Non-Invasive Monitoring of Arterial Stiffness with mmWave Radar

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***Abstract*—Non-contact monitoring of vital signs has been increasing demand in healthcare applications. Monitoring biomarkers such as heart rate variability (HRV) or pulse wave velocity (PWV) in home scenarios is crucial for early detection of cardiovascular diseases or the worsening of these diseases. This paper presents a mmWave radar network to measure pulse transit time (PTT) and PWV without contact. The experiment is carried out measuring simultaneously the carotid, the heart and the femoral of a healthy subject. Results reflect that the system is capable of measuring the carotid-femoral PWV (cfPWV), which is the *gold standard* for measuring arterial stiffness.**

I. Introduction

The COVID-19 pandemic showed the importance of adopting remote patient monitoring technology, with the aim of reducing the pressure on hospital care facilities. Therefore, home monitoring and telehealth, which are possible by non-contact monitoring, offer solution for early detection of health complications, reducing hospital costs and bed days of care [1].

The advances on radar technology during the last decades make this technology a strong candidate to address this issue. Radar monitoring allows not only to extract heartbeat and respiration [2], but also to measure other biomarkers that can provide significant information in the detection of certain cardiovascular diseases such as heart rate variability (HRV) [3] or pulse wave velocity (PWV).

PWV is widely used as an indicator of arterial stiffness and has been shown to predict mortality from cardiovascular disease, ischemic heart disease, stroke and atherosclerosis [4]. According to data provided by the World Health Organization (WHO) for the 2000-2019 period [5], approximately 40% of deaths are attributed to cardiorespiratory diseases, with cardiovascular diseases accounting for almost a third of these cases. Ischemic heart disease stands out as the leading cause of global mortality, responsible for 16% of total deaths worldwide, with 8.9 million deaths reported in 2019. This is followed by stroke, which accounts for 11% of total deaths [5], as illustrated in Fig. 1. Therefore, being able to monitor the PWV can play a key role in the early detection of these diseases, offering an advantage in timely intervention and treatment.

Moreover, it has been demonstrated that PWV is inversely related with blood pressure (BP) and can be used for an indirect estimation of this indicator [6]. Carotid-femoral pulse wave velocity (cfPWV) is considered the *gold standard* for measuring aortic stiffness [4]. PWV can be monitored with invasive methods such as pressure catheter recordings or with electromechanical solutions which require direct contact with the patient's tissue [7]. It has been reported that these probes require to be placed over the widest pulsation area and require support from solid structures such as the bones, being uncomfortable for the patient. Additionally, there are other non-invasive techniques for monitoring PWV based on the time delay, which is also known as pulse transit time (PTT), that a pulse experiences when traveling between two different locations. These techniques involve the use of electrocardiograms (ECG) and photoplethysmograms or piezoelectric pressure transducers. However, all of these methods require direct contact with the subject [7].

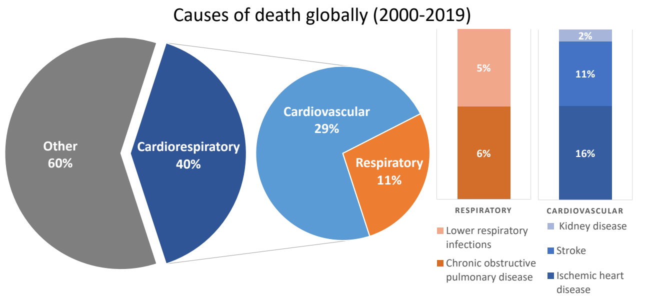


Fig. 1. Causes of death globally reported by the WHO in the 2000-2019 period.

For that reason, this study proposes a 3-node mmWave radar network for non-contact measuring the PWV of a patient. The PTTs are measured simultaneously pointing to the carotid, heart and femoral. Moreover, a 3-lead ECG and the pulse pressure measured in a finger are recorded as reference. The radar node is described in Section II. The rest of the paper is organized as follows. Section III describes the algorithm followed to calculate the PTT and PWV. The results are presented and discussed in Section IV, and conclusions are drawn in Section V.

II. Hardware

This study is carried out using a radar network which is composed by three radar nodes, that are synchronized using an external clock.

A. Radar module

The radar module used is based on a 124 GHz FMCW radar developed by Silicon Radar [8]. The radar modifications carried out are described in [3]. The RF front-end used is the commercial TRA_120_045 Microwave Integrated Circuit (MMIC) from Silicon Radar. The main radar characteristics are summarized in Table I.

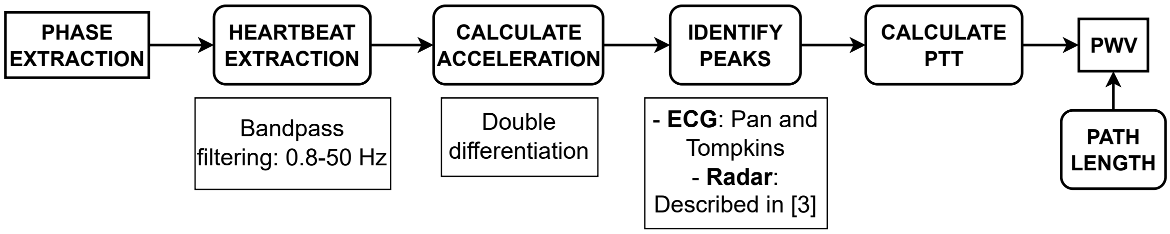


Fig. 2. Signal processing flow followed to obtain the PWV measurement with the radar.

TABLE I
MAIN CHARACTERISTICS OF THE RADAR MODULE USED.

Characteristic	Value
Model	*TRA_120_045*
Central Frequency (GHz)	124
Output Power (dBm)[a]	-3
Bandwidth (GHz)	20 (max)
Sweep Time	12 μs to 18 ms
Sampling frequency	10 MHz
Beamwidth (with the lens)[b]	4°(E) 4°(H)
Lens Gain	12 dB
Board dimensions (cm)	5x5

[a] Chip output power (without antennas).
[b] The use of dielectric lens allows a narrower focusing.

B. *Task Force Monitor*

The reference signals are acquired using the Task Force Monitor (TFM) developed by CNSystems [9]. This system measures synchronously the ECG and the BP obtained from the middle finger. The radar sensor and the TFM are synchronized using a gold-code sequence, which is generated in the clock board.

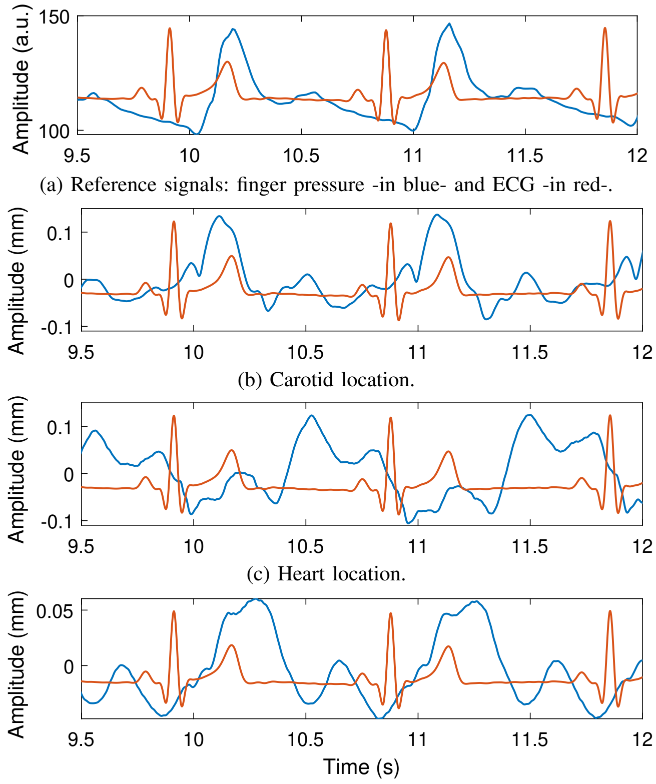
III. SIGNAL PROCESSING

This section describes the procedure followed to calculate the PTT measures from the radar and the TFM. The signal processing flow is shown in Fig. 2.

The signal acquired with the radar is obtained using the same procedure described in [3]. The displacement signal obtained from the radar is filtered from 0.8 to 50 Hz to obtain the heartbeat signal. After that, this signal is double differentiated to obtain the acceleration signal, where the maxima are identified. The heartbeat waveforms extracted for each location and the identification of the aforementioned points are exemplified in Fig. 3 and 4. It is important to highlight that, in order to mitigate the selection of false pulses induced by noise or motion artifacts, a minimum distance between maxima is enforced to be greater than $1/(2f_{hb})$, where f_{hb} represents the fundamental frequency of the heartbeat.

After that, two approaches are followed to calculate the PTT: one for the radar signal and one for the reference signal.

- The PTTs from the reference signals are obtained following the procedure described in [10], which define the PTT as the time interval between the R-peak from the ECG and the point where the pulse pressure reaches the fifty percent of its peak.

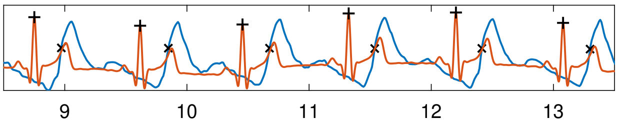


(a) Reference signals: finger pressure -in blue- and ECG -in red-.
(b) Carotid location.
(c) Heart location.
(d) Femoral location.

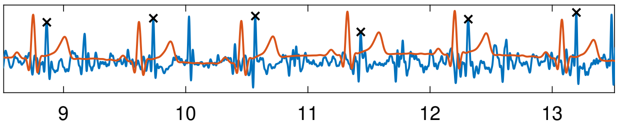
Fig. 3. Heartbeat waveform extracted with the radar from different locations compared with the ECG reference.

- The PTTs from the radar signals are obtained as the difference between the R-peak from the ECG and the maximum of the acceleration signal of each pulse, which was shown in [3] that was related with the heart sound S1.

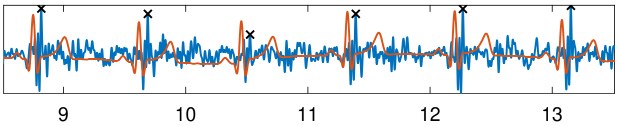
The ECG R-peaks instants are identified using the Pan and Tompkins algorithm [11]. Once the ECG and the radar points are identified, the PTT is calculated as the time difference between them. It is important to highlight that a double differentiation is a noisy process, thus, the noise present is magnified, as stated in [3]. Despite that, since it is a short-distance application where the subject remains quite still, the heart information is not masked by noise or motion artifacts. This fact can be observed in the scalograms depicted in Fig. 5, where the heartbeat extracted at each location is analyzed using time-frequency analysis. It shows that heart information activity can be retrieved successfully from the three locations. Moreover, the frequency components associated with the



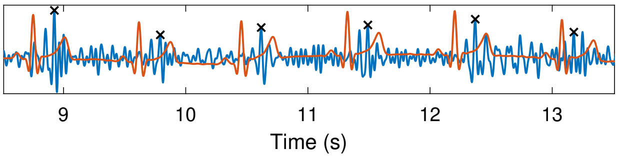
(a) Reference signals: finger pressure -in blue- and ECG -in red-.



(b) Carotid location.



(c) Heart location.



(d) Femoral location.

Fig. 4. Temporal instants obtained from the acceleration signals acquired with the radar (after double differentiation) compared with the ECG reference.

temporal instants identified in Fig. 4 can be distinguished from the surrounding noise.

It is important to note that using the double differentiation approach to identify time instants yields better results in extracting cardiac temporal information, as demonstrated in [3]. Since heartbeat waveform changes between measuring locations, as can be seen in Fig. 3, and can be distorted by breathing harmonics or motion artifacts, which often lie within the same frequency range as the heartbeat.

IV. Results

This section presents the results obtained with the radar setup described in Fig. 6. The three radars are pointing to the region of interest (carotid, heart or femoral) at a distance of 0.3 m. The radars are configured to transmit with a sweep time of 1 ms and a bandwidth of 6 GHz centered in 121 GHz. The person-under-test is a healthy 27 years male, which has been monitored four times in intervals of 60 s.

The acceleration of the different heartbeat signals obtained from each location are presented in Fig. 4, where it is noticeable the delay in the maxima of each signal with respect to the R-peak. This is due to the path that each heartbeat travels from the heart through the arteries.

The PTTs obtained are displayed in Fig. 7. This figure shows that the PTT measurements are consistent for each location, thus, it proves that the pulse delay can be calculated using radar techniques. It also shows how the delay is greater in the locations that are further from the heart location, as was expected. With these data it is possible to calculate the PWV, estimating the pulse travel path, as indicated in Fig. 6. Thus, knowing the delay between the heartbeat measured in the carotid, the one measured in the femoral, and the distance between these two points, the carotid-femoral PWT (cfPWV) is calculated and represented in Fig. 8. The cfPWV has a mean value of 6.75 m/s and a standard deviation of 1.07 m/s. The

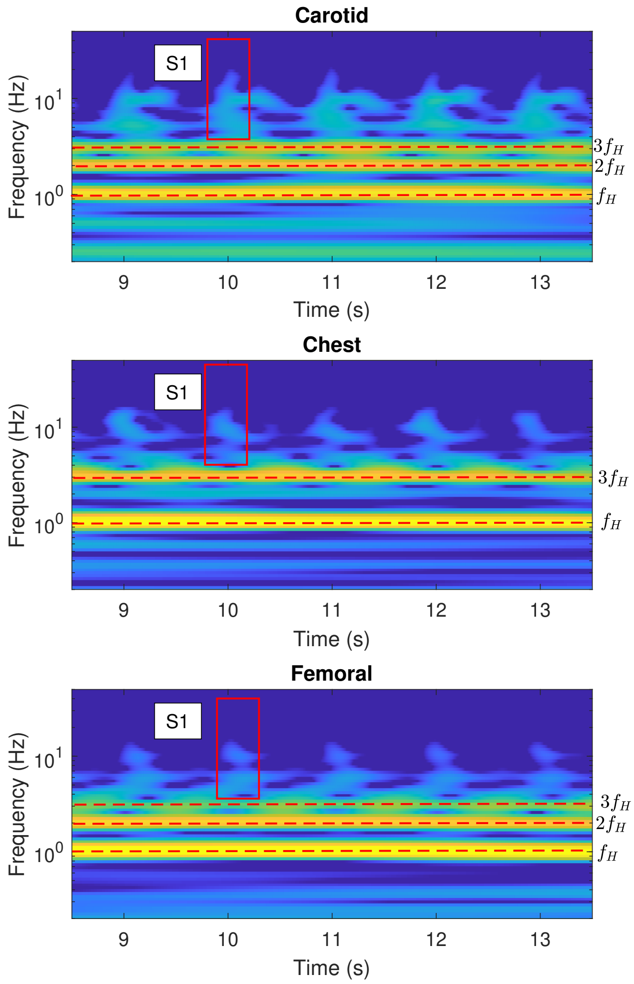


Fig. 5. Time-frequency analysis of the heartbeat waveform extracted with the radar from each location. The fundamental heartbeat frequency and its harmonics are identified with a red dashed line. Fundamental heartbeat frequency and harmonics (red dashed line) and first heart sound frequency complex (red square) are identified.

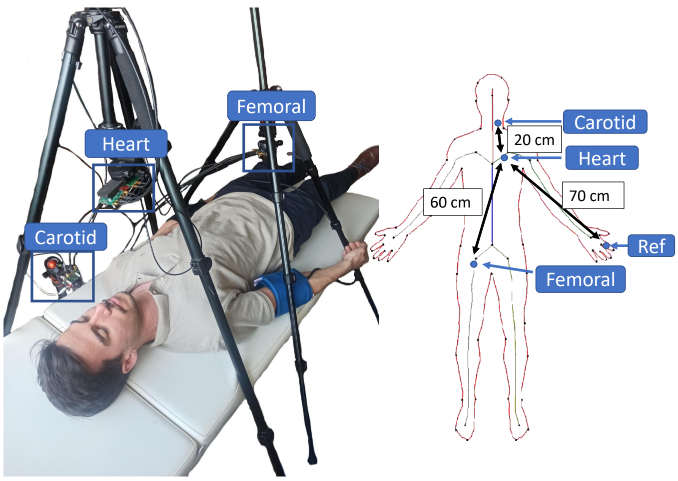


Fig. 6. Experimental radar setup. Body diagram with the radar pointing locations and the estimated path distance between the different measuring points. The distances have been measured using a tape measure on the surface of the body.

mean value ($\pm 2\ SD$) for the PWV in a healthy male under 30 years old, as defined in the literature, is 6.6 (4.9-8.2) m/s [12]. It is important to clarify that these measurements are based on a gross estimation of blood vessel length, so further analysis is required with medical experts.

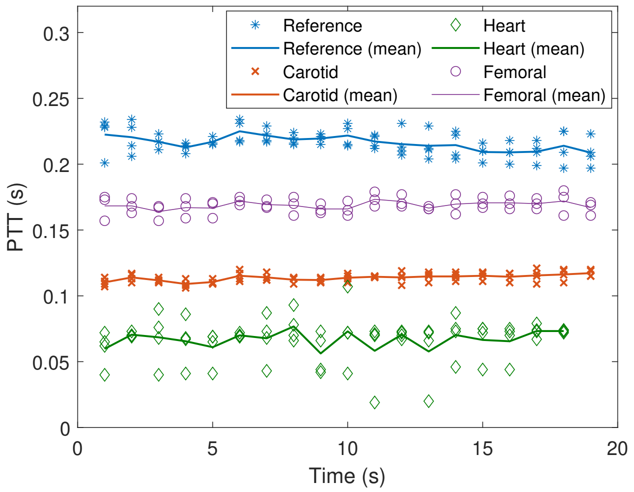


Fig. 7. PTT measurements calculated for each location: finger reference -in blue-, carotid -in red-, heart -in green- and femoral -in purple-. Solid lines represent the mean of the four measurements carried out.

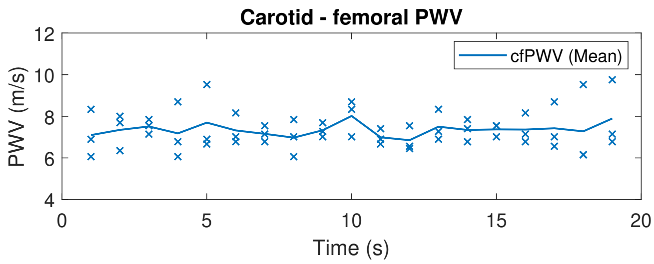


Fig. 8. Results of cfPWV measurement obtained with the radar setup measuring an under 30 years old healthy male, with a mean of 6.75 m/s and a standard deviation of 1.06 m/s.

V. Conclusion

A 3-node mmWave radar network for monitoring PTTs and PWV is presented. Results show that the system can measure the PTT at different locations and extract the cfPWV. This allows the implementation on monitoring scenarios, such as hospitals, where the patient is lying practically still on a stretcher. However, this is a first step that brings us closer to the possibility of monitoring these parameters at home, for instance, placing a radar network under the mattress of a bed. Being able to measure breathing and heart rates, HRV and PWV of the person monitored and thus being able to detect early possible diseases.

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