to an external memory element.

Alternatively, or additionally, the heart rate information may be displayed by one or more displays that are part of and/or separate from the system 100, or may otherwise be provided to a user, and in accordance with the understanding of the skilled person.

In further embodiments, one or more of the estimated heart rates may undergo further processing after they are estimated. The further processing may be carried out before or after the heart rate information is stored in a memory element and/or provided to a display.

In further embodiments, one or more of the estimated heart rates may undergo further processing after they are estimated, and without the heart rate information being stored in a memory element and/or provided to a display.

For example, in a specific embodiment, the heart rate information may undergo a process that results in the alerting of a medical service if one or more of the heart rates falls below a certain level. Emergency services may then be alerted to the fall in heart rate without the heart rate information being stored in a memory element and/or provided to a display.

Figure 1(c) is a schematic of a vital signs monitoring system 116 showing a specific embodiment of the receiver 110 in which the receiver 110 comprises two antennas 118, 120, and in accordance with a third embodiment of the present disclosure. Each of the two antennas 118, 120 is configured to receive the RF signal 106. One or both of the antennas 118, 120 may be omnidirectional receiving antennas.

Figure 2(a) is a graph of an example of the RF signal 106, where the RF signal 106 comprises a plurality of subcarriers 200. The subcarriers 200 are sidebands of a carrier wave 202 of the RF signal 106. Subcarriers 200 may be a result of a modulation process applied to the RF signal 106, where the subcarriers 200 are used to transmit information. RF signals 106 transmitted as part of a WiFi or 5G system comprise subcarriers for the transmission of information. The RF signal 106 may, for example, comprise 52 subcarriers.

In a further embodiment, the RF signal 106 may be, for example, a 5G signal, which may be a 5G NR signal. The 5G NR signal may comprise 3,300 subcarriers. The 5G

NR signal may function at higher frequencies than other 5G signals in accordance with the understanding of the skilled person.

In a further embodiment, the RF signal 106 may be, for example, a 6G signal.

In a further embodiment. the RF signal 106 may be for example, a WiFi signal, for example a 2.4GHz, 5GHz or 6GHz WiFi signal.

Figure 2(b) is an alternative schematic of the vital signs monitoring system 116 as illustrated in Figure 1(c). In the present schematic, the RF signal 106 is shown as being received by each of the two antennas 118, 120. The RF signal 106 as detected by the antenna 118 is labelled as 106a, with the RF signal 106 as detected by the antenna 120 is labelled as 106b. It will be appreciated that despite resulting from the detection of the same RF signal 106, the signal as detected by each of the antennas 118, 120 may differ, for example in phase and/or signal strength, as a result of the different positions of the two antennas 118, 120.

Figure 2(c) is a schematic of a specific embodiment of the estimation system 112 in accordance with a fourth embodiment of the present disclosure. The estimation system 112 comprises a phase signal circuit 204 that is configured to determine a phase signal 206a for each of the subcarriers of the RF signal received at the antenna 118, and denoted by numeral 106a. In operation, the phase signal circuit 204 computes the Fast Fourier Transform (FFT), which is a complex signal and provides the phase of the data. The estimation system 112 further comprises a phase signal circuit 207 that configured to determine a phase signal 206b for each of the subcarriers of the RF signal received at the second antenna, and denoted by numeral 106b. Each phase signal comprises the phase information of the subcarrier used to generate the phase signal. The vital signs determination unit 114 is configured to estimate the heart rates using the phase signals 206a, 206b.

The heart rate information may be provided to an external display 208. As discussed previously, the heart rates may otherwise be processed, stored or displayed.

Each subcarrier may be accumulated at the receiver 110 over a time period.

Figure 2(d) is a schematic of a specific embodiment of the estimation system 112 in accordance with a fifth embodiment of the present disclosure. In addition to the features as shown in Figure 2(c), the estimation system 112 of the present embodiment comprises a subtraction unit 210. The subtraction unit 210 is configured to determine a phase difference signal 212 for each subcarrier by subtracting the phase signals 206a, 206b for each subcarrier. The vital signs determination unit 114 is configured to estimate the heart rates using the phase signals 206a, 206b by using the phase difference signal 212 for each subcarrier to estimate the heart rates.

Figure 2(e) is a schematic of a specific embodiment of the estimation system 112 in accordance with a sixth embodiment of the present disclosure. In addition to the features as shown in Figure 2(d), the estimation system 112 of the present embodiment comprises a data processing module 214 for processing the phase difference signals 212 prior to estimation of the heart rates. The data processing module 214 may comprise a data calibration Hampel filter and/or a wavelet interval denoising circuit.

Figure 2(f) is a schematic of a specific embodiment of the estimation system 112 in accordance with a seventh embodiment of the present disclosure. It will be appreciated that the estimation system 112 of Figure 2(f) may comprise any other feature of the estimation systems as described herein in accordance with the understanding of the skilled person.

The estimation system 112 comprises a pre-processing circuit 216 that is configured to pre-process the RF signal 106 prior to the estimation of the heart rates by filtering out portions of the phase difference signals 212 of each of the subcarriers that are unrelated to heart rates.

The pre-processing circuit 216 may be configured to pre-process the RF signal 106 prior to estimation of the heart rates by filtering out portions of the phase difference signals 212 of each of the subcarriers comprising frequencies unrelated to heart rates. The pre-processing circuit 216 may comprise a band stop filter configured to filter out frequencies unrelated to heart rates and/or a band pass filter configured to pass frequencies related to heart rates.

The pre-processing circuit 216 may be configured to pre-process the RF signal 106 prior to estimation of the heart rates by filtering out portions of the phase difference signals 212 of each of the subcarriers comprising frequencies related to respiration rates.

The pre-processing circuit 216 may comprise a power spectral density determination unit configured to process the phase difference signals 212 to determine a power spectral density of each of the subcarriers. The power spectral density is computed by the power spectral density determination unit by calculating the FFT of the signal; then halving and multiplying the signal by two; then the absolute value of the signal is determined; then the values are squared; the values are then divided by the length of the signal and the sampling frequency, thereby resulting in the determination of the power spectral density of each of the sub carries.

It will be appreciated that although in the present embodiment the pre-processing circuit 216 receives the phase difference signals 212, in further embodiments, the pre-processing may be undertaken by the pre-processing circuit 216 prior to the generation of the phase difference signals 212.

The vital sign determination unit 114 may be configured to, for each of the subcarriers, identify n peaks in power spectral density, where n is an integer that is equal to the number of subjects.

Figure 2(g) is a schematic of a specific implementation of the phase signal circuit 204. It will be appreciated that the phase signal circuit 207 may also be implemented using the specific implementation shown in Figure 2(g). The phase signal circuit 204 comprises a FFT computation unit 218 for computing the FFT. The phase signal circuit 204 further comprises a sampling unit 220 for sampling the phase of the signal on a per frame basis. The phase signal circuit 204 further comprises a phase output unit 222 for outputting the phase signal 206a. In an embodiment where the implementation shown in Figure 2(g) is applied to the phase signal circuits 204,207, the subtraction unit 210 may be configured to compute the difference of two phases obtained from the FFT computation units of 204 and 207.

Figure 2(h) is a schematic of a specific implementation of the vital signs monitoring system 116 as shown in Figure 1(c). In the present embodiment, the system 116 further comprises a software defined radio receiver 224 and a gain block 226. In the present embodiment the components within the phase signal circuit 204 are labelled as follows 218a, 220a, 222a; and the components within the phase signal circuit 207 are labelled as follows 218b, 220b, 222b.

Figure 2(i) is a schematic of a specific implementation of pre-processing circuit 216 comprising a power spectral density determination unit 228 which comprises a first component 230 for determining a FFT of the signal; a second component 232 for halving the length of the signal and multiplying it by 2; a third component 234 for determining the absolute value of the signal and then squaring the absolute value; and a fourth component 236 that is configured to divide the signal by the length of the signal and the sampling frequency.

Figure 3(a) is a schematic of a vital signs monitoring system 300 in accordance with an eighth embodiment of the present disclosure. The vital signs monitoring system 300 is configurated to estimate the respiration rates of the two or more subjects 104, 106 using the plurality of subcarriers of the RF signal 106, in addition to their heart rates. The process for estimating heart rates may include one or more of the features as described herein in relation to heart rate estimation. It will be appreciated that in further embodiments, the respiration rates may be measured for different subjects from those whose heart rates are measured.

In the present illustration, the labels “HR” and “BR” are shown within the system 300 to illustrate that the heart rates (HR) and respiration rates (BR for breathing rate) of the two subjects 102, 104 are determined using the system 300.

The RF signal 106 may be pre-processed prior to estimation of the respiration rates to filter out portions of the RF signal 106 that are unrelated to respiration rates. the RF signal 106 may be pre-processed prior to the estimation of the respiration rates to filter out frequencies unrelated to the respiration rates. The RF signal 106 may, for example, be pre-processed prior to estimation of the respiration rates to filter out frequencies that are related to heart rates.

In a specific embodiment the RF signal 106 may be duplicated or otherwise separated into two or more signals so that one of the signals as extracted from the RF signal 106 is processed to estimate respiration rates whilst another of the signals as extracted from the RF signal 106 is processed to estimate the heart rates.

Figure 3(b) is a schematic of a vital signs monitoring system 302 in accordance with a ninth embodiment of the present disclosure. In the present embodiment, the vital signs determination unit 114 is configured to estimate the respiration rate of the subjects 102, 104 using the plurality of subcarriers of the RF signal 106 received by the receiver 110. In the present embodiment, the heart rate and respiration rate information is provided to external components 306, 308 however it will be appreciated that in further embodiments one or both of the heart rate and respiration rate information may be otherwise stored, processed or displayed, as discussed previously for embodiments relating to heart rate information and in accordance with the understanding of the skilled person.

Figure 3(c) is a schematic of a vital signs monitoring system 304 in accordance with a tenth embodiment of present disclosure. Figure 3(d) is an alternative schematic of the vital signs monitoring system 304 of Figure 3(c).

Figure 3(e) is a schematic of a specific implementation of the estimation system 112 which functions substantially as described for the embodiment shown in Figure 2(c), and in accordance with an eleventh embodiment of the present disclosure. In the present embodiment, the vital signs determination unit 114 is configured to estimate the respiration rates using the phase signals 106a, 106b.

Figure 3(f) is a schematic of a specific implementation of the estimation system 112 which functions substantially as described for the embodiment shown in Figure 2(d), and in accordance with a twelfth embodiment of the present disclosure. In the present embodiment, the vital signs determination unit 114 is configured to estimate the respiration rates using the phase signals 106a, 106b by using the phase difference signal 212 for each subcarrier to estimate the respiration rates.

Figure 3(g) is a schematic of a specific implementation of the estimation system 112 which functions substantially as described for the embodiment shown in Figure 2(e), and in accordance with a thirteenth embodiment of the present disclosure. In the present embodiment, the data processing module 214 is for processing the phase difference signals 206a, 206b prior to the estimation of the respiration rates and/or the heart rates. The data processing module 214 may comprise a data calibration Hampel filter and/or a wavelet interval denoising circuit.

Figure 3(h) is a schematic of a specific embodiment of the estimation system 112 in accordance with a fourteenth embodiment of the present disclosure. It will be appreciated that the estimation system 112 of Figure 3(h) may comprise any other feature of the estimation systems as described herein in accordance with the understanding of the skilled person.

The estimation system 112 comprises a pre-processing circuit 310 that is configured to pre-process the RF signal 106 prior to the estimation of the respiration rates by filtering out portions of the phase difference signals 212 of each of the subcarriers that are unrelated to respiration rates. The pre-processing circuit 310 may function as described for the pre-processing circuit 216. Furthermore, the pre-processing circuit 310 may be implemented as shown for the pre-processing circuit 216 in Figure 2(i).

The pre-processing circuit 310 may be configured to pre-process the RF signal 106 prior to estimation of the respiration rates by filtering out portions of the phase difference signals 212 of each of the subcarriers comprising frequencies unrelated to the respiration rates. The pre-processing circuit 310 may comprise a band pass filter configured to pass frequencies related to respiration rates and/or a band stop filter configured to filter out frequencies unrelated to the respiration rates.

The pre-processing circuit 310 may comprise a power spectral density determination unit configured to process the phase difference signals 212 to determine a power spectral density of each of the subcarriers.

It will be appreciated that although in the present embodiment the pre-processing circuit 310 receives the phase difference signals 212, in further embodiments, the pre-processing may be undertaken by the pre-processing circuit 310 prior to the generation of the phase difference signals 212.

The vital sign determination unit 114 may be configured to, for each of the subcarriers, identify n peaks in power spectral density, where n is an integer that is equal to the number of subjects.

Figure 4(a) is a schematic of a specific embodiment of the estimation system 112 in accordance with a fifteenth embodiment of the present disclosure. In the present example the vital signs determination unit 114 is configured to estimate both the respiration rates and the heart rates, and its operation may be understood with reference to the previously described embodiments, and in particular the embodiments presented in Figure 2(c) and Figure 3(e).

Figure 4(b) is a schematic of a specific embodiment of the estimation system 112 in accordance with a sixteenth embodiment of the present disclosure. In the present example the vital signs determination unit 114 is configured to estimate both the respiration rates and the heart rates, and its operation may be understood with reference to the previously described embodiments, and in particular the embodiments presented in Figure 2(d) and Figure 3(f).

Figure 4(c) is a schematic of a specific embodiment of the estimation system 112 in accordance with a seventeenth embodiment of the present disclosure. In the present example the estimation system 112 comprises the pre-processing circuits 216, 310 as described previously. The operation of the present embodiment may be understood with reference to the previously described embodiments, and in particular the embodiments presented in Figure 2(f) and Figure 3(h).

Figure 5(a) is a schematic of a specific embodiment of the estimation system 112 in accordance with an eighteenth embodiment of the present disclosure. Figure 5(b) is a schematic of a specific embodiment of the estimation system 112 in accordance with a nineteenth embodiment of the present disclosure.

Figure 6(a) is a schematic of a vital signs monitoring system 600 in accordance with a twentieth embodiment of the present disclosure. The present example is shown for four subjects 602, 604, 606, 608.

The vital signs monitoring system 600 comprises a CSI collection unit 610 which is used to collect the RF signal 106 as received from each of the antennas 118, 120 and provide the functions of the phase signal circuits 204, 207 and the subtraction unit 210 in determining the phase difference signal 212 (not shown).

The system 600 may comprise a Universal Software Radio Peripheral (USRP) device comprising the receiver 110 working in the 5G spectrum. The CSI values received from the USRP device and received simultaneously form both the antennas 118, 120, are processed for phase information, as discussed previously. Each signal from the receiver antennas 118, 120 may, for example, comprise 52 subcarriers as discussed previously for Figure 2(a). The phase signals from different antennas 118, 120 for the corresponding subcarriers are subtracted to obtain the phase difference signals for each subcarrier, for example, as discussed previously in relation to Figure 2(d). Further description of how the phase difference signals are derived is provided below.

CSI represents the frequency response of the channel through which a wireless signal propagates at a given carrier frequency. Each CSI entry has amplitude and phase information associated with it, which is affected by the multipath propagation effects. The Channel Frequency Response (CFR), H_{sc}, for a subcarrier sc, with frequency f_{sc} over n multiple propagation paths can be given as:

$$H_{sc}(f_{sc}; t) = \sum a_n(t) e^{-j2\pi f_{sc}\tau_n(t)} \tag{1}$$

where $a_n(t)$ is the amplitude attenuation, whereas $\tau_n(t)$ is the propagation delay $\tau_n(t)$ for path n at subcarrier frequency f_{sc}. The CSI^{sc} value for a channel with subcarrier frequency f_{sc} represents an estimate of CFR and can be given as:

$$CSI_{sc} = |H_{sc}| e^{j sin(\angle H_{sc})} \tag{2}$$

The phase and amplitude of CSI values are affected by movements of humans and objects including vitals i.e. breathing motion of the chest. In a given environment and fixed positions for transmitters and receivers, CSI captures changes in channel response, due to any of these motions. OFDM divides a channel into a number of subcarriers. Let a channel with multiple-input multiple-output have N_{Tx} transmitting antennas and N_{Rx} receiving antennas, then the channel can be represented as:

$$r_{sc} = H_{sc}t_{sc} + n_{sc}, \quad sc \in \{1, \ldots, S\}$$

(3)

where $t_{sc} \in R^{N_{Tx}}$ and $r_{sc} \in R^{N_{Rx}}$ are transmitted and received signals, respectively. The H_{sc} represents the CSI matrix for a given subcarrier. The total subcarriers used in the present example are $S = 52$ and the n_{sc} represents the noise vector. The CSI complex values are obtained from the receivers, where CSI packet received is a $N_{Tx} \times N_{Rx} \times 52$ matrix. The $N_{Tx} \times N_{Rx}$ gives the number of stream. A system with two receivers (for example the antennas 118, 120) will have two streams, where each stream can be represented as:

$$CSI^1 = \{CSI_1^1, \ldots, CSI_{52}^1\}$$

(4)

$$CSI^2 = \{CSI_1^2, \ldots, CSI_{52}^2\}$$

(5)

here, CSI_{sc}^a, a represents the antenna and sc represents the subcarrier. In the present disclosure, only the phase of the CSI measurements have been used for vital signs detection (denoted by $\angle CSI$). The phase difference for CSI values of two receiving antennas 1 and 2 are obtained as follows:

$$\Delta\angle CSI_{sc} = \angle CSI_{SC}^1 - \angle CSI_{sc}^2$$

(6)

Figure 7(a) is a graph showing the phase difference signals for five subcarriers 10, 20, 30, 40 and 50. In the present example the CSI phase differences are initially processed to remove any subcarriers containing mostly zero, and only subcarriers with sufficient information content are retained.

The data processing module 214 comprises a data calibration Hampel filter 612 and a wavelet interval denoising circuit 614. The pre-processing circuit 216 comprises a band stop filter 616, a band pass filter 618 and a power spectral density determination unit 620. The pre-processing circuit 310 comprises a band pass filter 622 and a power spectral density determination unit 624.

As shown in Figure 6(a), the heart rates and/or breathing rates may be provided to an application programming interface (API) 625.

The system 600 is an example of a contactless radio frequency sensing system for multi-subject heart rate detection. The system 600 measures vital signs in particular heart rate through non-intrusive contactless RF sensing.

The system 600 can provide contactless sensing and does not require any wearables or invasive methods for monitoring heart rate. The system 600 uses orthogonal frequency-division multiplexing (OFDM) signals which are ubiquitous in home WiFi.

5G technology will be pervasive in future smart homes and will supplement or replace existing communication systems. The system 600 allows us to determine vital signs, such as heart rate with devices which will be ubiquitous and will not require any special wearable instruments. This will allow continuous, round the clock monitoring of patients including older adults in care homes.

In the present embodiment, the system 600 comprises a directional transmitting antenna 626 and two omnidirectional antennas 118, 120 connected to a Universal Software Radio Peripheral Device (USRP) 633 which is operated in the 5G spectrum at 3.75GHz.

As discussed previously, US20200300972A1 indicates that it is necessary to use 60GHz WiFi for multi-subject heart rate detection. This is presented as being due to an improved spatial resolution over lower frequency technology.

The present disclosure provides a system for multi-subject heart rate detection which is not dependent on the use of 60GHz WiFi and has been demonstrated experimentally for 3.75GHz. The system of the present disclosure is enabled by the use of subcarriers of the RF signal in determination of the heart rates of multiple subjects. Furthermore, the aggregation of frequencies belonging to different subcarriers and different heart rates belonging to multiple subjects allows the system disclosed herein to reject spurious frequencies and enhance the weak heart rate signals for multiple subjects which would otherwise have disappeared in noise.

The transmitting antenna 626 sends data which is received at the receiving antennas 118, 120 to obtain phase difference information of CSI (being the RF signal 106 as previously discussed). The CSI phase values are then processed through signal processing algorithms and a frequency selective method to obtain vital signs measurements. In a preferred embodiment, the RF signal 106 is a 5G signal, such that there is provided a multi-subject vital signs monitoring system based on CSI phase values leveraging the 5G spectrum. To the best of our knowledge this is the first instance of multi-subject heart rate monitoring utilising CSI values, along with utilisation of 5G spectrum for vital signs monitoring.

It should be noted that the CSI signal may be referred to as simply “CSI” and may be used interchangeably with them term RF signal 106. It should also be appreciated that after detection by the antennas 118, 120, the RF signal 106 undergoes processing to extract the relevant information for making an estimate of the heart beats and/or respiration rates, as has been discussed previously. Therefore, for convenience, the signal passing through the system and processed at different stages may still be referred to as the CSI signal or the RF signal, even though the signal after each processing step will differ from the signal having been initially detected.

The system 600 uses CSI variations (being variations in the RF signal 106) due to heart beat movement and uses signal processing methods and an estimation algorithm “MultiBeat” to determine the heart rates of multiple subjects from the data.

The data processing module 214 may be configured to remove of mostly zero valued subcarriers and/or detrending of the CSI phase difference data, as follows. Data calibration is performed using the Hampel filter 612. The CSI data may contain DC components introduced by movements. The DC components can affect frequency estimation and spectrum peak detection due to their higher energy or amplitudes. The Hampel filter 612 may be used to calibrate the data by removing the DC component. Using a sufficiently large window size for the Hampel filter 612 can help to remove DC components as well detrend longer-term trends present in the data. In a preferred embodiment, the Hampel filter 612 may have a “large” window size of 4000 samples, with 2000 samples being used for detrending, with a small threshold ranging from 0.05 to 0.01. The detrending may be performed by subtracting the trend from the original signal. It will be appreciated that the processing carried out by the data processing module 214 may be considered as being “pre-processing” as it is carried out before other processes used to estimate the heart rates and/or breathing rates.

The high frequency noise from the RF signal 106 may be obtained by denoising the signal with wavelet based interval denoising 614. The wavelet utilised for denoising may be the Daubechies symmetric ’sym5’ with three vanishing moment and two level thresholding. Figure 7(b) is a graph showing the CSI signal prior to denoising (denoted by trace 700), and after denoising (denoted by trace 702).

After processing by the data processing module 214, the data is provided to the pre-processing circuits 216, 310. The module for heart rate processing (the circuit 216) applies band-stop and band pass filters to obtain the Power Spectral Density (PSD) for each of the subcarriers for the heart rate signal. The module for breathing rate processing (the circuit 310) filters the signal for breathing frequency range and provides the PSDs for each subcarrier for the breathing rates. Both PSDs are then processed by the MultiBeat estimation algorithm to determine the breathing and heart rates.

Multi-subject heart rate estimation may be performed by applying a stop band filter (provided by the band stop filter 616) to the processed signal obtained after detrending and denoising (as provided by the Hampel filter 612 and the wavelet interval denoising circuit 614).

The band stop filter 616 is applied in the breathing frequency range to reduce the effects of harmonics from higher strength breathing movements on the heart signals. After the band-stop filter 616 the RF signal 106 is passed through a band-pass IIR Butterworth filter 618 with the pass band range from 1 to 2Hz for the heart rate corresponding to 60bpm and 120 bpm, respectively.

The fast Fourier Transform (FFT) of the signal obtained from the band pass filter 618 is then computed for each of the subcarriers using the power spectral density determination unit 620. The one sided power spectral density (PSD) of the signal is computed from the FFT using the power spectral density determination unit 620. The PSDs of all the subcarriers are sent to the vital signs determination unit 114 which is configured to use a “MultiBeat” estimation algorithm for selection of frequencies and the estimation of multi-subject heart rate peaks from the selected frequencies of the PSDs of the subcarriers.

The breathing signal has higher energy since the breathing chest movements are higher than subtle beats of the heart rate. The calibrated and wavelet denoised signal obtained from processing the CSI phase difference values is passed through the IIR Butterworth passband filter 622. The filter 622 may be used to obtain frequencies in the 0.1 to 0.5Hz range for breathing. The filtered CSI subcarriers are then converted to frequency domain by computing the FFT of all the subcarriers. The FFT signal is then used to compute the PSD of all the signals. The PSDs obtained from this stage are passed onto the vital signs determination unit 114 which is configured to use the “MultiBeat” estimation algorithm for selection of frequencies and calculation of peaks for the multi-subject breathing rates from the selected frequencies in the PSDs.

The MultiBeat estimation algorithm, may be summarised as follows:

Input: $sc \leftarrow$ Subcarrier index $sc = \{1, \ldots, 52\}$

$\Delta \angle y_{sc} \leftarrow$ Processed, denoised and filtered phase difference CSI vector

Output: f_{vitals} Estimated frequencies for vitals

while $sc \in \{1, \ldots, 52\}$ do

```
/* Compute power spectral density of each  subcarrier */
```
$[power_{sc}, f_{psd}] \leftarrow powerSpectralDensity(\Delta \angle y_{sc})$
```
/* Compute first n peaks of each power spectral density  */
```
$peaks_{sc} \leftarrow maxPeaks\ (psd_{sc}, n = 3)$
```
/* Determine index values for all peaks */
```
$index_{sc} \leftarrow indexValues\ (peaks_{sc})$
```
/* Determine frequency matrix F for selected peaks */
```
$F \leftarrow f_{psd}(u)\ \forall u \in index_{sc}\ \forall sc$
```
/* Determine unique set of frequencies θ */
```
$\theta(F) = \{F_{ij}\}\ \forall i \forall j$

$k = findIndex\ (f_{psd} == \theta)\ \forall \theta$

$e(k) = [\sum_{sc} psd_{sc}(k)] / n_{pks}(k)\ \ \forall k$, where $n_{pks}(k)$ are all peaks at each k

$e_{pks}(n) = maxPeaks(e, n = 3)$

$f_{vitals}(n) \leftarrow f_{psd}(m)\ \forall m \in indexValues(e_{pks})$

It will be appreciated that in further embodiments, one of the pre-processing circuits 216, 310 may be omitted to provide a system for estimation of one of breathing rates or heart rates, in accordance with the understanding of the skilled person.

Figure 6(a) shows a specific implementation of the vital signs determination unit 114 which may be implemented in any of the embodiments as disclosed herein, and in accordance with the understanding of the skilled person.

Figure 6(a) includes specific implementations of the data processing module 214, pre-processing circuits 216, 310, and the vital signs determination unit 114. Any one, or any combination, of these specific implementations may be implemented in any of the other embodiments as disclosed herein to form a new embodiment, in accordance with the understanding of the skilled person.

The following description relates to the processed RF signal 106 received from the pre-processing circuit 216 for the determination of heart rates. It will be appreciated that the same description is applicable for the processed RF signal received from the pre-processing circuit 310 in determination of the breathing rates.

The vital signs determination unit 114 comprises a frequency selection unit 628 that receives the processed RF signal 106 from the pre-processing circuit 216 for the determination of heart rates. As discussed previously, the processed RF signal 106 provided to the vital signs determination unit 114 is the PSDs of all the subcarriers.

The frequency selection unit 628 is configured to select suitable frequencies from each subcarrier. The suitable frequencies are the maximum ‘n’ frequencies in each subcarrier with the highest power (Watt/Hz), where n depends on the number of subjects, whose heart rates are to be determined, e.g. n=3 for three subjects. Figure 6(b) is a schematic illustrating the operation of the frequency selection unit 628. An input of the power (watt/Hz) for each frequency for each subcarrier is provided as an input. The maximum n peaks per subcarrier are determined. And the maximum power (watt/Hz) peaks for n frequencies per subcarrier is provided as an output.

In summary, the frequency selection unit 628 is configured to, for each of the subcarriers, select the n frequencies having the highest power, thereby identifying the n peaks in power spectral density.

The peaks may correspond to same or different frequencies in each subcarrier. A unique set of ‘k’ frequencies are determined, which comprises all unique frequencies from the previous step. The inputs are the power peaks (Watt/Hz) for each frequency in the subcarriers and outputs are the maximum ‘n’ frequencies and their power peaks (Watt/Hz).

The frequency selection unit 628 is configured to provide the power (Watt/Hz) of previously mentioned k frequencies to an aggregate unit 630. The aggregate unit 630 is configured to accumulate the frequencies selected by the frequency selection unit 628, in units of energy (Watt/Hz), to accumulate the signal strengths of weaker heart beat signals across all the subcarriers, according to their respective frequencies and is divided by the number of peaks at each given frequency in ‘k’. The inputs are the peaks from each subcarrier and their respective frequencies and outputs are the aggregated peaks per frequency divided by the number of peaks at each frequency. Figure 6(c) is a schematic of an example embodiment of the aggregate unit 630. In operation, the aggregate unit 630 receives, as an input power Watt/Hz peak values for ‘k’ unique frequencies obtained from previous ‘n’ frequencies per subcarrier; the unit 630 the sums peak values per each frequency k (illustrated by block 630a); the unit then divides the number of peaks at each frequency k (illustrated by block 630b); the unit then provides, as an output aggregated Power (Watt/Hz) peaks for k frequencies.

In summary, the vital signs determination unit 114 comprises the aggregate unit 630 that is configurated to aggregate the power spectral density peaks from the n frequencies selected by the frequency selection unit 628 of each of the subcarriers.

The aggregate unit 630 is configured to provide a signal to a peak determination unit 632. The peak determination unit 632 uses the signal to determine the frequency peaks that are indicative of a heart beat of a subject. The inputs to this function are the aggregated power peaks from previous step, which are divided by the number of peaks at each frequency k and the output provides the power peaks according to the maximum peak values. The peaks may correspond to the number of subjects whose heart rates are to be determined, e.g. highest 3 peaks for 3 subjects. Figure 6(d) is a schematic of the peak determination unit 632. In operation, the unit 632 receives, as an input Aggregated Power (Watt/Hz) peaks for k frequencies; the unit 632 then determines the maximum n peaks overall; the unit 632 then provides, as an output, the maximum Power Watt/Hz peaks for n frequencies from aggregated frequencies of all subcarriers.

In summary, the aggregate unit 630 is configured to aggregate power density peaks per frequency, which may result in k frequencies, where k>=n, as the values of n frequencies may be different for each subcarrier resulting in k frequencies overall during aggregation. In a specific embodiment, each of the peaks is multiplied by the weight of the reciprocal of the number of frequencies obtained at each k.

In summary, the vital signs monitoring system 114 comprises the peak determination unit 632 that is configured to determine the maximum n peaks from the aggregated frequencies as provided by the aggregate unit 630.

The frequency selection and accumulation process can potentially diminish the impact of signal peaks due to spurious frequencies that do not belong to vital signs and may exist in some of the subcarriers. So intuitively, for finding multiple heart rates that may exist in most or all of the carriers, the method described herein can resort to the removal of spurious peaks by informed frequency selection and aggregation.

Figure 8(a) is a schematic of an experimental setup that was used to test the system 600. Figure 8(b) is a photograph of the room set up (front view), Figure 8(c) is a photograph of grounded truth sensors, and Figure 8(d) is a photograph of the room set up (right view).

Figure 8(a) illustrates the dimensions and distance used for measurements. 5 subjects took part in the process for data collection. Twenty samples for different combinations of three subjects were collected separately for breathing and heart rate monitoring, along with twenty samples each for single and two subjects. The two ground truth sensors utilised for the experiment were Polar H10 heart beat sensor and Vernier go direct respiration belt.

The experiments were conducted with a single Universal Software Radio Peripheral (USRP) software defined radio device operating at 3.75GHz at 400 samples/sec. The device utilised one directional transmitting antenna and two VERT2450 omnidirectional antennas. The USRP device was connected to the desktop computer for processing of CSI signals. The desktop computer comprised of Intel(R) Core (TM) i7-7700, 3.60 GHz processors, 16 GB RAM and Ubuntu 18.04 operating system. The USRP is a programmable device and was programmed with the Gnu Radio. The Gnu Radio allows creation of flow diagrams from Orthogonal Frequency-Division Multiplexing (OFDM) blocks. The flow diagrams are then converted to python scripts for running on the USRP device. The OFDM transmitter transmits random signals between 0 - 255, which are then received on the omnidirectional antennae. The CSI complex values are then extracted for each receiver antenna. The complex numbers are then processed to obtain amplitude and phase information. In this experiment, we only used the phase information for vital signs monitoring. The main configuration parameters for the USRP device are given in the table below.

Parameter	Value
Platform	USRP X300/310
OFDM Subcarrier	52
Operating Frequency	3.75GHz
Sampling Rate	400Hz
Transmitter Gain (dB)	40
Receiver Gain (dB)	40

Figure 9(a) is a graph showing the power spectral density of the subcarriers and the algorithm estimation which is used to determine the heart rate of the three subjects (labelled A, B and C), for a first experiment. The algorithm estimation is labelled with numeral 900. The results are as determined from a practical experiment, as described in Figure 8(a).

From the algorithm estimation, there are three peaks denoting the heart rates of the three individuals. Subject A has a heart rate of 1.17Hz, Subject B has a heart rate of 1.36Hz, and Subject C has a heart rate of 1.46Hz.

Also shown for comparison are the results acquired from the ground truth sensors which show Subject A having a heart rate of 1.12Hz, Subject B having a heart rate of 1.33Hz, and Subject C having a heart rate of 1.48Hz.

Figure 9(b) is a graph showing the power spectral density of the subcarriers and the algorithm estimation which is used to determine the heart rate of the three subjects (labelled A, B and C), for a second experiment. The algorithm estimation is labelled with numeral 902. The results are as determined from a practical experiment, as described in Figure 8(a).

From the algorithm estimation, there are three peaks denoting the heart rates of the three individuals. Subject A has a heart rate of 1.07Hz, Subject B has a heart rate of 1.26Hz, and Subject C has a heart rate of 1.36Hz.

Also shown for comparison are the results acquired from the ground truth sensors which show Subject A having a heart rate of 1.10Hz, Subject B having a heart rate of 1.23Hz, and Subject C having a heart rate of 1.33Hz.

Figure 9(c) is a graph showing the power spectral density of the subcarriers and the algorithm estimation which is used to determine the breathing rate of the three subjects (labelled A, B and C), for a third experiment. The algorithm estimation is labelled with numeral 904. The results are as determined from a practical experiment, as described in Figure 8(a).

From the algorithm estimation, there are three peaks denoting the breathing rates of the three individuals. Subject A has a breathing rate of 0.156Hz, Subject B has a breathing rate of 0.234Hz, and Subject C has a breathing rate of 0.35Hz.

Also shown for comparison are the results acquired from the ground truth sensors which show Subject A having a breathing rate of 0.14Hz, Subject B having a breathing rate of 0.25Hz, and Subject C having a breathing rate of 0.34Hz.

Figure 9(d) is a graph showing the power spectral density of the subcarriers and the algorithm estimation which is used to determine the breathing rate of the three subjects (labelled A, B and C), for a fourth experiment. The algorithm estimation is labelled with numeral 906. The results are as determined from a practical experiment, as described in Figure 8(a).

From the algorithm estimation, there are three peaks denoting the breathing rates of the three individuals. Subject A has a breathing rate of 0.16Hz, Subject B has a breathing rate of 0.32Hz, and Subject C has a breathing rate of 0.35Hz.

Also shown for comparison are the results acquired from the ground truth sensors which show Subject A having a breathing rate of 0.143Hz, Subject B having a breathing rate of 0.3Hz, and Subject C having a breathing rate of 0.36Hz.

The technique of the present embodiment uses a frequency selection method instead of selecting subcarriers, and provides good accuracy in multisubject vital sign monitoring. The proposed MultiBeat estimation algorithm provides low error and good accuracy. The error may be determined by:

$$\left\|\frac{f_e - f_r}{f_r}\right\| \tag{7}$$

where f_e is the estimated frequency with the MultiBeat algorithm and f_r is the frequency obtained from the ground truth sensors.

The accuracy is determined from the error rate as:

$$1 - \left\|\frac{f_e - f_r}{f_r}\right\| \tag{8}$$

Figure 10(a) is a graph showing the average accuracy estimates for the heart rate and breathing rate experiments. The average accuracy is determined by averaging out the estimation for all the samples. Using the experimental set up as shown in

Figure 8(a), the average accuracy was found to be 96.2% for single subject heart rate (labelled 1000) and decreases to 94.06% and 92.25% for 2 (labelled 1002) and 3 (labelled 1004) subjects, respectively, as illustrated in Figure 10(a). The decrease in accuracy is due to a cumulative effect of the error:

$$\sum \left\| \frac{f_e - f_r}{f_r} \right\| \qquad (9)$$

for the multi-subject case. Figure 10(a) also illustrates the average accuracies for the estimated breathing rate of 90.48%, 83.95% and 81.09% for single subject (labelled 1006), 2 subjects (labelled 1008) and 3 subjects (labelled 1010), respectively.

The difference in estimated frequencies and ground truth rates is of the order of 10^{-1} and is affected by the resolution of the PSD, however normalisation by much lower rate in breathing frequency estimates results in higher percentage error and lower average accuracy. Similar, to the heart rate estimation, the accuracy for the breathing rate decreases with the increase in subjects due to the cumulative effect of the error.

The heart rate estimation uses a window size of 10 seconds and breathing rate is determined from a window size of 30 seconds. Window size describes the time over which the RF signal 106 is acquired.

The decrease in window size also affects the accuracy of estimated frequencies from the CSI phase values. Figure 10(b) is a graph showing how accuracies for heart rate estimation vary with the size of the window.

Different window sizes of 2.5, 5.0, 7.5 and 10.0 seconds for 1 (labelled 1012), 2 (labelled 1014) and 3 (labelled 1016) subjects were tested. Figure 10(b) shows that the highest average accuracy of 92.25% was obtained for 3 subjects with a window of 10 seconds. The accuracy decreases with decreasing window size and approximately 50% average accuracy is obtained for a window size of 2.5 seconds. This is because smaller window size translates to fewer heart beats, and affects the accuracy of the frequency peaks obtained in the PSDs of all the subcarriers by the proposed algorithm.

Similarly, the single and 2 subject heart rates have higher accuracies and the value decreases to 71.5% and 66.7% for single subject and 2 subject heart rates, respectively. The graph representing single subject heart rate is slightly higher than the two subject and three subject graphs due to the accumulation of error for multiple subjects in estimation of heart rates.

The breathing rate is of the order of 10^{-1}Hz and varies between 0.1Hz and 0.5Hz, due to which higher window size of 30 seconds is utilised for estimation of breathing cycles.

Figure 10(c) illustrates the effect of window size 10, 20 and 30 seconds on the accuracies of breathing rates for single (labelled 1018), two (labelled 1020) and three subjects (labelled 1022). The highest estimation accuracies are obtained for 30 second windows with an accuracy of 81.09% for the three subjects. The highest accuracy of 90.48% is obtained for the single subject over all subjects. The breathing rate also decreases with decreasing window size, as smaller number of breathing cycles are captured by smaller windows. The window size of 10 seconds obfuscates longer breathing cycles and may result in incomplete cycles resulting in much lower accuracies of 54.76% and 37.11% for two and three subjects, respectively, as compared to the accuracies for multi-subject heart rates. The breathing rates for single subject has slightly higher accuracy than the accuracies for two subjects for all the window sizes, due the cumulative effect of the error in the two subject case. Similarly, the two subject breathing rate estimation has slightly higher accuracy than the three subject case for all the window sizes.

Figure 11 is a schematic of a practical implementation of an alternative implementation of the system 600 of Figure 6(a) in use.

Vital signs monitoring including heart rate and respiration rate are two of the most important indicators of public health and wellbeing. There is high demand for contactless sensing of vital signs that do not require intrusive wearable devices to enable continuous contactless monitoring.

Existing solutions using WiFi sensing are limited to single subject heart rate monitoring and use subcarrier selection methods based on high signal variance or stable frequency indicators. Multi-subject heart rates can exist in the same or a different subset of carriers, and therefore subcarrier selection methods do not work well with them.

As disclosed herein, there is provided a novel algorithm for multi-subject monitoring of both heart and breathing rates, which relies on frequency selection method, rather than subcarrier selection.

This is the first time multi-subject heart rates have been sensed and detected with the use of 5G spectrum for vital signs monitoring. The technique accumulates the strength of weak heart signals by selection of frequencies and aggregation, and removes spurious frequency peaks not related to vital signs that may be present in some subcarriers.

The proposed system and algorithm achieve a high average accuracy of 92.25% and 81.05% for up to 3 subject heart and breathing rates, respectively and provides an error of the order of 10^{-1} Hz.

The proposed system and technique provide high accuracy multi-subject vital signs detection as well lower error compared to known systems. The overall higher average accuracy for all the subjects provides a viable mechanism for detection of multi-subject breathing and heart rates.

In summary, specific embodiments of the systems disclosed herein can provide detection of multi-subject vital signs monitoring, namely breathing and heart rates.

The systems can use the 5G spectrum for detection and monitoring at 3.75GHz with CSI phase difference values.

The proposed MultiBeat algorithm leverages frequency selection methods for multiple subjects, unlike traditional techniques which use subcarrier selection. Subcarrier selection methods obfuscate weak heart signals by selecting subcarriers which may contain single relatively higher strength heart signal and therefore are unsuitable for detecting multi-subject heart rates.

It has been shown that an implementation of the system 600 of the present disclosure can achieve a high accuracy of 92.25% for 3 subjects and 94% for 2 subjects. Furthermore, the breathing rates have the maximum error of 10^{-1} Hz, with an accuracy of 81% for 3 subjects breathing rates and 83.95% for 2 subjects.

Figure 12 shows graphs of nine subcarriers (labelled 1200, 1202, 1204, 1206, 1208, 1210, 1212, 1214, 1216). For each subcarrier, the graph is shown as power spectrum versus frequency. Each subcarrier has three frequencies selected (in the present example, n is equal to three), with each frequency being associated with a peak and indicated by a vertical line.

Figure 13 is a graph of the power spectrum versus frequency for a plurality of subcarriers as provided to the aggregate unit 630 presented on a single graph. Labelling has been omitted for clarity.

As discussed previously, the aggregate unit 630 is configured to aggregate power density peaks per frequency, which may result in k frequencies (labelled 1300, 1302, 1304, 1306, 1308), where k>=n, as the values of n frequencies may be different for each subcarrier resulting in k frequencies overall during aggregation. In summary, k unique frequencies are obtained from the n frequencies per subcarrier.

Figure 14 is a graph of power spectrum versus frequency showing the sum of the peak values per frequency for the k frequencies 1300, 1302, 1304, 1306, 1308 as provided as an output by the block 630a of the aggregate unit 630 as illustrated in Figure 6(c). This procedure provides a power spectrum versus frequency trace 1400.

Figure 15 is a graph of power spectrum versus frequency showing the processed trace 1400 (denoted by numeral 1500) as processed by the block 630b in multiplying the aggregated peaks by the weight of the reciprocal of the number of frequencies. The trace 1500 is provided as an output from the aggregate unit 630 to the peak determination unit 632.

Figure 16 is a graph of the trace 1500 showing estimated heart rate frequencies as may be determined using the peak determination unit 632 as labelled by 1600, 1602, 1604. Actual heart rates as measured by a ground truth sensor are labelled 1606, 1608, 1610 for comparison. The heart rate frequencies 1600, 1602, 1604 may be provided as outputs of the peak determination unit 632.

It will be appreciated that the graphs shown in Figures 12 to 16 are provided to show examples of the operation of the frequency selection unit 628, the aggregate unit 630 and the peak determination unit 632. It will be appreciated that these graphs are provided for illustrative purposes and to aid the clarity of the previous description.

It will be appreciated that common refence numerals and variables between Figures have been used to represent common features.

Various improvements and modifications may be made to the above without departing from the scope of the disclosure.

CLAIMS

1. A vital signs monitoring system configured to estimate the heart rates of two or more subjects using a plurality of subcarriers of a radio frequency (RF) signal.

2. The vital signs monitoring system of claim 1 comprising:

a receiver configured to receive the RF signal after its interaction with the two or more subjects; and

an estimation system comprising a vital signs determination unit configured to estimate the heart rates of the two or more subjects using the plurality of subcarriers of the RF signal received by the receiver.

3. The vital signs monitoring system of claim 2, wherein:

the receiver comprises a first antenna and a second antenna; and

each of the first and second antennas are configured to receive the RF signal.

4. The vital signs monitoring system of claim 3, wherein:

the estimation system comprises:

i) a first phase signal circuit configured to determine a first phase signal for each of the subcarriers of the RF signal received at the first antenna; and

ii) a second phase signal circuit configured to determine a second phase signal for each of the subcarriers of the RF signal received at the second antenna; wherein:

the vital signs determination unit is configured to estimate the heart rates using the first and second phase signals.

5. The vital signs monitoring system of claim 4, wherein the estimation system comprises:

a subtraction unit configured to determine a phase difference signal for each subcarrier by subtracting the first phase signal from the second phase signal for each subcarrier; wherein:

the vital signs determination unit is configured to estimate the heart rates using the first and second phase signals by using the phase difference signal for each subcarrier to estimate the heart rates.

6. The vital signs monitoring system of claim 5, wherein the estimation system comprises a first pre-processing circuit configured to pre-process the RF signal prior to estimation of the heart rates by filtering out portions of the phase difference signals of each of the subcarriers that are unrelated to heart rates.

7. The vital signs monitoring system of claim 6, wherein the first pre-processing circuit is configured to pre-process the RF signal prior to estimation of the heart rates by filtering out portions of the phase difference signals of each of the subcarriers comprising frequencies unrelated to heart rates.

8. The vital signs monitoring system of claim 7, wherein the first pre-processing circuit is configured to pre-process the RF signal prior to estimation of the heart rates by filtering out portions of the phase difference signals of each of the subcarriers comprising frequencies related to respiration rates.

9. The vital signs monitoring system of claim 7 or 8, wherein the first pre-processing circuit comprises a band stop filter configured to filter out frequencies unrelated to heart rates and/or a band pass filter configured to pass frequencies related to heart rates.

10. The vital signs monitoring system of any of claims 7 to 9, wherein the first pre-processing circuit comprises a power spectral density determination unit configured to process the phase difference signals to determine a power spectral density of each of the subcarriers.

11. The vital signs monitoring system of claim 10, wherein the vital sign determination unit is configured to, for each of the subcarriers, identify n peaks in power spectral density, where n is an integer that is equal to the number of subjects.

12. The vital signs monitoring system of claim 11, wherein the vital signs determination unit comprises a frequency selection unit that is configured to, for each of the subcarriers, select the n frequencies having the highest power, thereby identifying the n peaks in power spectral density of each subcarrier.

13. The vital signs monitoring system of claim 12, wherein the vital signs determination unit comprises an aggregate unit configured to aggregate the power spectral density peaks from the n frequencies selected by the frequency selection unit of each of the subcarriers.

14. The vital signs monitoring system of any of claims 4 to 13, wherein each subcarrier is accumulated at the receiver over a time period.

15. The vital signs monitoring system of any preceding claim configured to estimate respiration rates of the two or more subjects using the plurality of subcarriers of the RF signal.

16. The vital signs monitoring system of claim 2, wherein the vital signs determination unit is configured to estimate the respiration rate of the two or more subjects using the plurality of subcarriers of the RF signal received by the receiver.

17. The vital signs monitoring system of claim 16, wherein:

the receiver comprises a first antenna and a second antenna;

each of the first and second antennas are configured to receive the RF signal; and

the estimation system comprises:

i) a first phase signal circuit configured to determine a first phase signal for each of the subcarriers of the RF signal received at the first antenna; and

ii) a second phase signal circuit configured to determine a second phase signal for each of the subcarriers of the RF signal received at the second antenna; wherein:

the vital signs determination unit is configured to estimate the respiration rates using the first and second phase signals to estimate the respiration rates.

18. The vital signs monitoring system of claim 17, wherein the estimation system comprises:

a subtraction unit configured to determine a phase difference signal for each subcarrier by subtracting the first phase signal from the second phase signal for each subcarrier; wherein

the vital signs determination unit is configured to estimate the respiration rates using the first and second phase signals using the phase difference signal for each subcarrier to estimate the respiration rates.

19. The vital signs monitoring system of claim 18, wherein the estimation system comprises a second pre-processing circuit configured to pre-process the RF signal prior to estimation of the respiration rates by filtering out portions of the phase difference signals of each of the subcarriers that are unrelated to the respiration rates.

20. The vital signs monitoring system of any of claims 19, wherein the second pre-processing circuit comprises a second power spectral density determination unit configured to process the phase difference signals to determine a power spectral density of each of the subcarriers.

21. The vital signs monitoring system of claim 20, wherein the vital signs determination unit configured to, for each of the subcarriers, identify n peaks in power spectral density, where n is an integer that is equal to the number of subjects.

22. The vital signs monitoring system of claim 21, wherein the vital signs determination unit comprises a frequency selection unit that is configured to, for each of the subcarriers, select the n frequencies having the highest power, thereby identifying the n peaks in power spectral density of each subcarrier.

23. The vital signs monitoring system of claim 22, wherein the vital signs determination unit comprises an aggregate unit configured to aggregate the power spectral density peaks from the n frequencies selected by the frequency selection unit of each of the subcarriers.

24. The vital signs monitoring system of any preceding claim comprising a transmitter configured to transmit the RF signal.

25. A method of estimating the heart rate of two or more people using the vital signs monitoring system of any preceding claim comprising:

receiving an RF signal; and

estimating the two or more heart rates using the plurality of subcarriers of the RF signal.

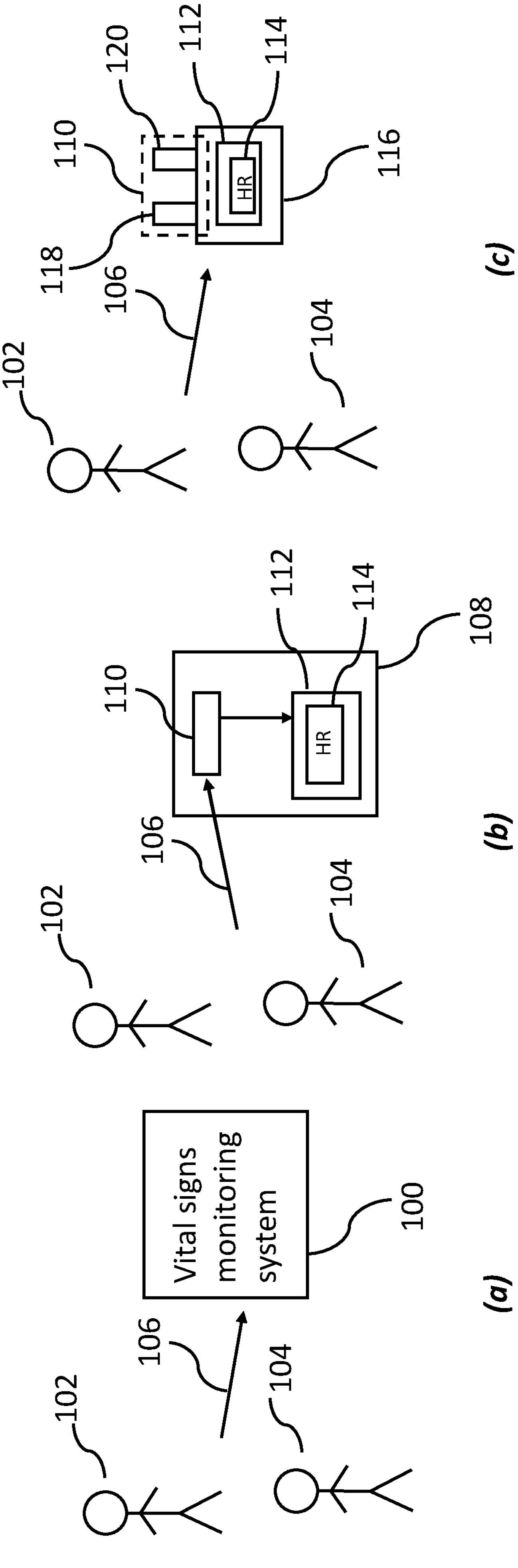

Vital signs monitoring system
100
102
104
106
(a)
110
112
114
HR
108
(b)
120
118
116
(c)

Figure 1

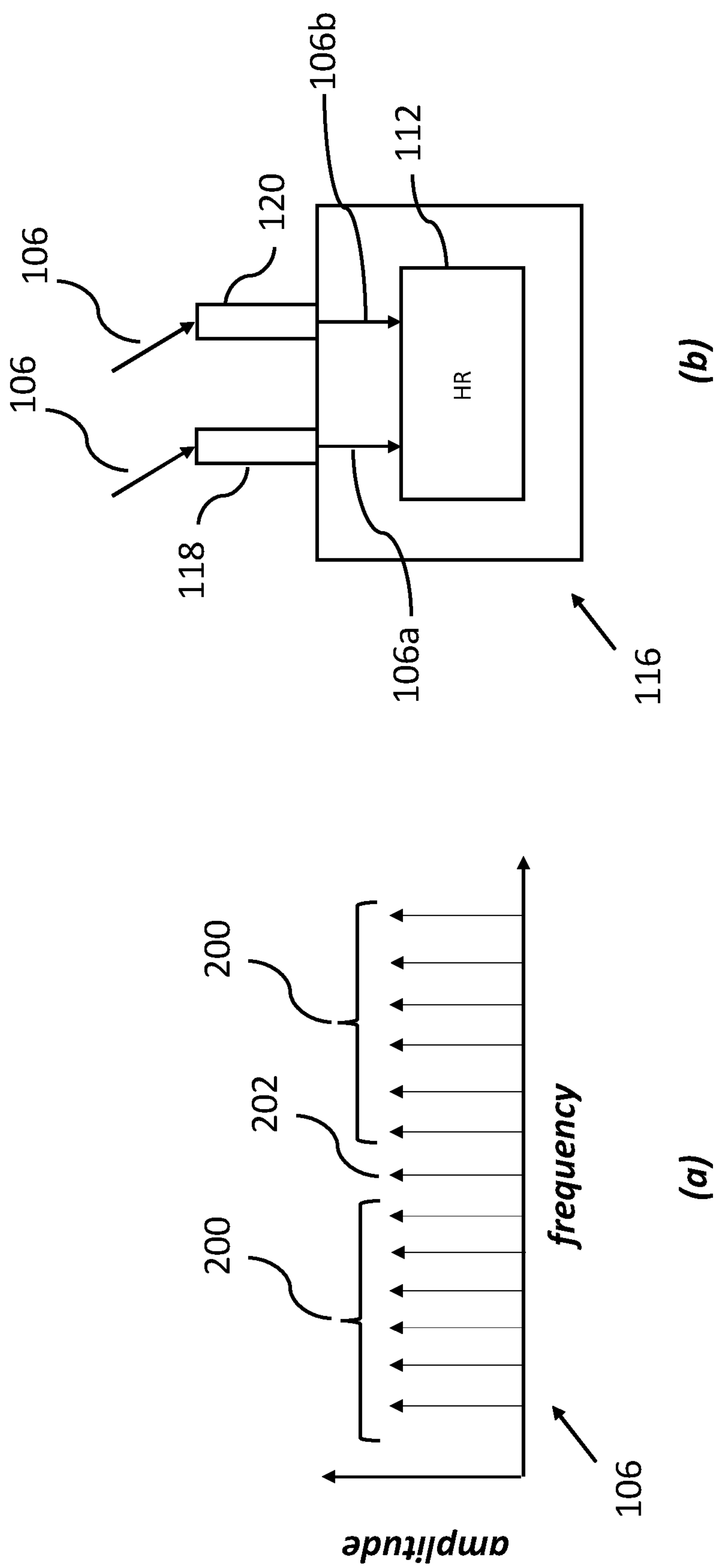

106
106
120
106b
112
HR
118
106a
116
(b)
200
202
200
frequency
amplitude
106
(a)

Figure 2

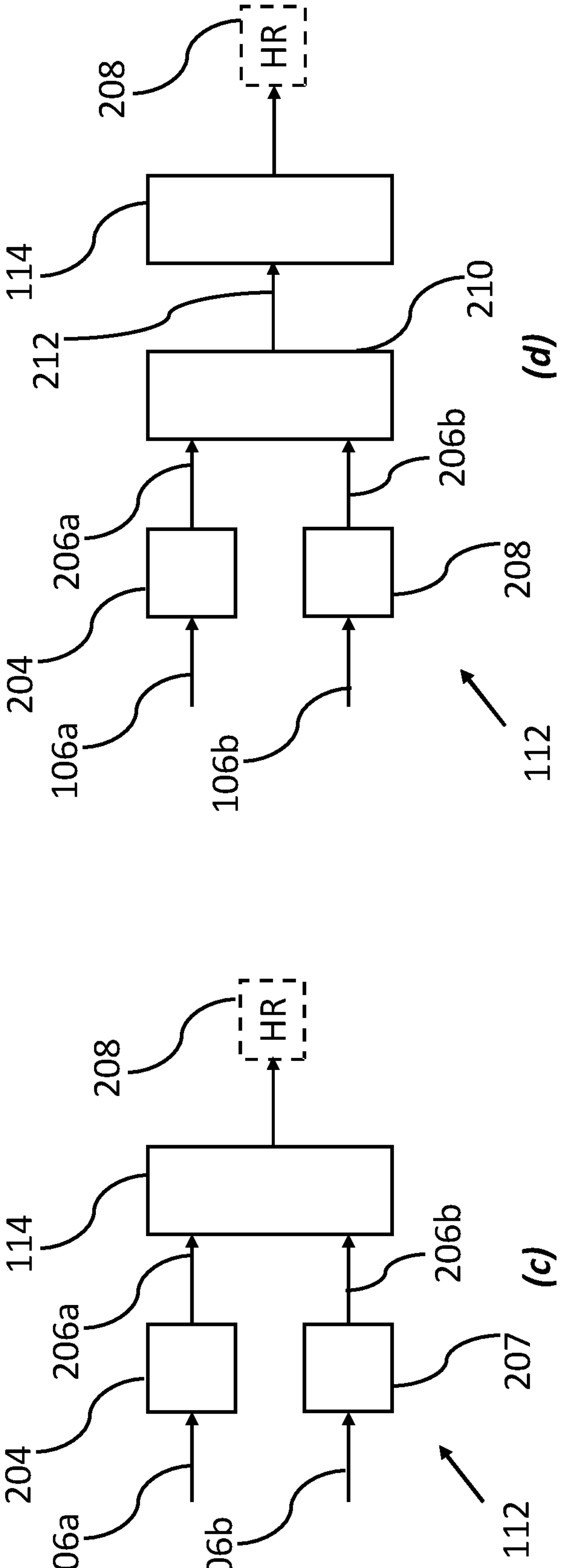

208
HR
114
212
210
206a
206b
204
208
106a
106b
112
(d)
208
HR
114
206a
206b
204
207
106a
106b
112
(c)

Figure 2

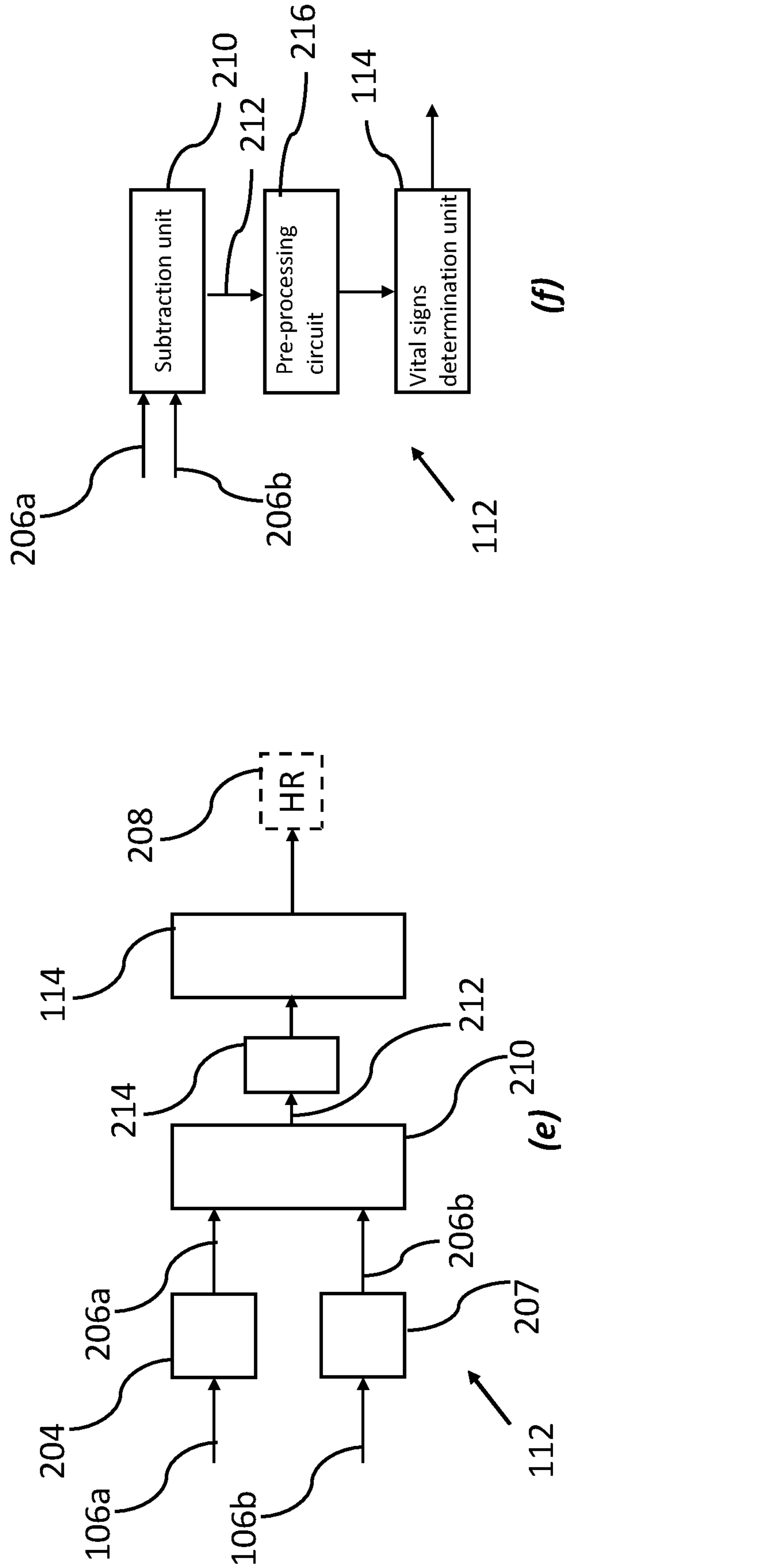
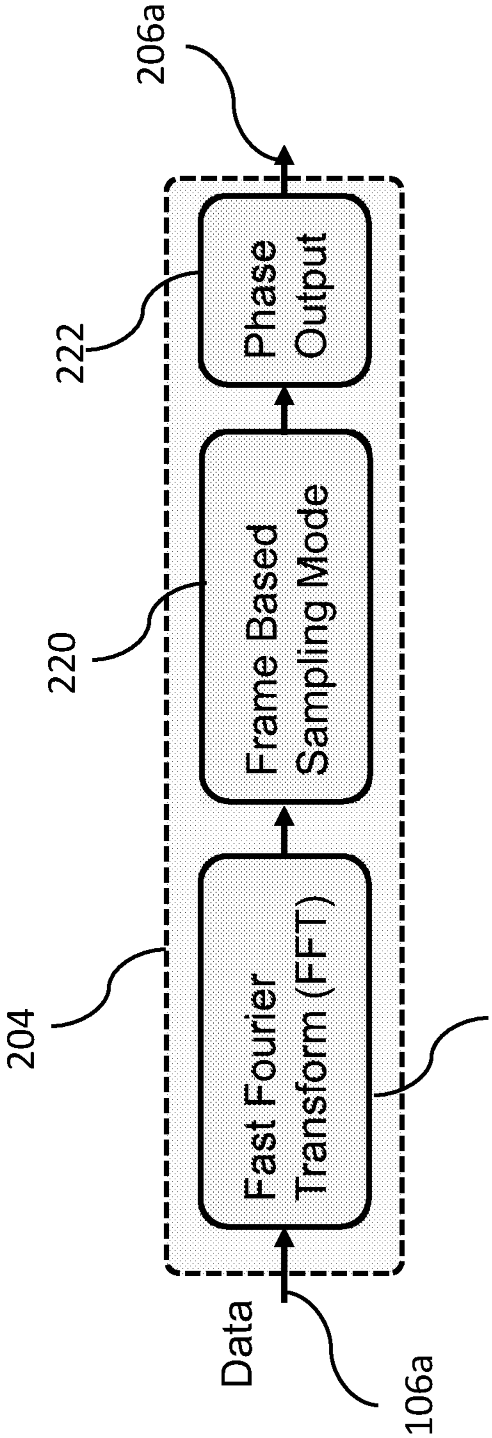
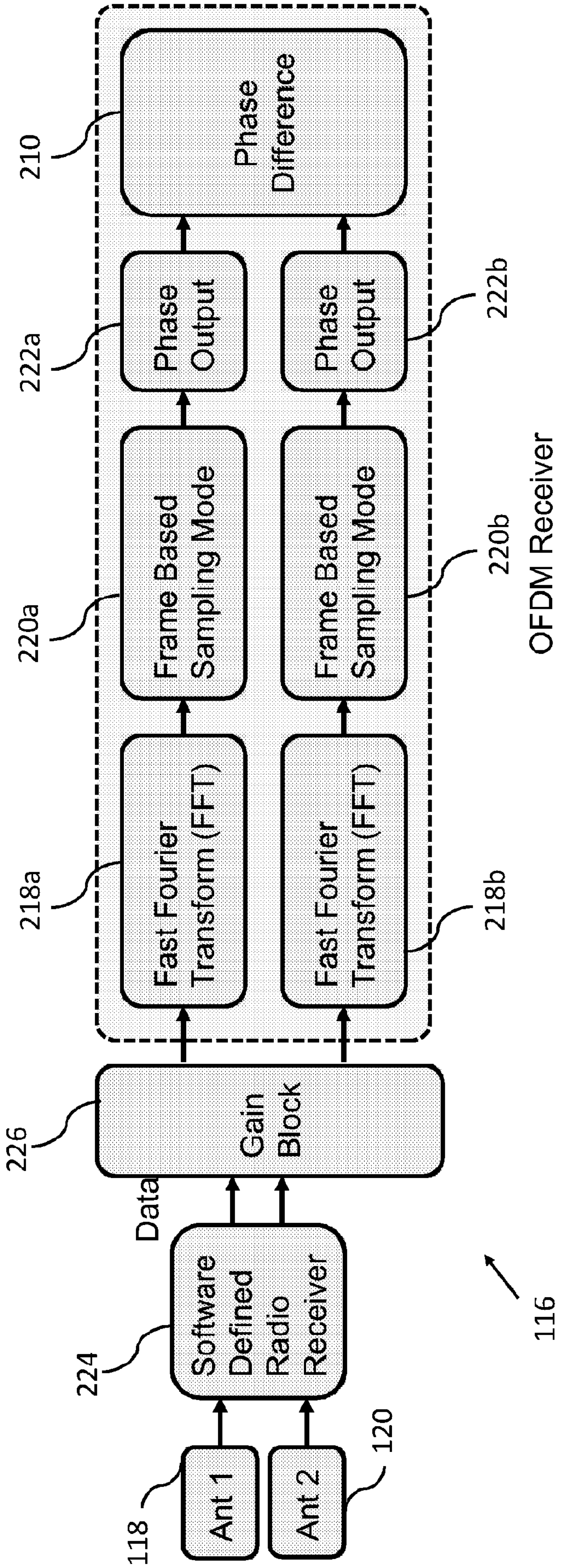

210
216
114
212
Subtraction unit
Pre-processing circuit
Vital signs determination unit
206a
206b
112
(f)
208
HR
114
212
214
210
(e)
206a
206b
207
204
106a
106b
112

Figure 2


Data
106a
Fast Fourier Transform (FFT)
218
204
Frame Based Sampling Mode
220
Phase Output
222
206a

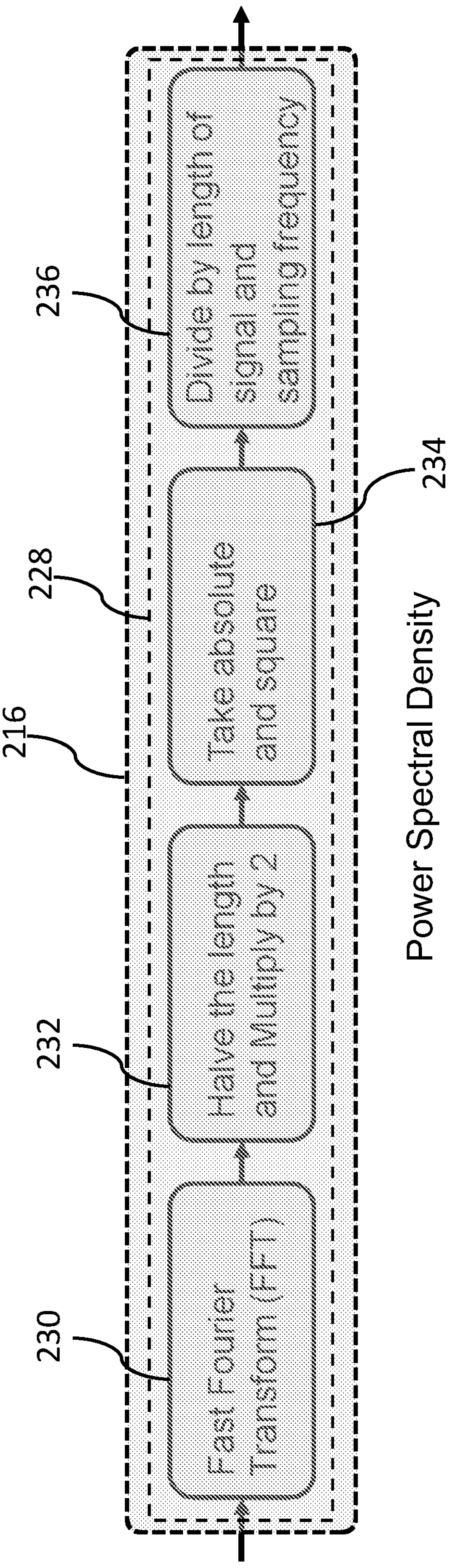
(g)

Figure 2


Ant 1
Ant 2
118
120
Software Defined Radio Receiver
224
Data
Gain Block
226
Fast Fourier Transform (FFT)
218a
Fast Fourier Transform (FFT)
218b
Frame Based Sampling Mode
220a
Frame Based Sampling Mode
220b
Phase Output
222a
Phase Output
222b
Phase Difference
210
OFDM Receiver
116

(h)

Figure 2


236
Divide by length of signal and sampling frequency
234
228
Take absolute and square
216
232
Halve the length and Multiply by 2
230
Fast Fourier Transform (FFT)
Power Spectral Density

(i)

Figure 2

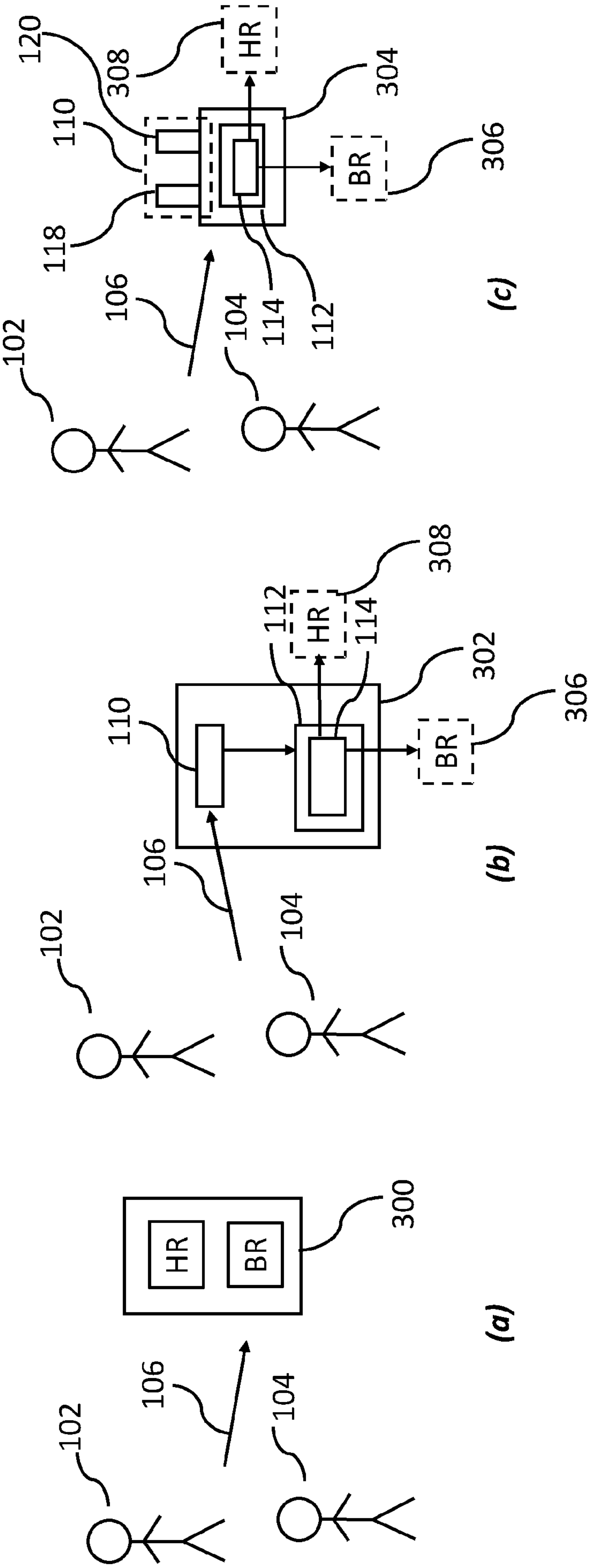

(a)
102
104
106
300
HR
BR
(b)
102
104
106
110
112
114
302
306
308
HR
BR
(c)
102
104
106
110
112
114
118
120
304
306
308
HR
BR

Figure 3

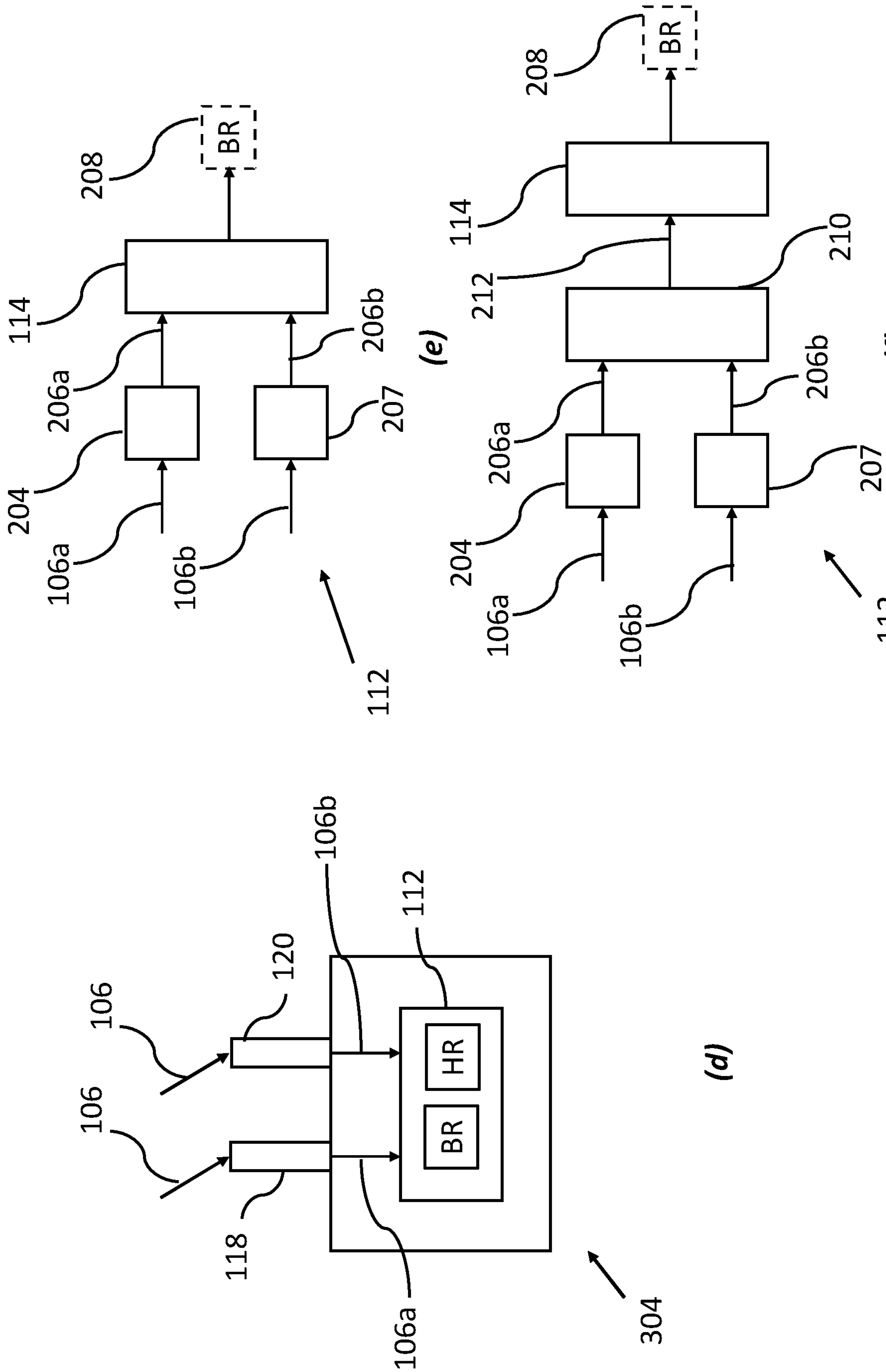
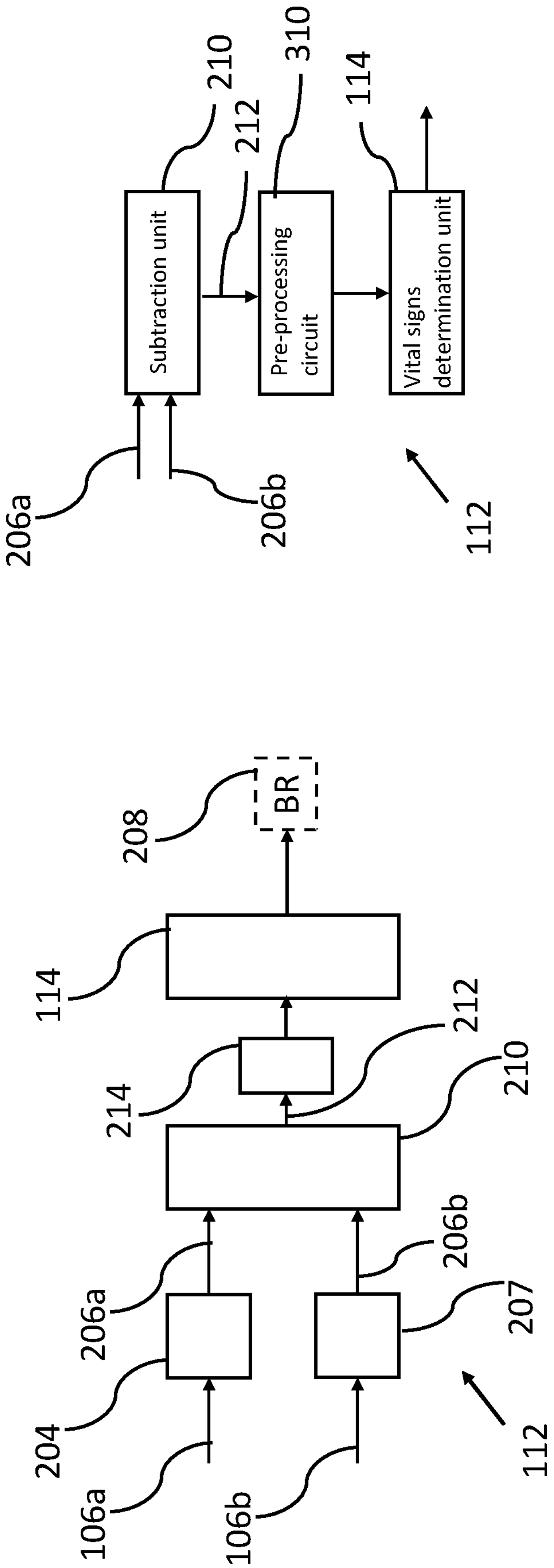

208
BR
114
206a
206b
204
207
106a
106b
112
(e)
208
BR
114
212
210
206a
206b
204
207
106a
106b
112
(f)
106
106
120
106b
112
HR
BR
118
106a
304
(d)

Figure 3


206a
206b
Subtraction unit
210
212
Pre-processing circuit
310
Vital signs determination unit
114
112
106a
106b
204
206a
207
206b
210
212
214
114
208
BR
112

(h)

(g)

Figure 3

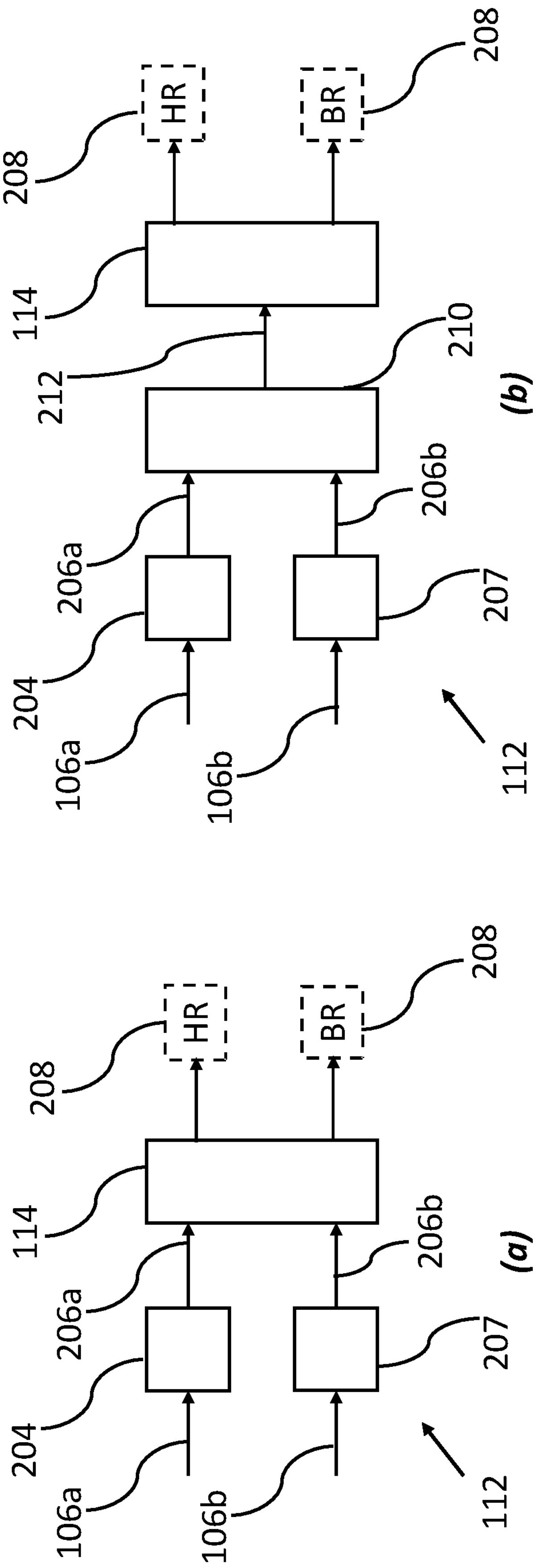
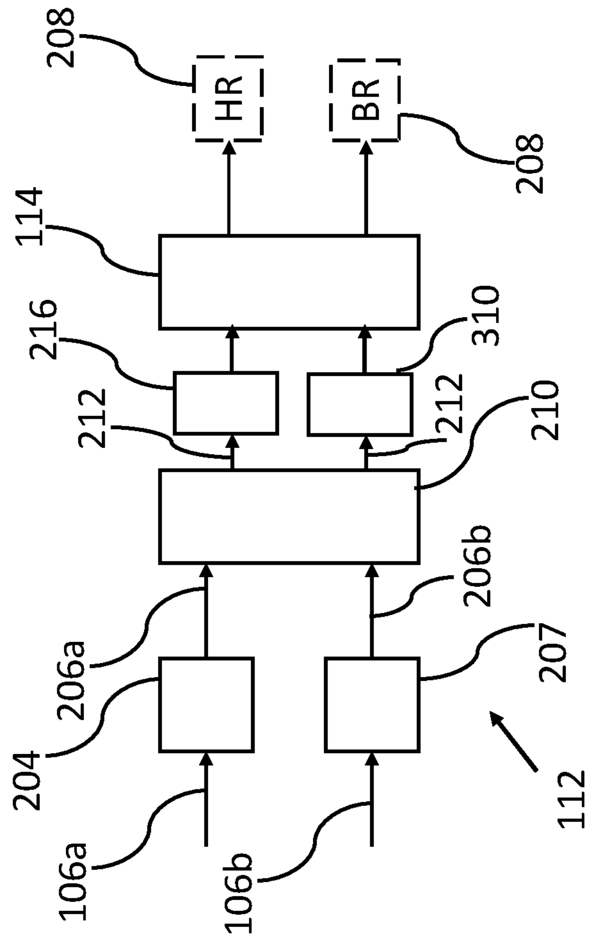

208
HR
BR
208
114
212
210
206a
206b
204
207
106a
106b
112
(b)
208
HR
BR
208
114
206a
206b
204
207
106a
106b
112
(a)

Figure 4


208
HR
BR
208
114
216
310
212
212
210
206a
206b
204
207
106a
106b
112

(c)

Figure 4

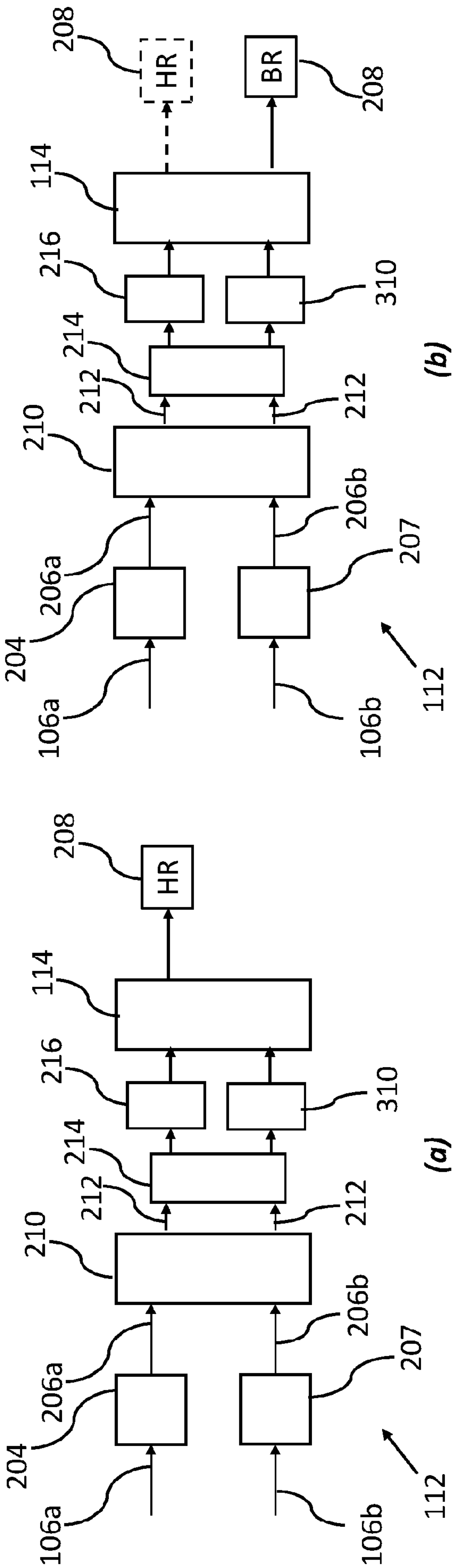
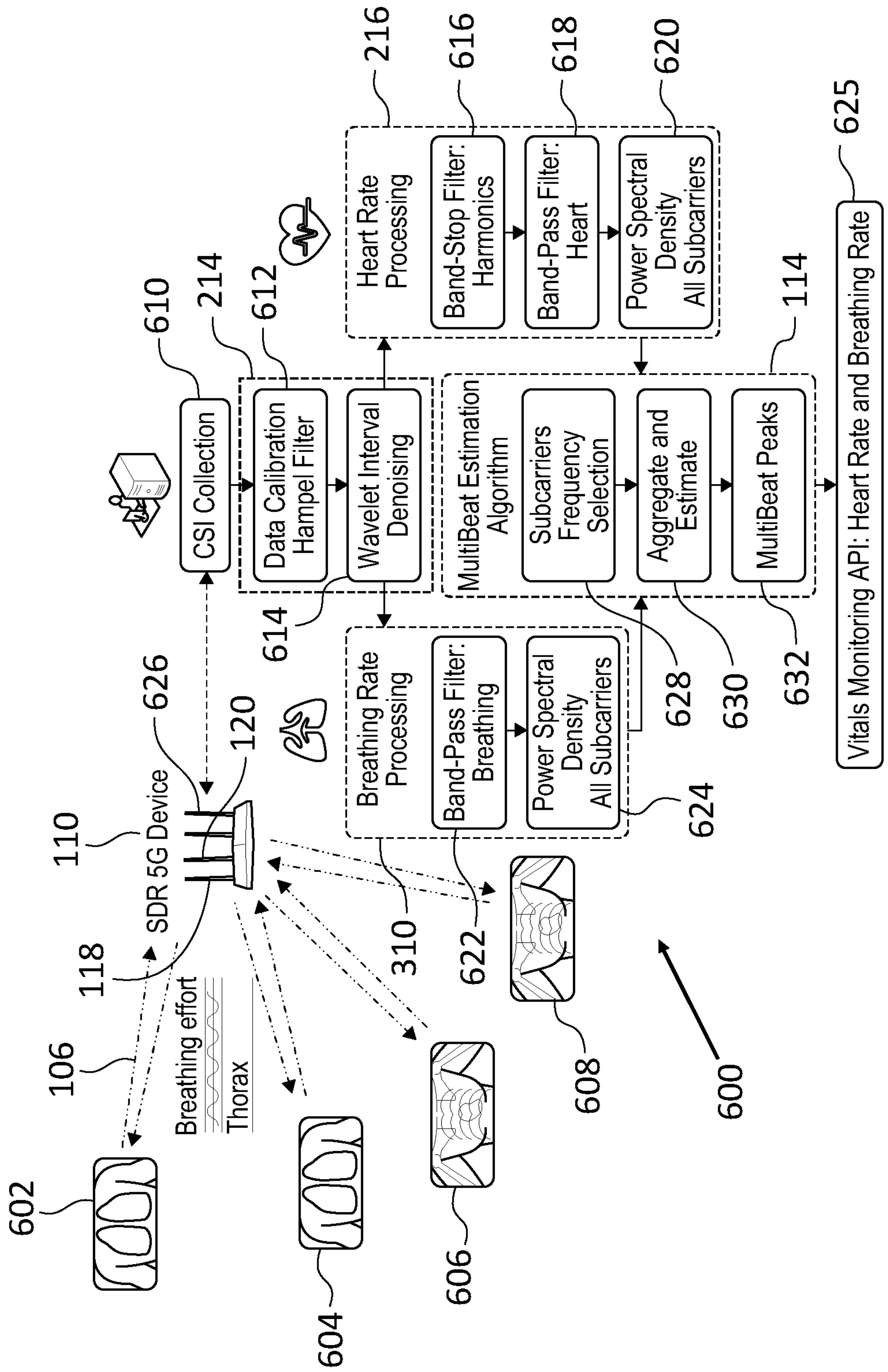
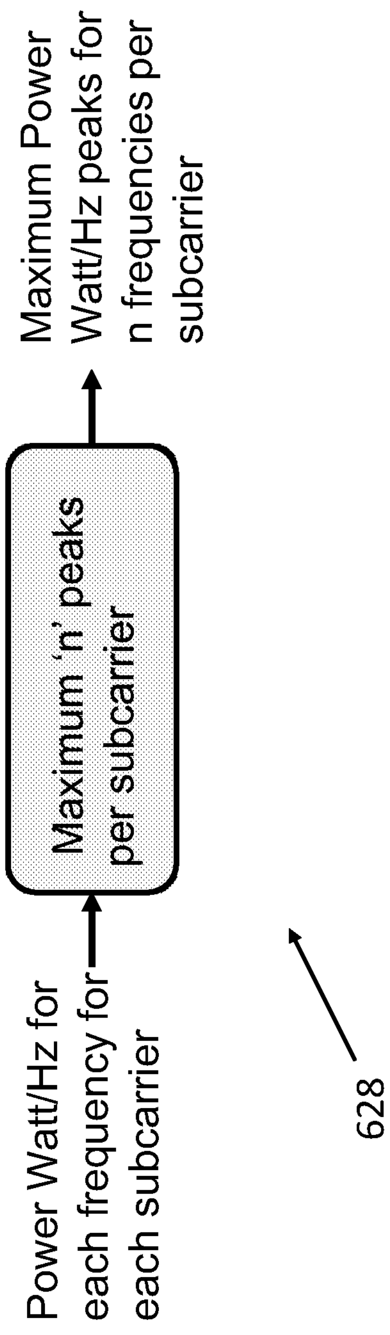

208
HR
BR
208
114
216
310
214
212
212
210
206a
206b
207
204
106a
106b
112
(b)
208
HR
114
216
310
214
212
212
210
206a
206b
207
204
106a
106b
112
(a)

Figure 5


216
616
618
620
Heart Rate Processing
Band-Stop Filter: Harmonics
Band-Pass Filter: Heart
Power Spectral Density All Subcarriers
610
214
612
CSI Collection
Data Calibration Hampel Filter
Wavelet Interval Denoising
614
114
MultiBeat Estimation Algorithm
Subcarriers Frequency Selection
Aggregate and Estimate
MultiBeat Peaks
628
630
632
625
Vitals Monitoring API: Heart Rate and Breathing Rate
626
120
110
SDR 5G Device
118
106
Breathing effort
Thorax
Breathing Rate Processing
Band-Pass Filter: Breathing
Power Spectral Density All Subcarriers
624
310
622
602
604
606
608
600

(a)
Figure 6


Power Watt/Hz for each frequency for each subcarrier
Maximum 'n' peaks per subcarrier
Maximum Power Watt/Hz peaks for n frequencies per subcarrier
628

(b)

Figure 6

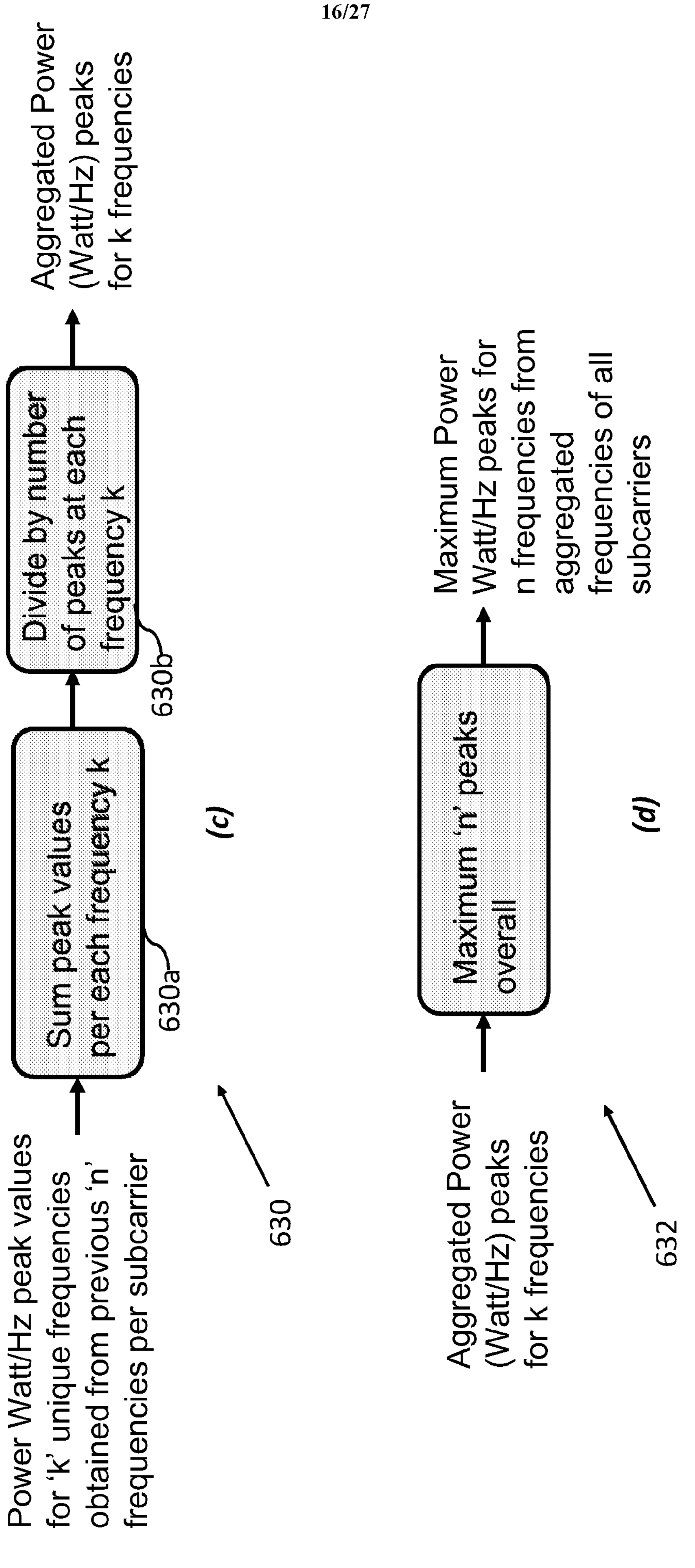

Power Watt/Hz peak values for 'k' unique frequencies obtained from previous 'n' frequencies per subcarrier
Sum peak values per each frequency k
630a
Divide by number of peaks at each frequency k
630b
Aggregated Power (Watt/Hz) peaks for k frequencies
(c)
630
Aggregated Power (Watt/Hz) peaks for k frequencies
Maximum 'n' peaks overall
Maximum Power Watt/Hz peaks for n frequencies from aggregated frequencies of all subcarriers
(d)
632

Figure 6

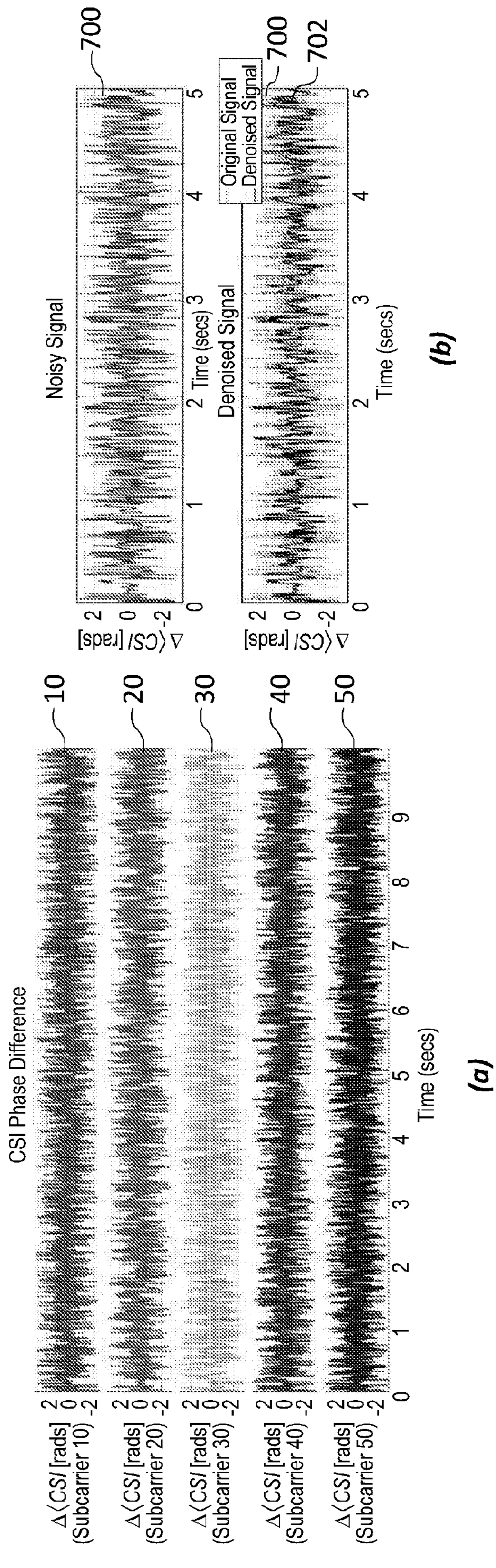

CSI Phase Difference
Δ∠CSI [rads] (Subcarrier 10)
Δ∠CSI [rads] (Subcarrier 20)
Δ∠CSI [rads] (Subcarrier 30)
Δ∠CSI [rads] (Subcarrier 40)
Δ∠CSI [rads] (Subcarrier 50)
10
20
30
40
50
Time (secs)
(a)
Noisy Signal
Δ∠CSI [rads]
Time (secs)
700
Denoised Signal
Original Signal
Denoised Signal
700
702
Time (secs)
(b)

Figure 7

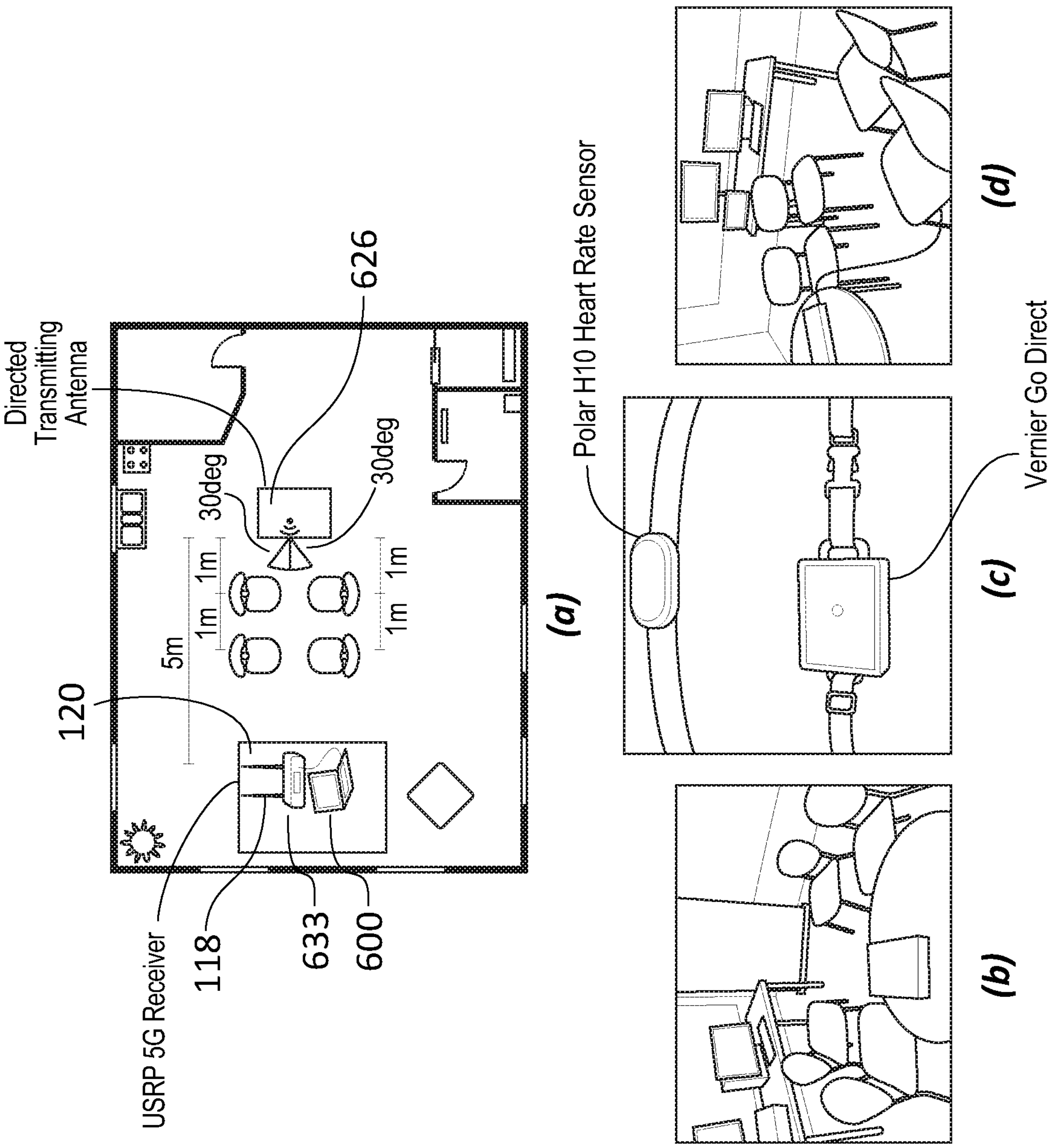

Directed
Transmitting
Antenna
626
30deg
30deg
1m
1m
1m
1m
5m
120
USRP 5G Receiver
118
633
600
(a)
Polar H10 Heart Rate Sensor
Vernier Go Direct
Respiration Sensor
(b)
(c)
(d)

Figure 8

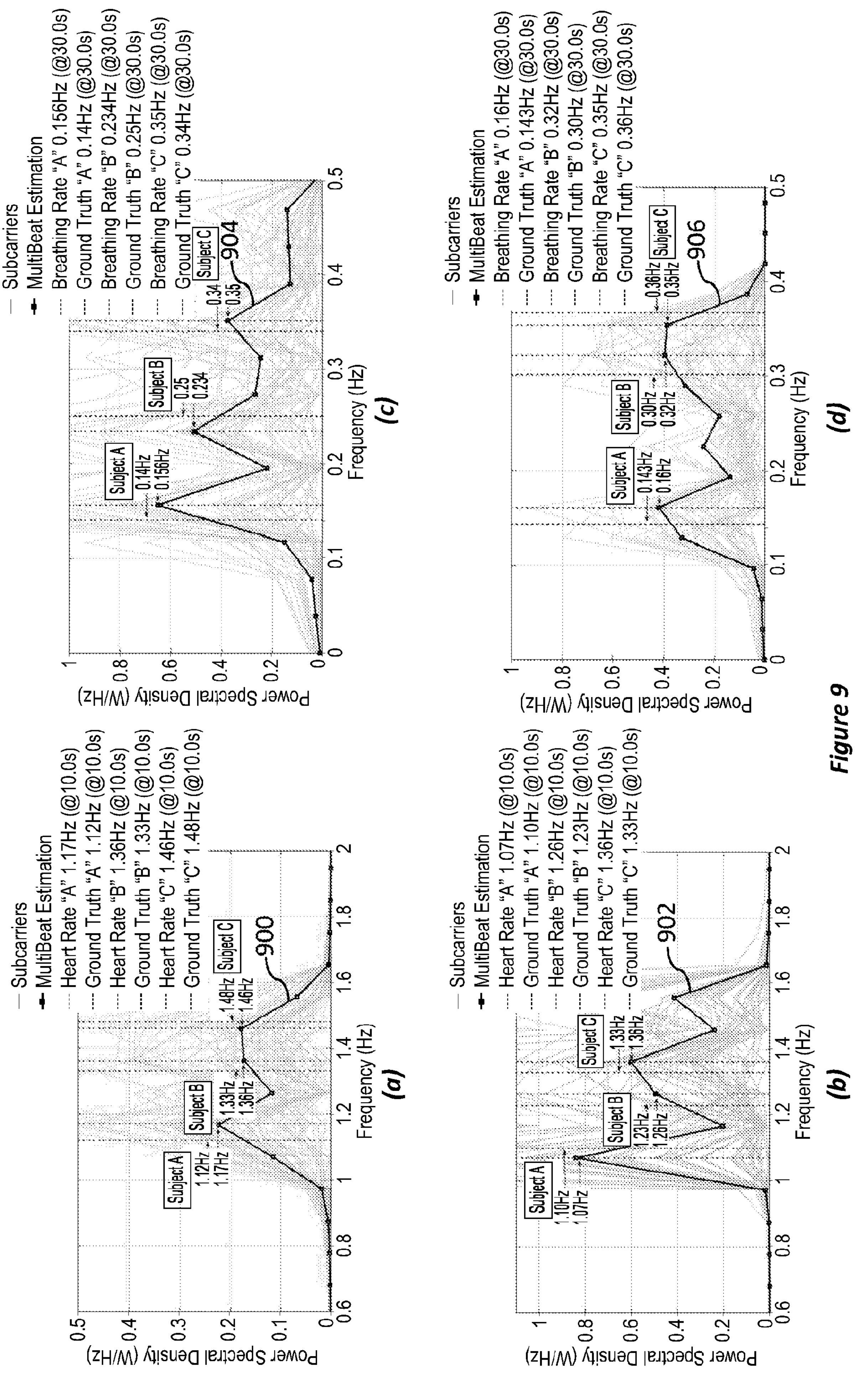

Subcarriers
MultiBeat Estimation
Breathing Rate "A" 0.156Hz (@30.0s)
Ground Truth "A" 0.14Hz (@30.0s)
Breathing Rate "B" 0.234Hz (@30.0s)
Ground Truth "B" 0.25Hz (@30.0s)
Breathing Rate "C" 0.35Hz (@30.0s)
Ground Truth "C" 0.34Hz (@30.0s)
Subject A
0.14Hz
0.156Hz
Subject B
0.25
0.234
Subject C
0.34
0.35
904
Power Spectral Density (W/Hz)
Frequency (Hz)
(c)
Subcarriers
MultiBeat Estimation
Breathing Rate "A" 0.16Hz (@30.0s)
Ground Truth "A" 0.143Hz (@30.0s)
Breathing Rate "B" 0.32Hz (@30.0s)
Ground Truth "B" 0.30Hz (@30.0s)
Breathing Rate "C" 0.35Hz (@30.0s)
Ground Truth "C" 0.36Hz (@30.0s)
Subject A
0.143Hz
0.16Hz
Subject B
0.30Hz
0.32Hz
Subject C
0.36Hz
0.35Hz
906
Power Spectral Density (W/Hz)
Frequency (Hz)
(d)
Subcarriers
MultiBeat Estimation
Heart Rate "A" 1.17Hz (@10.0s)
Ground Truth "A" 1.12Hz (@10.0s)
Heart Rate "B" 1.36Hz (@10.0s)
Ground Truth "B" 1.33Hz (@10.0s)
Heart Rate "C" 1.46Hz (@10.0s)
Ground Truth "C" 1.48Hz (@10.0s)
Subject A
1.12Hz
1.17Hz
Subject B
1.33Hz
1.36Hz
Subject C
1.48Hz
1.46Hz
900
Power Spectral Density (W/Hz)
Frequency (Hz)
(a)
Subcarriers
MultiBeat Estimation
Heart Rate "A" 1.07Hz (@10.0s)
Ground Truth "A" 1.10Hz (@10.0s)
Heart Rate "B" 1.26Hz (@10.0s)
Ground Truth "B" 1.23Hz (@10.0s)
Heart Rate "C" 1.36Hz (@10.0s)
Ground Truth "C" 1.33Hz (@10.0s)
Subject A
1.10Hz
1.07Hz
Subject B
1.23Hz
1.26Hz
Subject C
1.33Hz
1.36Hz
902
Power Spectral Density (W/Hz)
Frequency (Hz)
(b)

Figure 9

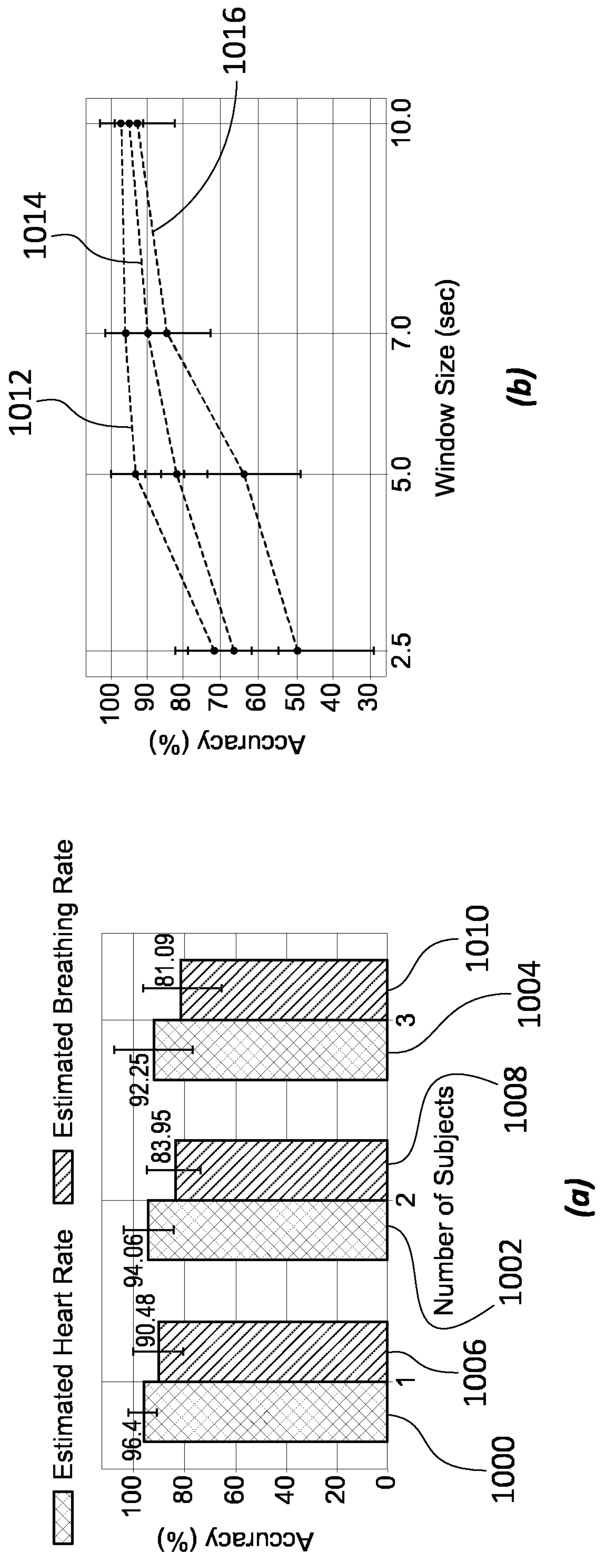

Estimated Heart Rate
Estimated Breathing Rate
Accuracy (%)
100
80
60
40
20
0
96.4
90.48
94.06
83.95
92.25
81.09
1
2
3
Number of Subjects
1000
1006
1002
1008
1004
1010
(a)
Accuracy (%)
100
90
80
70
60
50
40
30
2.5
5.0
7.0
10.0
Window Size (sec)
1012
1014
1016
(b)

Figure 10

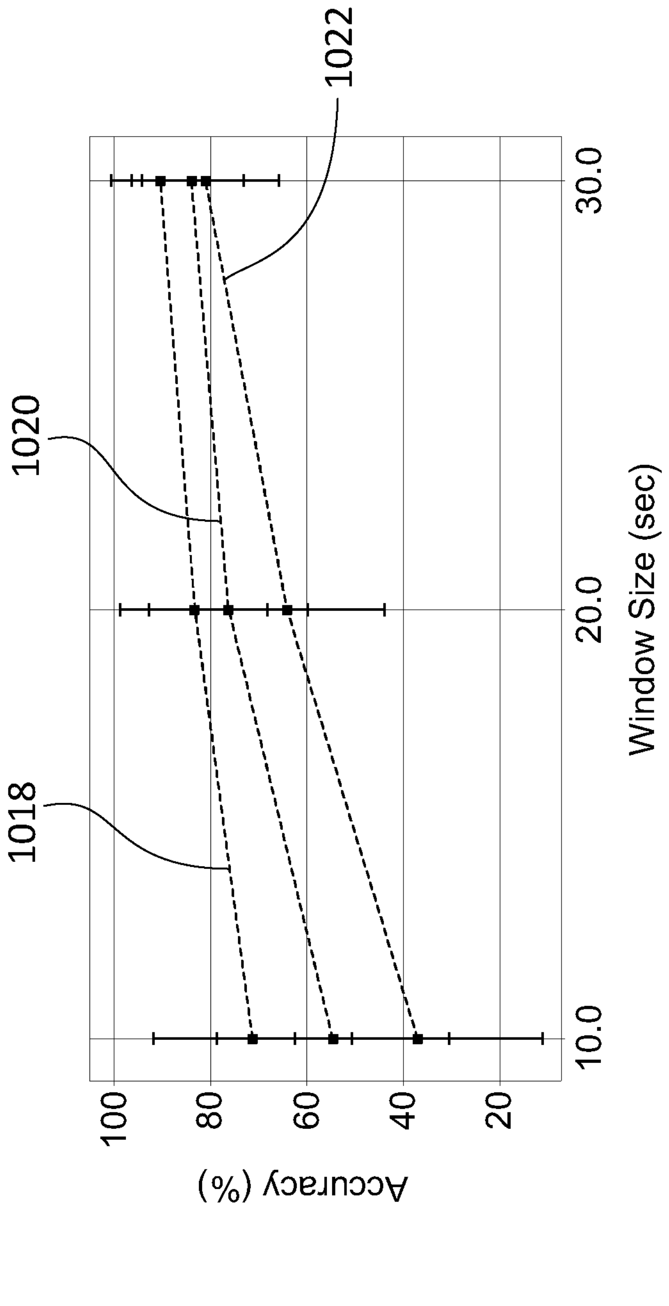

1018
1020
1022
Accuracy (%)
100
80
60
40
20
10.0
20.0
30.0
Window Size (sec)

Figure 10(c)

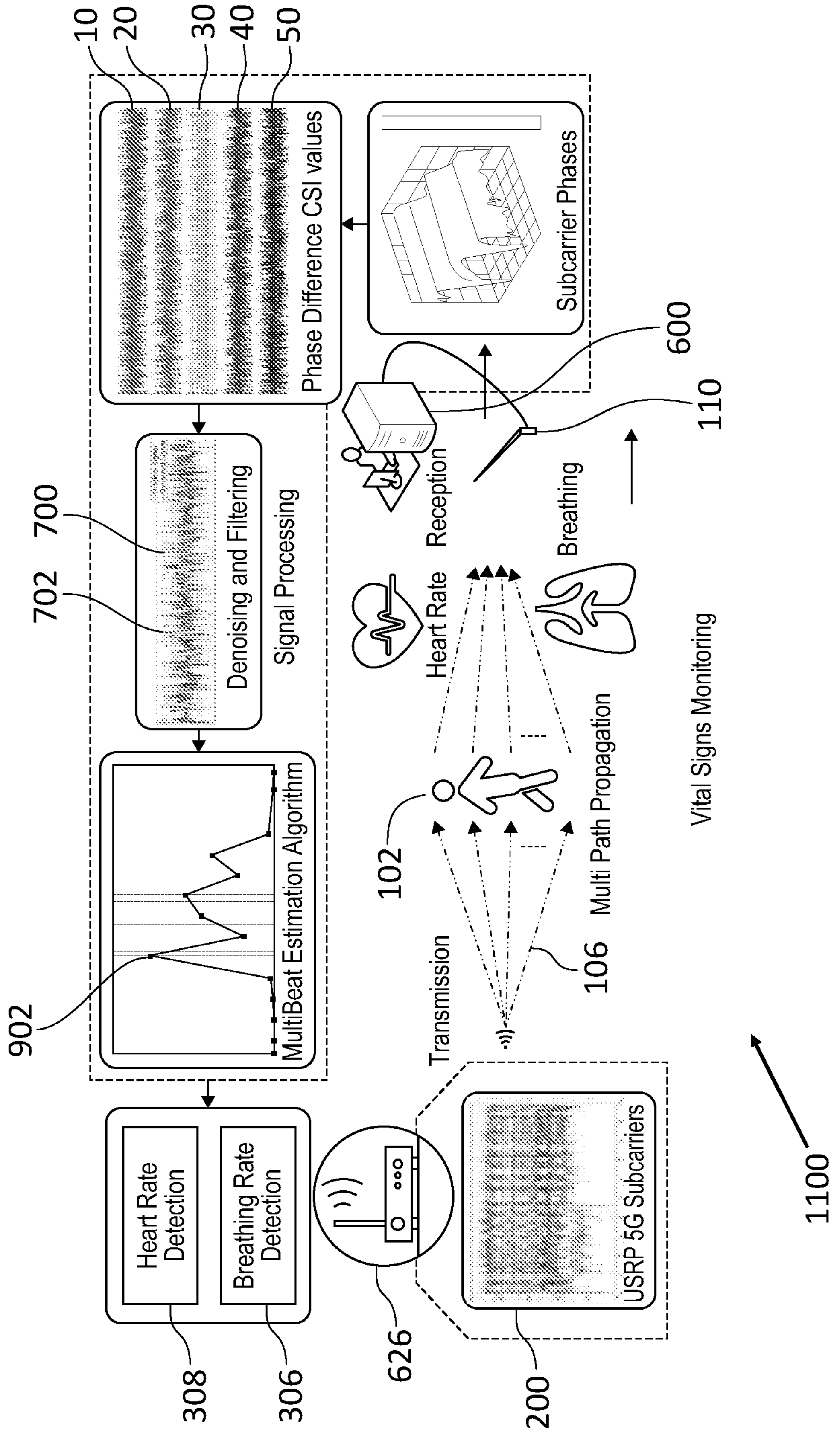

10
20
30
40
50
Phase Difference CSI values
Subcarrier Phases
600
110
700
702
Denoising and Filtering
Signal Processing
Reception
Breathing
Heart Rate
902
MultiBeat Estimation Algorithm
102
Multi Path Propagation
Vital Signs Monitoring
Transmission
106
Heart Rate Detection
Breathing Rate Detection
USRP 5G Subcarriers
308
306
626
200
1100

Figure 11

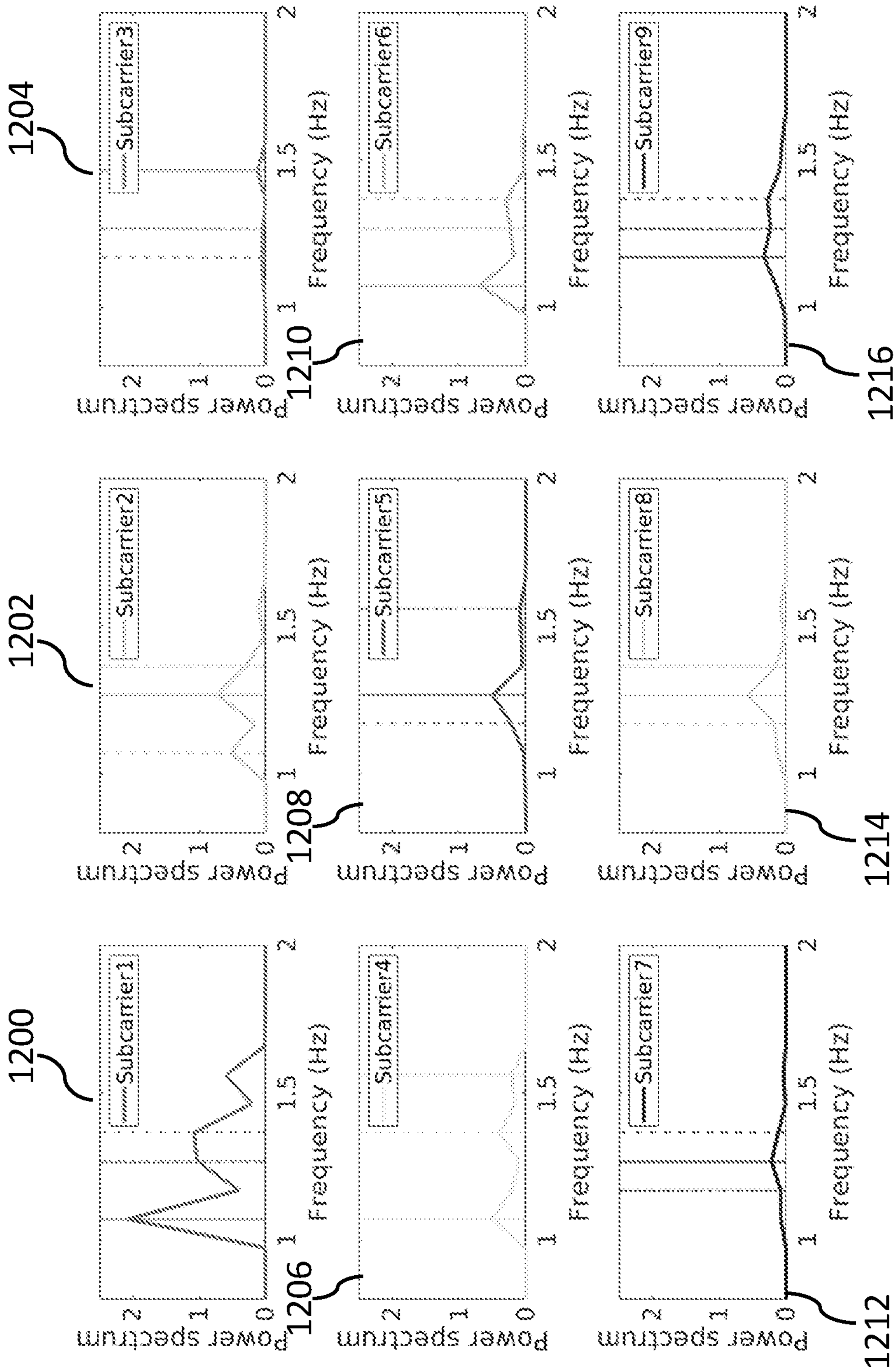

1200
1202
1204
1206
1208
1210
1212
1214
1216
Subcarrier1
Subcarrier2
Subcarrier3
Subcarrier4
Subcarrier5
Subcarrier6
Subcarrier7
Subcarrier8
Subcarrier9
Power spectrum
Frequency (Hz)

Figure 12

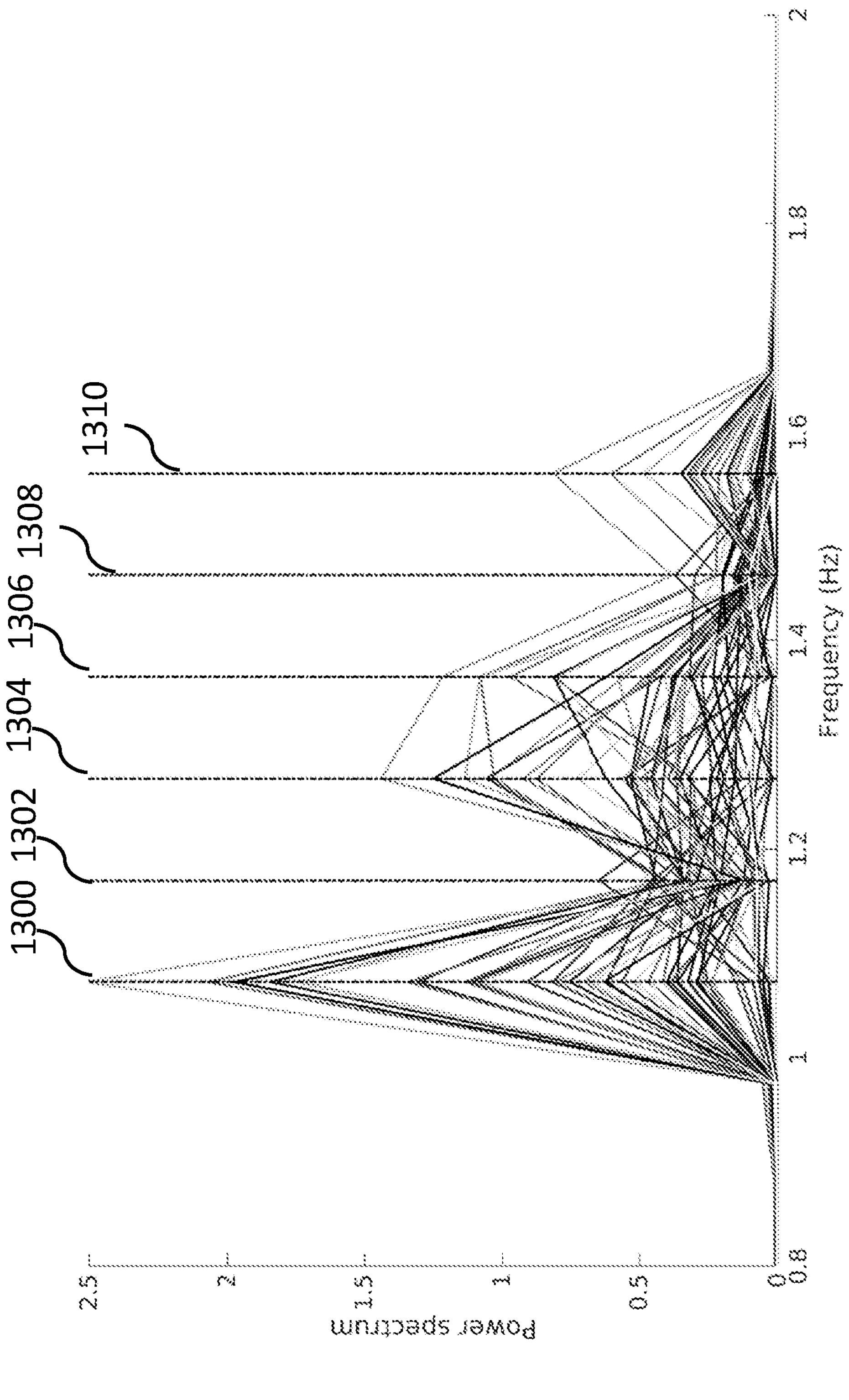

1300
1302
1304
1306
1308
1310
Power spectrum
0
0.5
1
1.5
2
2.5
Frequency (Hz)
0.8
1
1.2
1.4
1.6
1.8
2

Figure 13

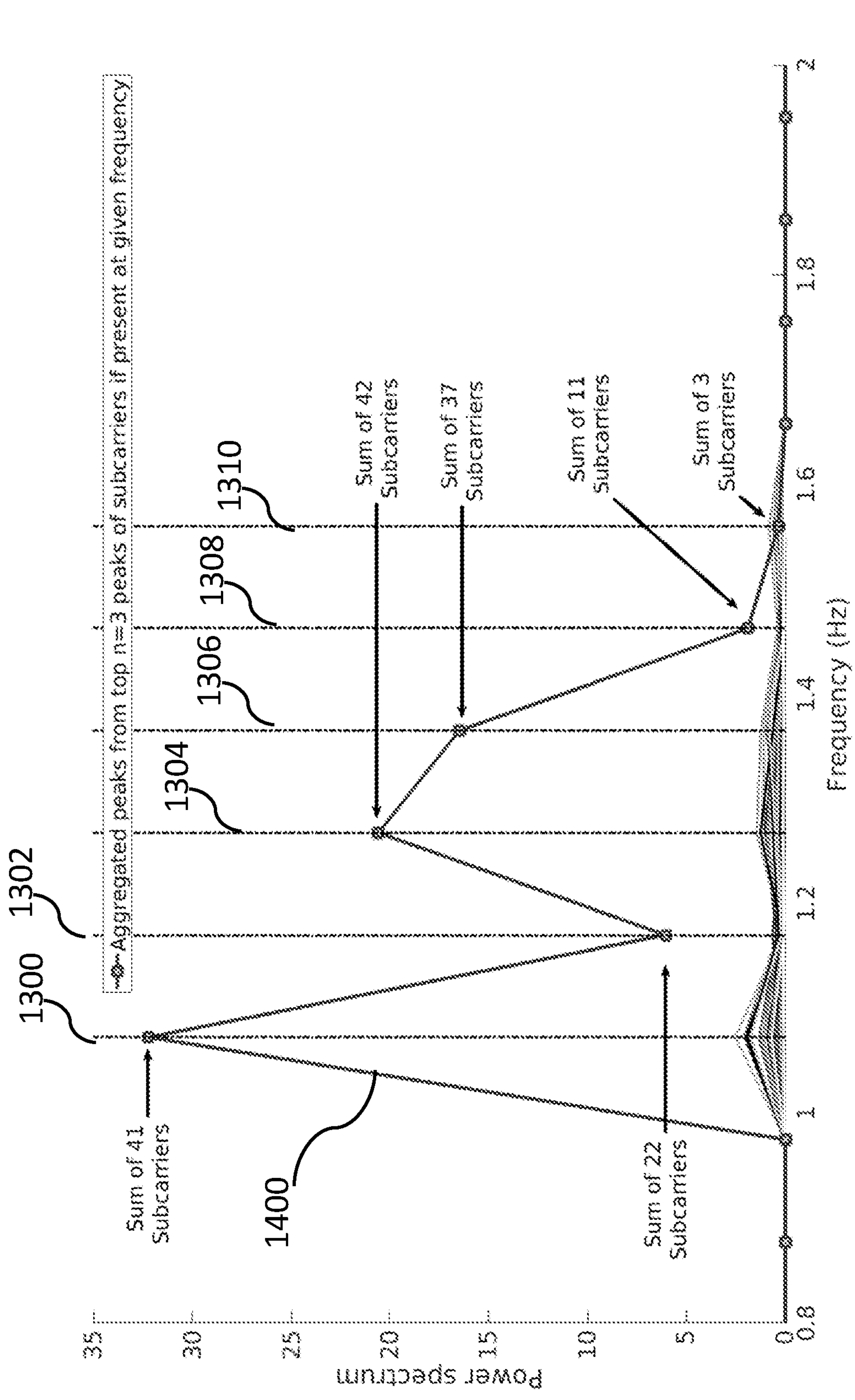

Aggregated peaks from top n=3 peaks of subcarriers if present at given frequency
1300
1302
1304
1306
1308
1310
1400
Sum of 41 Subcarriers
Sum of 22 Subcarriers
Sum of 42 Subcarriers
Sum of 37 Subcarriers
Sum of 11 Subcarriers
Sum of 3 Subcarriers
Power spectrum
0
5
10
15
20
25
30
35
Frequency (Hz)
0.8
1
1.2
1.4
1.6
1.8
2

Figure 14

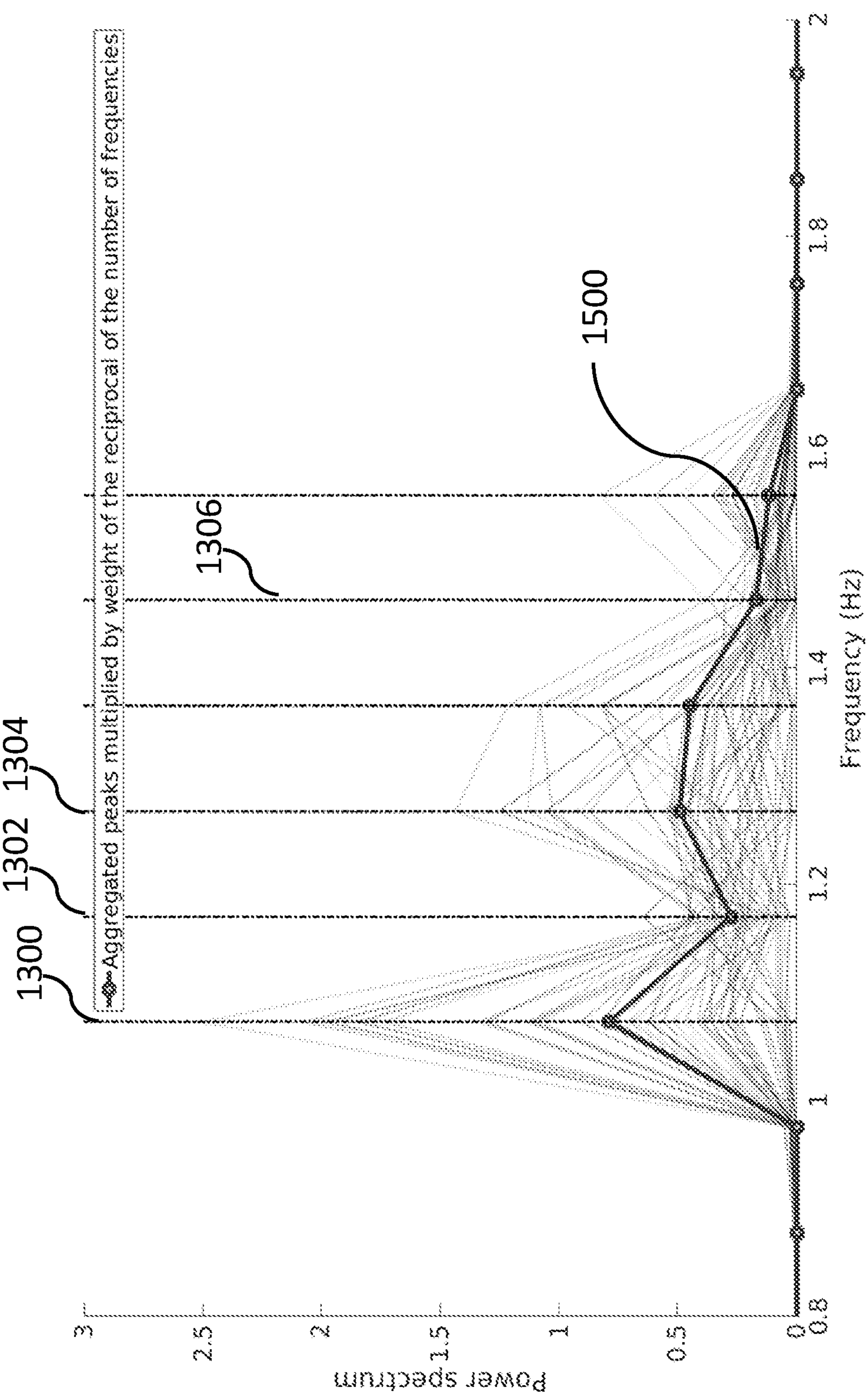

Aggregated peaks multiplied by weight of the reciprocal of the number of frequencies
Power spectrum
Frequency (Hz)
0
0.5
1
1.5
2
2.5
3
0.8
1
1.2
1.4
1.6
1.8
2
1300
1302
1304
1306
1500

Figure 15

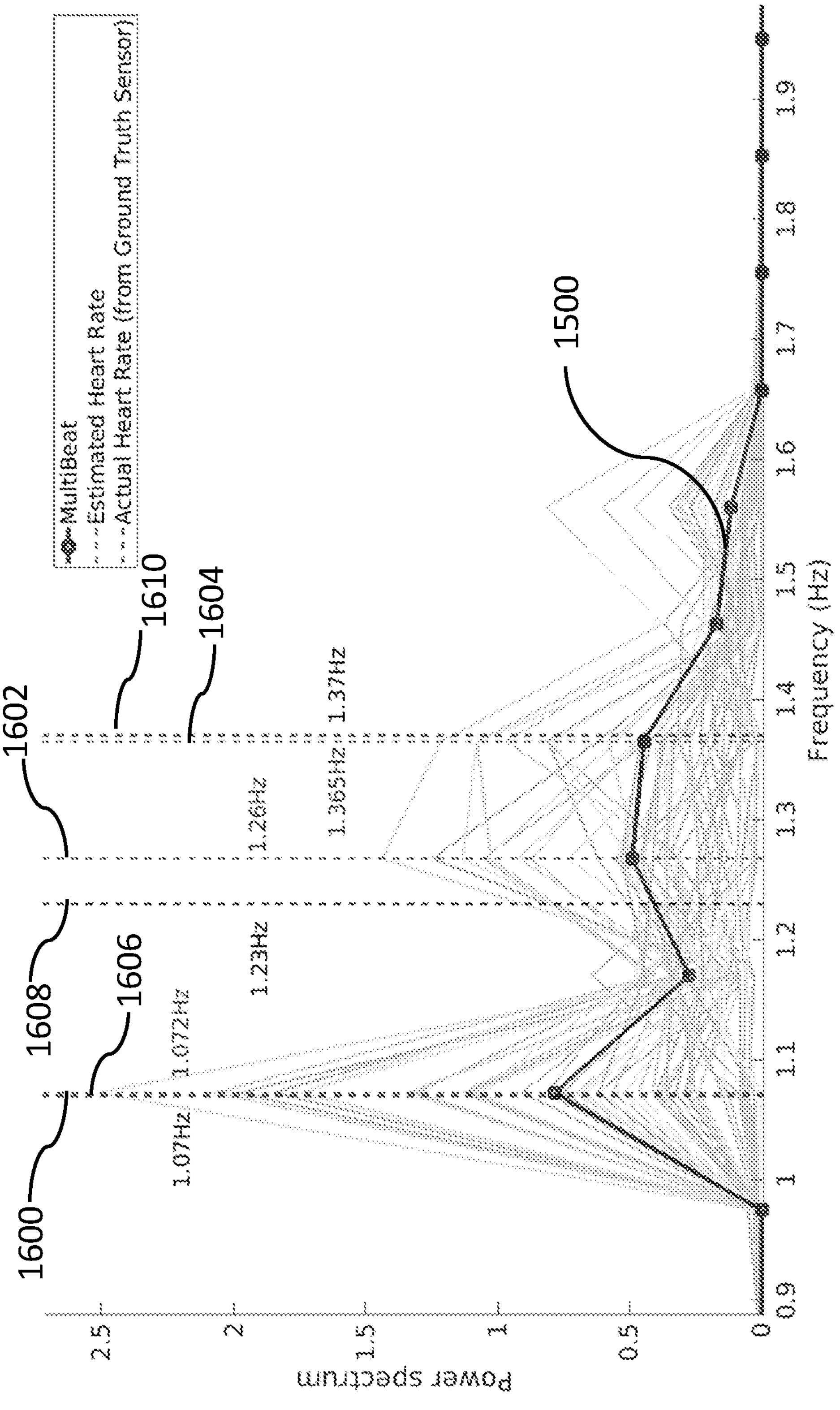

MultiBeat
Estimated Heart Rate
Actual Heart Rate (from Ground Truth Sensor)
1600
1608
1606
1602
1610
1604
1500
1.07Hz
1.072Hz
1.23Hz
1.26Hz
1.365Hz
1.37Hz
Power spectrum
0
0.5
1
1.5
2
2.5
0.9
1
1.1
1.2
1.3
1.4
1.5
1.6
1.7
1.8
1.9
Frequency (Hz)

Figure 16

INTERNATIONAL SEARCH REPORT

International application No
PCT/GB2023/050697

A. CLASSIFICATION OF SUBJECT MATTER
INV. A61B5/0507 A61B5/00
ADD. A61B5/08 A61B5/11

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

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)
A61B

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

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	WANG XUYU ET AL: "PhaseBeat: Exploiting CSI Phase Data for Vital Sign Monitoring with Commodity WiFi Devices", PROCEEDINGS OF THE INTERNATIONAL CONFERENCE ON DISTRIBUTED COMPUTING SYSTEMS, IEEE COMPUTER SOCIETY, US, 5 June 2017 (2017-06-05), pages 1230-1239, XP033123076, ISSN: 1063-6927, DOI: 10.1109/ICDCS.2017.206 [retrieved on 2017-07-13]	1-9, 15-19,25
Y	abstract page 1230, right-hand column, line 25 - page 1236, left-hand column, line 9 -----	10-14

-/--

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

[ ] See patent family annex.

* Special categories of cited documents :

"A" document defining the general state of the art which is not considered to be of particular relevance

"E" earlier application or patent but published on or after the international filing date

"L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)

"O" document referring to an oral disclosure, use, exhibition or other means

"P" document published prior to the international filing date but later than the priority date claimed

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"X" document of particular relevance;; the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone

"Y" document of particular relevance;; the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art

"&" document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
30 May 2023	07/06/2023

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

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INTERNATIONAL SEARCH REPORT

International application No
PCT/GB2023/050697

C(Continuation). DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	LIU JIAN JLIU28@STEVENS EDU ET AL: "Tracking Vital Signs During Sleep Leveraging Off-the-shelf WiFi", PROCEEDINGS OF THE 34TH ACM SIGMOD-SIGACT-SIGAI SYMPOSIUM ON PRINCIPLES OF DATABASE SYSTEMS, ACMPUB27, NEW YORK, NY, USA, 22 June 2015 (2015-06-22), pages 267-276, XP058511255, DOI: 10.1145/2746285.2746303 ISBN: 978-1-4503-3550-8	1,20-25
Y	page 267, right-hand column, line 13 - page 272, line 24 -----	10-14
A	HE YING ET AL: "WiFi Vision: Sensing, Recognition, and Detection With Commodity MIMO-OFDM WiFi", IEEE INTERNET OF THINGS JOURNAL, IEEE, USA, vol. 7, no. 9, 22 April 2020 (2020-04-22), pages 8296-8317, XP011809192, DOI: 10.1109/JIOT.2020.2989426 [retrieved on 2020-09-14] the whole document -----	1-25