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Noncontact Screening System with Two Microwave Radars for the Diagnosis of Sleep Apnea-Hypopnea Syndrome

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Abstract— **There were two key problems in applying Doppler radar to a diagnosis system for sleep apnea-hypopnea syndrome. The first is noise associated with body movements and the second is the body positions in bed and the changes of the sleeping posture. We focused on the changes of the amplitude of the radar output signal corresponding to the changes in the tidal volume, and proposed a method of detecting the change of the respiratory amplitude value without the influence of body position in bed. In addition, we challenged the detection of the apnea-hypopnea event confirmed by accompanied rise of heart rates. To increase the accuracy of heart rate measurement, we propose a new automatic gain control and a real-time radar-output channel selection method based on a spectrum shape analysis. A prototype of the system was set up at a sleep disorder center in a hospital and field tests were carried out with eight subjects. Despite the subjects engaging in frequent body movements while sleeping, the system was quite effective in the diagnosis of sleep apnea-hypopnea syndrome (the correlation coefficient *r* = 0.98).**

I. Introduction

The number of patients with sleep apnea-hypopnea syndrome (SAHS) has increased, and two to four percent of middle-aged adults are considered to be the potential patients [1]. SAHS can cause and worsen other medical conditions, such as hypertension, heart failure and diabetes, and it is preferable to diagnose for the treatment at the early stage. The diagnosis of SAHS is based on the results of a sleep study (polysomnography: PSG). However, it is difficult for a subject to keep in a normal sleeping mood with many contact sensors such as electro encephalography, electrooculography, ECG, pulse oximeter, respiratory airflow, and respiratory effort indicators. Hence, reliability and accuracy of the PSG test may be reduced. To overcome this problem, we have developed a non-contact heart rate and respiratory monitoring system with microwave Doppler radars as an easier alternative to the formal sleep study (PSG).

There are two key problems in applying microwave Doppler radar to night-time vital signs monitoring systems. The first major obstacle to mainstream application is noise associated with body movement. Irregular movements such as limb movement and turning over in bed can be mixed with respiratory and pulse wave signals. Thus, it is difficult to separate and extract the target signal. Attempted solutions have included the use of two radars – at front and back – to cancel out body movement [2]. However, this is hard to implement in the bed-based environment, and cancellation is problematic in practice due to morphological differences between the body front and back. The second main obstacle is the degradation of the accuracy of heart rate determination due to the null detection point (NDP) problem [3], respiratory harmonic interference, and intermodulation [2]. The researches that combined the orthogonal I/Q channels such as arctangent demodulation [4] or complex signal demodulation have accomplished certain advancement for the problem. However, these current researches were done under the condition of short-time measurements with subjects on chairs without a single move. The researches cannot be applied for long-time measurements with frequent change of positions and postures in bed.

In this paper we discuss techniques for measuring the heart rates and monitoring the changes of the amplitude of radar-output signal corresponding to the changes in the tidal volume – important for the development of microwave Doppler-based diagnostic system for SAHS. For the accurate measurement of heart rates, we propose a new automatic gain control (AGC) to reduce the influence of body movement and a real time selection method to choose the best radar-output channel out of four signals based on a spectrum shape analysis (SSA). Moreover, we propose a amplitude baseline method (ABM) for the detection of respiratory disturbances. We also present the results of a practical evaluation of the diagnostic system for SAHS with eight outpatients at a hospital.

II. Methods

The SAHS monitoring system is shown in Fig. 1. Two 24 GHz microwave Doppler devices (NJR4262J) were installed

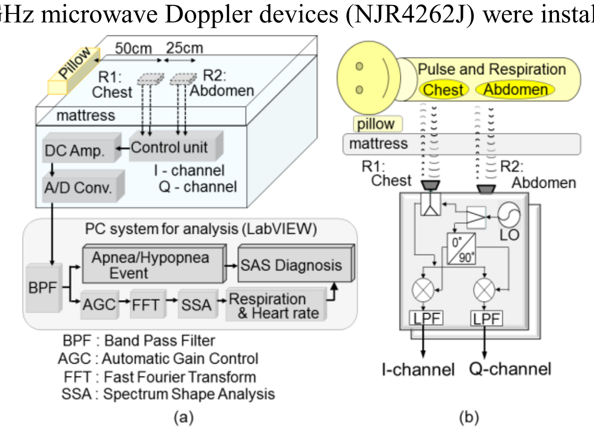

Figure 1. Monitoring system set-up: (a) Installation of radars and block diagram. (b) Two-channel Doppler-radar sensor and I/Q channels.

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separated from one another by the distance of 25 cm so that each may measure the vibrations of the chest and the abdomen respectively. The Doppler device incident power density was 1.5×10^{-2} mW/cm^2 at the body surface, and it was much lower than the Japanese electromagnetic wave safety guideline value of 1mW/cm^2 for radio frequency protection. The radar output signals were divided in two ways, and each signal was processed through band pass filters corresponding to respiration (i.e. 0.1 to 0.6 Hz) or pulse (i.e. 0.6 to 3.0 Hz). Fig. 1(b) shows the block diagram of a quadrature Doppler radar system. The local oscillator signal (LO) is divided by a two-way 90° power splitter to get two orthonormal baseband signals (I and Q: in-phase and quadrature baseband outputs) which are combined to demodulate the subject's motion.

Fig. 2 shows four radar output baseband signals with a sampling frequency of 100 Hz. Fig. 2(a) shows that slight vibrations due to heartbeat have been superimposed on the respiration wave-form. Fig. 2 (b) shows the example of the radar signal changed from normal breath to the apnea and hypopnea. Meanwhile, the radar output signal encounters regions of Null Detection Points (NDPs) at every quarter wave-length from the object [3]. As shown in Fig. 2(c), when either signal I or signal Q is in the optimum detection area, the other is in the NDP. Aside from the influence of NDPs, the radar output signal of the respiration and heartbeat drastically changed with body movements (Fig. 2(d)).

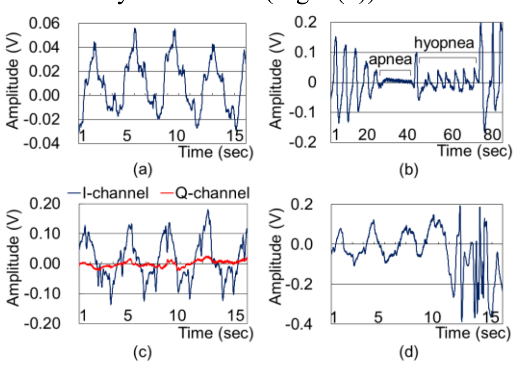

Figure 2. Samples of radar output signals: (a) Slight vibrations due to heartbeat were superimposed. (b) apnea signal (amplitude <20%) and hyopnea signal (amplitude <50%). (c) Characteristics of the I/Q channel signal. (d) Random body movement signal.

A. Automatic Gain Control

Cyclic chest motion due to respiration creates a displacement of approximately 1 mm, whereas chest and back vibrations induced by pulse does not exceed 0.02 mm. The radar output signals induced by random body movements are 10 to 1000 times larger than target signals. We determined maximum heartbeat output signal amplitude H_m during times without body movement for eight elderly subjects described below. In order to reduce radar output signals induced by random body movements, we introduced an automatic gain control (AGC), an amplitude auto-regulation method to limit the cardiac output signals to H_m. AGC processing was applied to each peak and trough when either the peak amplitude exceeded H_m or trough amplitude was less than - H_m (Fig. 3(a)). Each peak and trough was separated as a wave fraction from the zero amplitude point to the next zero amplitude point. When the max amplitude in the fraction is P, the amplitude was divided by P/H_m for each peak or trough. As a result, the frequency elements did not mix with the frequency that originally existed. AGC drastically improved the heartbeat peak extraction accuracy (Fig. 3(b)), because AGC enabled the frequency peak of the faint heartbeat hidden behind the large-amplitude body motion signal to be extracted through fast Fourier transform (FFT).

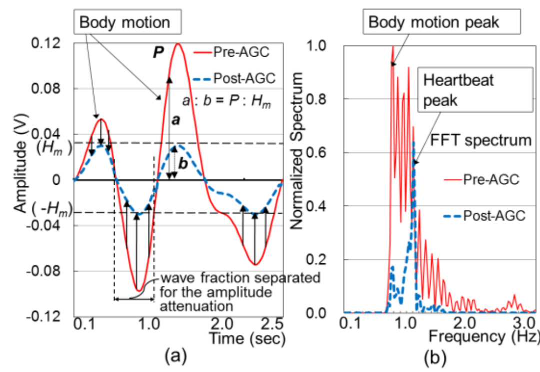

Figure 3. AGC processing: (a) When the max amplitude value of the wave fraction is P, if $P > H_m$ (=0.03V) then amplitudes at all sampling points of the wave fraction are divided by P/H_m. (b) AGC extracts the frequency peak of faint heartbeat hidden behind the large-amplitude body motion signal.

B. Spectrum Shape Analysis (SSA) for channel selection

We introduced a method for a real-time selection of the most appropriate radar channel from four radar output channels (I and Q channel of two radars) with least body motion influence, highest sensitivity, lowest respiratory harmonics, and least intermodulation interferences. For the accurate heart rate determination, we conducted a spectrum shape analysis (SSA) in order to select the most appropriate channel with the least number of large peaks and with the largest peak-to-peak ratio in the FFT spectrum. For a series of FFT spectrum peak values, P_1, P_2, P_3 … P_k and P_{k+1} (k is a positive integer), rearranged in order of decreasing height, we pick the smallest integer n which satisfies the inequality of $P_i > 3P_{i+1}$, where i runs from 1 to k. Then we say that the number of large peaks in the spectrum is n. This method is an empirical procedure to select the most appropriate channel by evaluation parameters V for SSA, defined as follows: P_t is an integral of the FFT spectrum of radar output signal from 0.6 to 3.0 Hz. P_1, P_2 and P_3 are the power values of large peaks.

(a) $V = 1 - (P_1/P_t)$, $(0<V<1)$ when only one large peak exists; (b) $V = 1 + (P_2/P_1)$, $(1<V<2)$ when two large peaks exist; (c) $V = 2 + (P_2/P_1) + (P_3/P_1)$, $(2<V<4)$ when three large peaks exist; and (d) $V = 5$ when four or more large peaks exist.

The FFT spectra of radar output signals corresponding to formulas above are shown in Fig. 4. A mono tone periodic wave was dominant in the real-time heartbeat signal while harmonics and intermodulation waves were not observed. Therefore, the corresponding FFT spectrum had only one large peak (Fig. 4(a)). In contrast, basic heartbeat waves were interfered with harmonics and intermodulation waves. Then, the corresponding FFT spectrum had two or more large peaks (Fig. 4(b), (c)). In addition, the main frequency element is not contained in the signal of the channel in NDP, a lot of spectrum peaks appear under the influence of the white noise or irregular body movements (Fig. 4(d). A radar channel with the lowest V value is selected automatically as the most appropriate channel. Therefore it could be said that SSA is effective in avoiding NDP, harmonics and intermodulation.

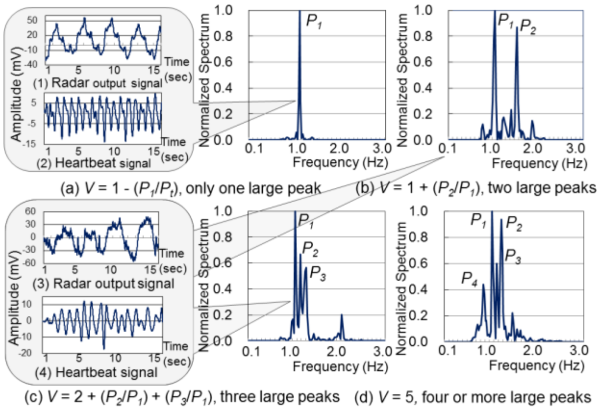

Figure 4. Four types of FFT spectral shapes with heart beat signals and evaluation parameters V according to the number of large peaks. Insets: real time radar signal (1) and heartbeat signal (2) correspond to to the only one large spectrum (a). Similarly, radar signal (3) and heartbeat signal (4) correspond to (b) and (c). (d) a lot of spectrum peaks are flooded when the channel is in NDP or target signals are interfered by body movements.

C. The diagnostic criteria for apneas and hypopneas with Doppler radar measurement

There are three forms of sleep apnea: central (CSA), obstructive (OSA), and mixed sleep apnea (MSA) constituting 0.4%, 84% and 15% of cases respectively [5]. OSA is the most common type of sleep apnea and is caused by obstruction of the upper airway in the presence of respiratory efforts, and has the feature that the rib cage and the abdomen move independently and it often shows the paradoxical movement of the reversed phase. The apnea-hypopnea index (AHI) is a sleep apnea-hypopnea severity score that combines apneas (pauses in breathing, or a 80% reduction in airflow) and hypopneas (a 50% reduction in airflow). These apneic events in breathing must last for 10 seconds or more. The AHI is calculated by dividing the number of events by the number of hours of sleep. (AHI values are typically categorized as 5-15/hr = mild; 15-30/hr = moderate; and > 30/hr = severe.) To detect the apneic events, we propose the amplitude baseline method (ABM) as follows. A respiratory baseline is set by the mean value of the amplitude of breathing radar signal in latest 120 seconds without the body movement and the apneic events. The diagnostic criteria for apneas and hypopneas with Doppler radar measurement includes a reduction (>50% or >80%) in the respiratory amplitude from the baseline. This determination is executed about the radar channel with the maximum output power in four channels. And, a supplementary decision is performed in the case with the paradoxical movement of the reversed phase and the breathing disorder accompanied by rise of heart rates.

D. Clinical testing of the dual radar system

We conducted a clinical test on eight elderly subjects (mean age, 87 years; range, 73-101 years, six females and two males) at a nursing home. Each subject lying on bed was monitored using the dual radar system between 9 pm and 5 am the following day. For eight hours, his or her Holter ECG was recorded simultaneously using electrodes as a reference. In addition, the diagnosis of sleep apnea-hypopnea syndrome with radar was compared with PSG for eight outpatients (mean age, 51 years; range, 45-63 years, two females and six males) at a sleep disorder center in a hospital. Ethics approval for the research was provided by the Tokyo Metropolitan University ethics committee and a written informed consent was obtained from each subject.

III. Results

A. Accuracy evaluation of heart rate

We determined the Pearson correlation coefficients between the heart rates determined by radars with four various conditions and those derived from Holter ECG. The Doppler heart rate was calculated every second with FFT: the signal in a 40 s Hamming window was processed, and the local maximum in the frequency domain was used to calculate the heart rate. The correlation coefficient of the heart rate was calculated based on both ECG and Doppler heart rates per second for the eight-hour measurement period (28,800 seconds). Pre-AGC indicates the average of four correlation coefficients of each raw radar output signal without AGC , and post-AGC indicates the average of four correlation coefficients of each radar output signal with AGC. The conventional method indicates the combination of I/Q channel selection with the multiplied FFT spectrum [6]. Correlation coefficients of post- and pre-AGC were 0.35±0.018 and 0.30±0.019 respectively. The correlation coefficients of post-AGC were significantly higher than those of pre-AGC (t-test p-value: $p<0.001$). Correlation coefficients of the conventional method +AGC were 0.56±0.035 and significantly higher than those of post-AGC, but did not reach 0.6. Correlation coefficients corresponding to SSA+AGC were 0.70±0.025 and significantly higher than those of the conventional method ($p<0.001$). The combination of AGC with SSA showed the highest correlation coefficients (Fig. 5).

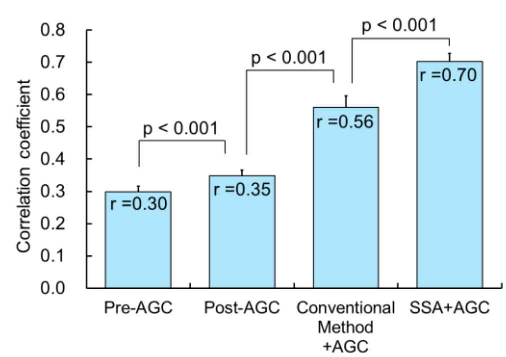

Figure 5. Correlation coefficients between the heart rates determined by radars and those derived from Holter ECG with eight elderly subjects

B. Evaluation of diagnosis accuracy of SAHS

Fig. 6 shows the radar output signals for the apnea event at the hospital. Because the respiratory effort was not observed in the example Fig. 6(a), it is thought to be a central sleep apnea (CSA). In contrast, an independent respiratory movement (paradoxical movement) of the chest and the abdomen was observed in the example Fig. 6(b), and this is thought to be an obstructive sleep apnea (OSA).

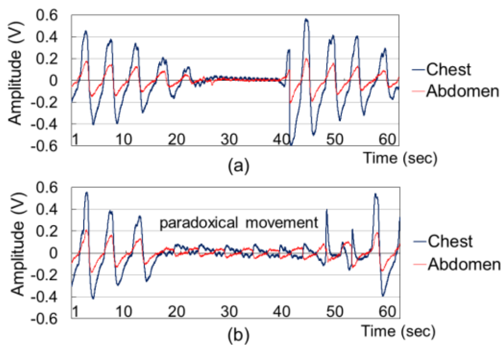

Figure 6. Radar signals of chest and abdomen walls during apnea. These are a typical central sleep apnea (a) and an obstructive sleep apnea with paradoxical movements (b).

The clinical test data collected at the hospital are shown in Table 1. The comparative tests of radar and PSG were executed with eight subjects from 10:00PM to 6:00AM the following day. We used the respiratory disturbance index (RDI) which indicates the number of apneic events per hour of recording (and not per hour of sleep). It was more adequate than AHI to compare the radar and PSG methods. For each subject, the mean value of RDI and the standard deviation were calculated with radar and PSG respectively. It could be said that RDIs obtained from radars and PSG correlated very closely (the Pearson correlation coefficient $r = 0.98$).

Table1. RDIs calculated with the radar signal and the PSG

Subject	S1		S2		S3		S4	
Time	Radar	PSG	Radar	PSG	Radar	PSG	Radar	PSG
22:00~	91	88	5	4	41	49	20	17
23:00~	96	89	2	1	48	46	11	16
0:00~	73	79	11	9	38	46	8	9
1:00~	58	54	24	20	26	29	2	1
2:00~	9	6	4	6	38	40	18	19
3:00~	17	22	32	40	15	17	63	55
4:00~	33	31	17	16	37	31	25	26
5:00~	57	59	27	33	35	31	34	31
mean	54.3	53.5	15.3	16.1	34.8	36.1	22.6	21.8
std dev	32.4	31.4	11.5	14.1	10.1	11.0	19.2	16.3

Subject	S5		S6		S7		S8	
Time	Radar	PSG	Radar	PSG	Radar	PSG	Radar	PSG
22:00~	19	21	24	30	17	23	5	0
23:00~	42	36	3	0	33	30	95	92
0:00~	75	73	7	15	9	12	91	99
1:00~	60	58	25	20	25	23	74	81
2:00~	21	29	9	12	30	33	82	78
3:00~	39	38	10	11	53	50	77	73
4:00~	48	49	28	24	6	7	73	74
5:00~	28	21	14	12	32	29	66	64
mean	41.5	40.6	15.0	15.5	25.6	25.9	70.4	70.1
std dev	19.4	18.3	9.4	9.2	15.1	13.2	28.1	30.4

The units are times per hour.

IV. Discussion

The SSA method was the most successful at extracting the heart rates. Conceivable reasons are as follows. The vibration surface caused by the pulse wave is very narrow, and the radar can detect the heartbeat signal well when the vibration surface is orthogonalized in the direction of the radar antenna. Therefore, under the condition that the position and the sleeping posture in bed always being changed while sleeping, it is not a priority matter to combine channel signals which are uncertain to capture the target signal, but to select the channel which definitely catches the heartbeat signal well in real time.

Many electrodes of PSG restrain the patient and obtrusive contact electrodes could disturb the measurement itself. We have proposed the non-contact heart rate and respiratory monitoring system with two radars as an easier alternative to PSG. We showed that RDIs obtained from radars and PSG correlated very closely. This result is thought to be the advancement in the SAHS diagnostic approach. However, the case where RDI with radar is less than that by the PSG measurement might lead the SAHS patients to misdiagnosis on the screening. Therefore, our following challenge is to improve the diagnosis system to extract the respiratory efforts from the disorder of corrugated shape compared with usual breathing and paradoxical movement of the rib cage and the abdomen in OSA.

V. Conclusion

We conducted a clinical test at the hospital in order to evaluate the microwave Doppler-based diagnostic system for SAHS that can deal with body movements and positional change in bed during sleep, using two radars and the newly developed analysis methods, i.e. the amplitude baseline method (ABM) and the combination of AGC with SSA. We achieved an accurate diagnosis with RDI of SAHS (the Pearson correlation coefficient $r = 0.98$) by observing the respiratory movement of the chest and the abdomen separately with two radar. Our non-contact diagnosis system for SAHS shows great promise as a new screening system.

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