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Detection of Cardiac Activity using a 5.8 GHz Radio Frequency Sensor

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Abstract—**A 5.8-GHz ISM-Band radio-frequency sensor has been developed for non-contact measurement of respiration and heart rate from stationary and semi-stationary subjects at a distance of 0.5 to 1.5 meters. We report on the accuracy of the heart rate measurements obtained using two algorithmic approaches, as compared to a reference heart rate obtained using a pulse oximeter. Simultaneous Photoplethysmograph (PPG) and non-contact sensor recordings were recorded over fifteen minute periods for ten healthy subjects (8M/2F, ages 29.6±5.6 yrs) One algorithm is based on automated detection of individual peaks associated with each cardiac cycle; a second algorithm extracts a heart rate over a 60-second period using spectral analysis. Peaks were also extracted manually for comparison with the automated method. The peak-detection methods were less accurate than the spectral methods, but suggest the possibility of acquiring beat by beat data; the spectral algorithms measured heart rate to within ±10% for the ten subjects chosen. Non-contact measurement of heart rate will be useful in chronic disease monitoring for conditions such as heart failure and cardiovascular disease.**

I. Introduction

Recent advances in radio-frequency engineering and signal processing technologies, have created the possibility of using Doppler-based radio-frequency sensors for non-invasive health monitoring [1]. The basic principle is that a radio-frequency sensor can provide a sensitive measurement of the underlying motion of a subject, and signal processing techniques can then be used to determine parameters such as gross bodily movement, respiration and even movement due to individual heart beats. Hence, Doppler radio-frequency sensing could enable a truly contactless application in healthcare technology, home monitoring and security and search scenarios. The main challenge in using the Doppler radar for vital signs detection is the analysis and processing of the data [2]. This paper focuses on the issue of detecting the heart rate of a subject, when the subject is stationary or semi-stationary (e.g., sleeping or working quietly at a desk). The technical challenge to overcome is isolating the heartbeat signal from other moving objects which cause interference and distortion [3].

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There has been a lot of previous interest in the detection of vital signs using radio-frequency sensing. Some of the signals processing techniques being currently investigated include time-frequency analysis, wavelets, fuzzy logic and neural networks [3]. It has been demonstrated that with the Doppler radar, an agreement of 88% can be achieved with the reference heart rate [4]. The signal processing technique used in that study is based on autocorrelation and uses several enhancement techniques, including a centre clipper [4]. Petrochilos *et al*. have compared ICA and RACMA methods of signal processing and found that ICA works at least as well as RACMA [5]. Brink *et al*. claim that their method, seismosomnography (SSG), has practical limitations whereby the maximum detectable heart rate is only 120 beats per minute and does not allow tolerance for all different scenarios of patients. They also observe considerable lack of agreement between the interbeat-interval variability measures of SSG and ECG. Most of these algorithms need to be further explored as current applications are limited due to the presence of multiple sources of interference in real applications outside the lab [6].

As a first step, in this paper, we will explore the performance of a sensor and signal processing algorithm for detection of heart rate in stationary subjects, as compared to a simultaneously acquired reference signal using photoplethysmography.

II. Methodology

A. Radio Frequency Sensor

The signals analyzed in this paper were acquired using a custom-designed 5.8-GHz Doppler radar sensor. The sensor is a proof-of-concept prototype aimed at evaluating different aspects and problems associated with remote sensing of human vital signs, especially regarding prospective signal levels, and additional functionality such as restriction of the measurement zone of interest. The sensor operates in a continuous wave mode, with an average emitted power level of approximately 8 mW .

The reflected radio frequency (RF) signal to the sensor from a nearby moving object can be received by either a patch or horn antenna. The received RF signal is demodulated to result in an I and Q channel baseband signal showing the Doppler shift which reflects movement of the subject. The measured signal is then amplified by three different amplifiers. The first amplifier provides a voltage gain of 1000. The second amplifier removes any remaining DC offsets by slowly tuning the offset voltage pin of the first amplifier. The final unity-gain operational amplifier is an active filter that limits the bandwidth of the IF signal to 100 Hz, both for noise performance and anti-aliasing purposes. The physical design of the sensor and the experiment set up is shown in Figure 1. In this study, two custom-designed horn antennas were used as the transmit and receive antenna. The gain of the horn antennas was 10 dB ±1 dB. When operating, the sensor produces two differential signals at its output which are 8 V voltage swing peak-to-peak, ±4 V with reference to system ground. The output signal voltage is fed into a data acquisition unit.

B. Data Acquisition

The National Instruments USB-6211 was used for data acquisition in this experiment. The NI USB-6211 is a plug-and-play device that can be installed directly to a USB drive on a computer to perform data acquisition. Wires representing each acquisition channel can be directly connected to terminals on the NI USB-6211. The NI USB-6211 samples the data at a chosen sampling rate and makes them available digitally to a suitable computer running Matlab or other data-logging software, for example NI LabVIEW SignalExpress. Data that is acquired can then be analyzed and signal processing can be carried out.

C. Measurements

Simultaneous photoplethysmograph (PPG) and radio-frequency sensor recordings were recorded for ten subjects consisting of eight male and two female, with a mean age of 29 years (SD5.75). The PPG reading was obtained using a pulse oximeter which illuminates skin with a red light from a light-emitting diode (LED) and measures the changes in light absorption. The amount of light reflected by the LED indicates the change in volume caused by the blood pressure pulse. Each cardiac cycle appears as a peak on a PPG reading. The PPG is used in this study as an accepted standard reference against which the non-contact signals are compared. The experiment was carried out in a laboratory setting with no nearby moving objects or people. The subjects were seated 70cm from the sensor antennas. The sensor antenna was placed on a table 93cm above the ground.

The non-contact sensor detects the raw movement of the person's chest, dominated by breathing and cardiac movement. A component of this movement will be the ballistocardiogram (BCG) signal from the surface of the skin. The BCG signal is a result of chest movement due to shifts in the centre of mass of the blood which is caused by cardiac activity. Any movement in a human body will cause disturbance to the BCG signal which is very weak compared to the signals caused by bigger body movements. These disturbances were minimized by making the subject feel as comfortable as possible during the measurement by remaining still in seat, staying relaxed and breathing normally throughout the course of the experiment, with little or no movement at all. BCG signal detection is confounded by the presence of a strong respiratory component, so it is necessary to separate the heartbeat signal from the respiration signal for analysis [7].

	Age (years)	Height (cm)	Weight (kg)	BMI
Mean	***29***	***172.7***	***64.55***	***21.25***
Std Dev	***5.75***	***11.79***	***16.64***	***3.15***

Table 1: Summary of subject details

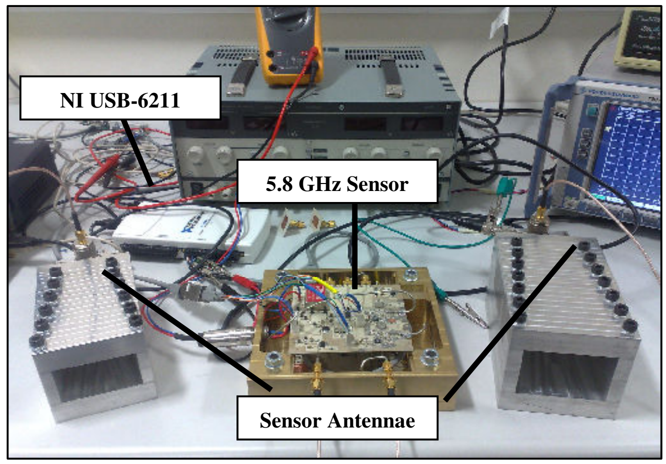

Figure 1: Experiment set-up in the lab

In this protocol, measurement took place over a 15 minute period, and the sampling rate was 60Hz. For the detection of heart rate, the sensor is somewhat sensitive to positioning and distance. However, initial feasibility experiments showed that if placed within a distance of 50cm and 100cm from the subject, the sensor can measure heart rate. When using the separate transmit and receive horn antennas, the antennas are positioned in a 30 degree angle facing the subject's chest. The sensor was non-invasive and none of the subjects reported any discomfort caused by the sensor (emitted power levels are orders of magnitude below recommended safety limits for non-ionising radiation at 5.8 GHz). Figure 2 shows an example of approximately 10 minutes of data obtained from Subject 1 using the non-contact sensor. The top two axes are the non-contact I and Q channels (NC-I, NC-Q) respectively, and the third axis is the PPG signal. The lower three axes are the respective spectrograms, which only show the information above 0.5 Hz (i.e. the respiratory component is not visible). In the spectrograms, the cardiac information is typically present at around 1.1 Hz or 65bpm. It is present on both the I and Q non-contact channels and the PPG channel. It is evident that some cardiac information is being detected by the non-contact sensor.

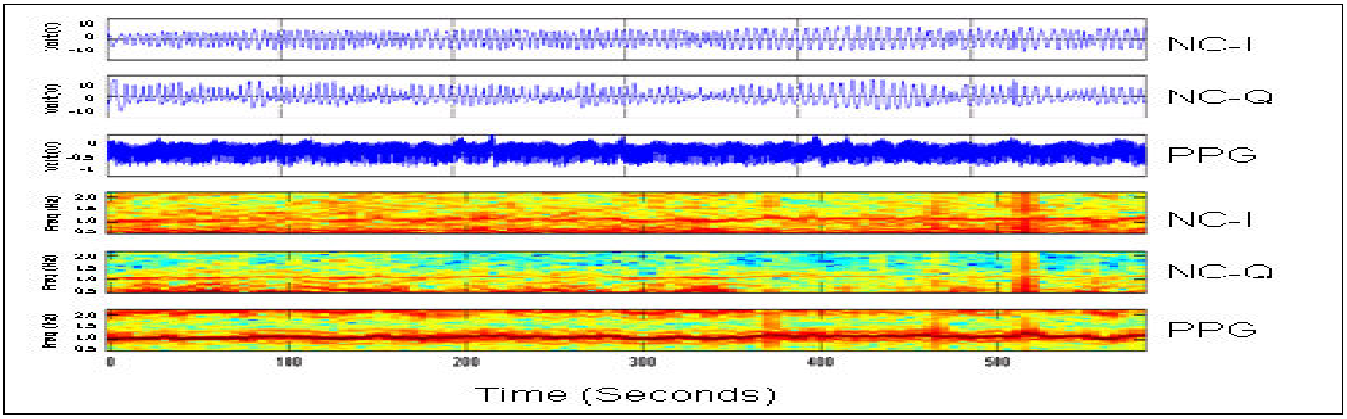

Figure 2: Spectrogram of data for Subject 1.

D. Signal processing

Measurement of heart rate using a non-contact radio-frequency sensor would open a number of interesting possibilities in health monitoring. In this paper, we investigate the accuracy of our proposed sensor and method for heart rate monitoring

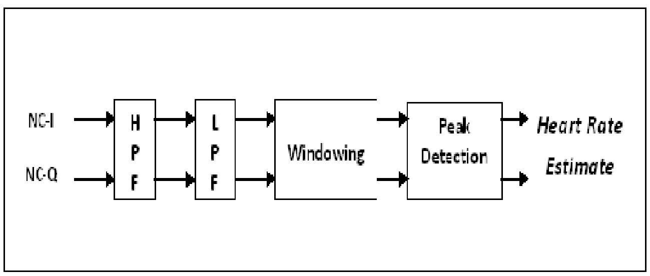

Figure 3: Block diagram of signal processing steps used in this paper.

First, sections of data containing movement are removed using visual inspection. A low-pass filter that filters signals with frequencies of 10Hz or higher was applied to both PPG and non-contact data to remove high frequency signals in the spectrum. The non-contact data is further processed whereby the signal is effectively band-passed by applying a high-pass followed by a low-pass filter to pass signals with frequencies between 1 and 2.5Hz. This selects the range from 60-150 beats per minute and removes the breathing frequency, thus allowing an easier extraction of the cardiac activity. A seventh-order low-pass Butterworth filter with a cutoff frequency at 1 Hz and an eleventh-order Butterworth filter with cutoff frequency at 2.5Hz were used. To ensure that the filtered data has zero phase distortion, after filtering in the forward direction, the filtered sequence is reversed and run back through the filter. This also modifies the magnitude by the square of the filter's magnitude response. Startup and ending transients are minimized by matching initial conditions. Figure 4 shows a power spectrum of a 30 second segment of the filtered signal. It indicates a main frequency component at approximately 1.1Hz (marked by the red circle in Figure 4) which represents the cardiac signal.

We investigated two techniques for heart rate detection. The first method is based on detection of individual peaks associated with each cardiac cycle. To achieve this, the raw data is segmented into 15 second windows with 1s overlap (segments of 900 samples). All signal processing was done within this window. A signal processing algorithm that detects peaks and troughs was applied to both the PPG and non-contact data. A threshold value was set by looking in a rolling window of data whose optimal window length depends on the sample rate. Then, the algorithm looks in the entire data set and assigns peaks based on the threshold value. A peak is considered valid if it begins below threshold, exceeds the threshold at some point, and then returns to a value below threshold. The maximum and minimum distance by which peaks can be separated are 1s (60bpm) and 0.4s (150bpm) respectively. The indices of the beginnings of the peaks and the total number of peaks found is then stored. To extract interpeak intervals, the number of discrete sample points between two subsequent troughs is counted. Based on the interpeak intervals, the heart rate can be derived. In order to avoid extracting incorrect peaks, the peaks must be pronounced and clearly distinguishable. Peaks are marked with red circles.

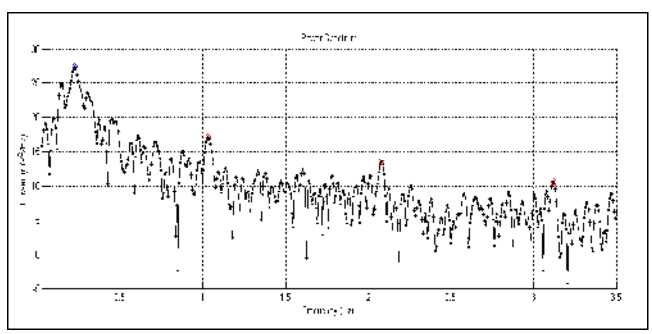

Figure 4: Power spectrum of 30 seconds of raw data showing peaks at frequencies corresponding to cardiac and respiration signals.

The peak marked with a blue circle at approximately 0.23Hz corresponds to respiration (14 breaths per minute). The first peak marked by a red circle at approximately 1.05

Hz indicates cardiac signal (63 beats per minute) and the subsequent red circles are harmonics of the cardiac frequency. A second technique was to take the power spectral density estimate of 60 second segments of the band pass filtered signal. A cardiac frequency was then selected as the frequency at which a peak occurs between 0.83Hz (50 bpm) and 2 Hz (120 bpm).

E. *Performance Measure*

The performance measure for the heart signal is epoch based. The data is segmented into 60 second epochs for both the PPG and non-contact channel signals. For the peak-detection methods, the difference between the number of peaks in the PPG signal and non-contact signals is calculated. The comparison is a measure of the accuracy of the non-contact channel data with reference to the PPG data. The absolute mean error is the difference in the number of peaks detected in the non-contact channel with reference to the PPG signal epoch. The channel with a lower error rate is more accurate in heart rate detection. A value of 0 indicates perfect agreement.

$$Error(beats\ \ per\ \min) = \left| PPG - Non - contact \right|$$

Or

$$Error(\,percentage) = \left| ((PPG - Non - contact) \div PPG) \times 100 \right|$$

III. RESULTS

Manual annotation was carried out for a random 60s period for all the 10 subjects to confirm that the non-contact sensor is indeed detecting heart rate. Manual annotation was done by visually inspecting the filtered signals and calculating the number of peaks in each 60 second segment manually. Table 3 shows the number of peaks found in a 1-minute segment for each subject (corresponds to human heart beat per minute), using the peak detection method described earlier and also manual annotation.

	Peak Detection Method (bpm)			**Manual (bpm)**		
Subject	**NC_I**	**NC_Q**	**PPG**	**NC_I**	**NC_Q**	**PPG**
1	63	51	66	65	65	65
2	57	55	62	62	62	62
3	48	54	55	48	56	56
4	47	69	80	48	69	81
5	39	53	64	51	64	64
6	53	49	79	60	59	81
7	56	56	80	76	70	82
8	66	56	83	72	58	83
9	47	100	67	66	67	67
10	68	67	77	71	72	77
Mean	***54.4***	***61***	***71.3***	***61.9***	***64.2***	***71.8***

Table 2: Manual annotation of heart rate

From Table 2, it is evident that manual annotation gives more accurate values with reference to the PPG heart rate than the peak detection method although some subjects still performed poorly for both. The advantage of having cardiac activity detection on a beat-by-beat basis is considerable – this would allow monitoring of conditions where inter-beat variability is important (e.g., subjects with atrial fibrillation).

	Mean Value (bpm)	**Mean Error (beats per minute error)**		**Standard Deviation (beats per minute)**	
Subject	**PPG**	**NC_I**	**NC_Q**	**NC_I**	**NC_Q**
1	65.33	3.6	5.43	3.73	4.72
2	63.73	2.75	3.84	2.43	2.69
3	55.18	5.21	10.58	3.88	8.79
4	81.67	27.94	9.74	7.57	7.31
5	61.75	12.7	5.41	5.61	3.71
6	85.94	15.78	35.17	10.74	5.52
7	82.07	17.82	21.55	7.75	9.21
8	79.75	9.85	22.57	5.45	5.99
9	66.22	8.42	32.14	5.07	17.22
10	75.26	5.51	5.36	6.68	7.04
Mean	***71.69***	***10.96***	***15.18***	***5.89***	***7.22***

Table 3: Results from peak detection method

Furthermore, Table 4 shows that a spectral technique which considers a longer period of time can be highly reliable in detecting the heart rate of the subject. The results show that the non-contact sensor can reliably detect heart rate but a more reliable algorithm is needed for heart rate estimation.

Results from the automated peak detection method show that the mean value of the heart rate from the PPG is 71.7 beats per minute across all 10 subjects. The mean absolute error for the non-contact channels based on the PPG value is 11.0% and 15.2% for channel I and Q respectively. The mean absolute error for the non-contact channels based on the PPG value from spectral analysis 10.4% and 7.8% for channel I and Q respectively. As an illustration of the potential of the method, the mean error for Subject 2 is low (2.8% for the I-channel and 3.8% for the Q-channel), which demonstrates that the non-contact sensor is very useful for heart rate detection.

		Heart Rate(bpm)			**Error (%)**	
Subject	**Length(s)**	**PPG**	**NC-I**	**NC-Q**	**NC-I**	**NC-Q**
1	60.00	62.67	61.39	62.06	3.95	1.70
2	60.00	63.25	63.20	63.16	1.22	0.35
3	60.00	53.42	54.59	53.49	3.68	1.67
4	60.00	84.12	72.66	81.49	14.20	3.70
5	60.00	57.82	57.57	54.52	8.36	7.14
6	60.00	89.78	73.86	54.79	18.13	38.88
7	60.00	79.52	66.57	71.99	18.50	9.72
8	60.00	83.35	62.98	75.22	24.30	10.03
9	60.00	66.60	60.37	65.44	9.71	3.75
10	60.00	76.08	75.35	75.91	1.69	0.79
Mean	*60.00*	*71.66*	*64.85*	*65.81*	*10.37*	*7.77*

Table 4: Results from spectral analysis

Conversely, the mean error for Subject 9 is very high (8.4% for the I-channel and 32.1% for the Q-channel). On visual inspection of this record, the peak detection method failed to identify some peaks that occurred although there is clear evidence of peaks in the non-contact channels. There is also clear evidence that the sensor tracks varying heart rate as shown in Figure 5. From Figure 5, it can be seen that at there is a clear trace of cardiac activity at around 1.1Hz which drops to 0.9 Hz at 570s. Both the non-contact channels also track this change in cardiac activity. Thus, further improvement of the peak detection method may yield higher performance.

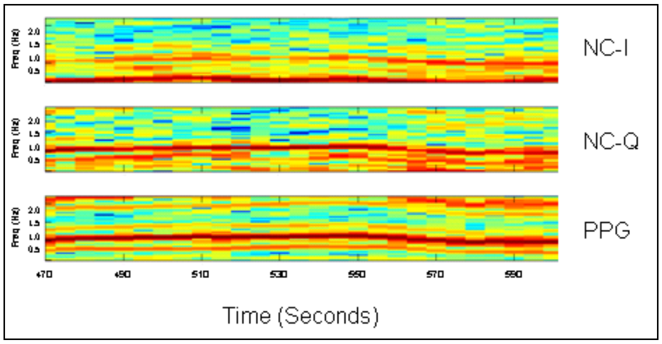

Figure 5: Tracking heart rate

IV. DISCUSSION

Figure 6 shows a section of the non-contact channels and the PPG signal in the time-domain. The top two axes are I and Q respectively, the third axis is the PPG signal. The number of peaks in 10s of data was manually calculated for each of the three axes. The peaks were calculated for the period of 10s between 300s and 310s.

Based on non-contact signal I between 300s and 310s:
Number of peaks = 0.5 + 10 = 10.5
Based on non-contact signal Q between 300s and 310s:
Number of peaks = 10 + 0.5 = 10.5
Based on PPG signal between 300s and 310s:
Number of peaks = 0.5 + 10 = 10.5

This demonstrates that the non-contact sensor has the ability to detect cardiac information on a beat per beat basis. Thus, a more reliable algorithm can be deployed on the non-contact data to estimate the heart rate.

V. CONCLUSION

It has been shown that heart rate information can be measured using a non-contact radio-frequency sensor and the signal processing methods described above. The use of the non-contact sensor was also convenient and unobtrusive. Future validation includes obtaining measurements from the non-contact sensor in different scenarios including from a moving target and with paced breathing. The non-contact respiration signal will be compared with a conductance band system as simultaneous reference. The capability of the sensor in real-time applications will also be investigated.

Proposed further algorithm development includes improving signal quality by detection and removal of movement intervals. Channel confidence measure will be developed such that the I or Q channel with the best signal is selected. This will adapt over time as the I/Q relative quality varies with respect to subject position.

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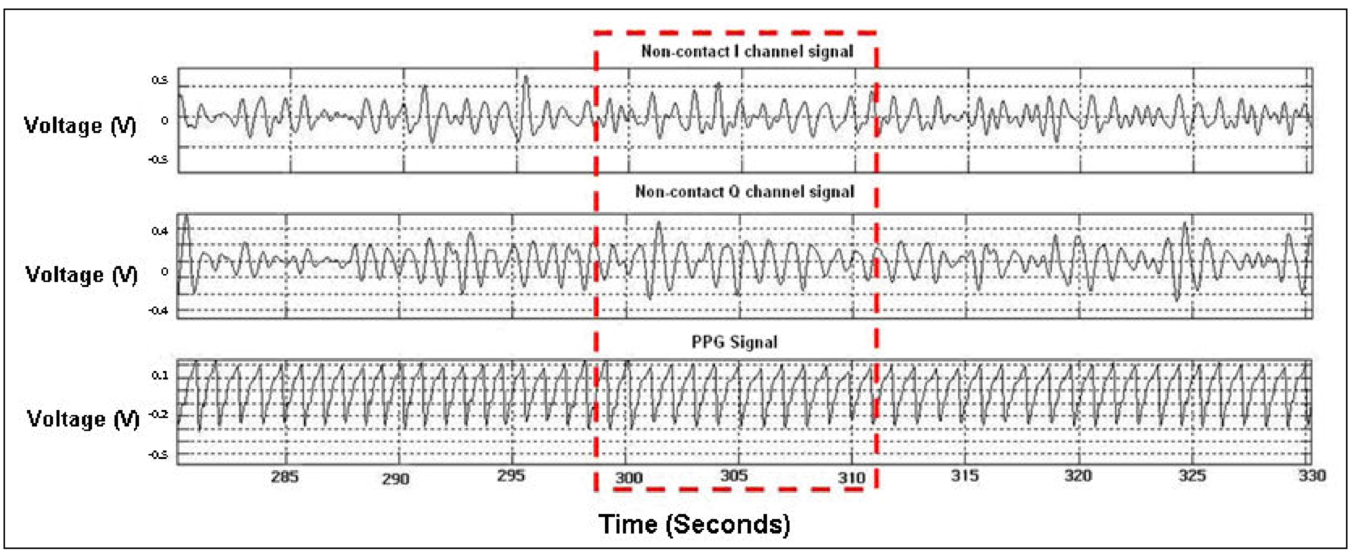

Figure 6: Data section in time-domain

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