Multi-Point Near-Field RF Sensing of Blood Pressures and Heartbeat Dynamics

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ABSTRACT Systolic and diastolic blood pressure estimation using arm-cuff monitors is one of the most common cardiovascular evaluation criteria in healthcare today, however these measures lack critical heartbeat and pressure dynamics. Pulse-transit time can be used as an alternative for arm-cuff monitors, but gives only one parameter for two pressure quantities. Ultrasound, computed tomography scan, and magnetic resonance imaging can retrieve geometrical features of the heart, but cannot directly estimate vascular pressures. Here we show a novel radio-frequency heartbeat sensor based on multi-point near-field observation. By comparing to synchronized electrocardiogram and auscultation, the multi-point sensor can assess motion and pressure in different parts of the heart following the Wiggers diagram. By applying the Hilbert-Huang frequency-time transform, the central blood pressure can be derived from the vascular vibration characteristics as continuous transients, including during the pulmonary cycle which is previously inaccessible from branchial measurements. Our scheme can be further extended to a full multiple-input-multiple-output (MIMO) channel arrangement, producing rich content for diagnostic and biometric applications. We employ dynamic time warping analyses to illustrate the sensor’s wealth of diversified information across different channels and persons. This new multi-point sensor, which can be worn conveniently over clothing, enables unprecedented monitoring capabilities of detailed central blood pressure transients and heartbeat dynamics under the given measurement instructions for at-home and clinical cardiovascular diagnostics.

INDEX TERMS Biomedical signal processing, cardiography, microwave sensors, MIMO, stethoscope, wearable sensors.

I. INTRODUCTION
Arm-cuff sphygmomanometers for blood pressure (BP) measurements are universally employed in modern at-home and in-patient healthcare practices. Systolic and diastolic BPs are estimated by branchial vascular occlusion once within a period of many pulses. However, the intrusive cuff inflation can disturb sleep as well as interrupt circadian rhythms and vigilance. In addition, continuous heartbeat time dynamics during central aortic and atrioventricular cycles remain unavailable, and the characteristics of the pulmonary circulation are entirely missing. Monitoring central aortic and pulmonary artery pressure transients results in more accurate evaluation of cardiovascular risk factors than the branchial BPs, however these crucial metrics can previously only be measured by invasive catheters [1], [2]. To reduce disturbance to patients, BP can be indirectly derived from the pulse-transit time (PTT) [3], [4], which is defined as the transit delay between two chosen proximal and distal points. PTT can be measured by either photo-plethysmography (PPG) [5], [6] or radio-frequency (RF) [7] methods to estimate the systolic pressure with personal calibration, while the diastolic pressure is derived from a fitting ratio. PTT sensors must be placed on two disjoint body positions and require synchronization with sub-millisecond precision, making PTT a cumbersome method for long-term ambulatory monitoring. Notice that PTT derives both systolic and diastolic pressures from a single measurement parameter, which is non-ideal for pathological evaluation, even with personal calibration. Alternatively, heartbeat geometrical features can be

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recovered by ultrasound, computed tomography (CT) scan, and magnetic resonance imaging (MRI) [8], each of which often require professional operators and extensive lab setup, and are hence infeasible for long-term or at-home monitoring. BP cannot be directly measured with these methods but can be estimated with personal calibration. Ultrasound has limited spatial resolution for accurate extraction of detailed valve dynamics, although the blood velocity and systolic BP can be derived from the Doppler effect [9]. CT scans and MRI have limited time resolution, and the heart valve timing and pressure dynamics are subject to large uncertainties.

Electromagnetic heartbeat sensing has been investigated for many years as well [10], [11]. Microwave radar-based approaches, by either the Doppler effect or the frequency-modulation chirps [12], [13], require the line-of-sight (LoS) alignment to the chest area by an off-body reader, although no wearable body tag is needed. As the minute chest movement caused by heartbeats is read at a distance, the signal is often weak and the spatial resolution of the different heart areas is low. Alternatively, notched or resonator-loaded transmission lines can be employed to sense the local dielectric properties [14], [15], which had been extended to vital-sign monitoring [7]. However, the blood pressure estimation has been restricted to PTT, and no multi-point observation on the heart had been attempted.

In this paper, we present a new RF-based method for tracking heart dynamics and central blood pressure based on the near-field coherent sensing (NCS) principle [4]. This method enables convenient and comfortable cardiac monitoring by placing multiple sensing antennas over clothing on top of the heart, allowing for heartbeat dynamics to be directly modulated onto the received RF signal. Due to the high sensitivity and broad-band coupling of NCS, both large cardiovascular motion and minute cardiac vibration characteristics can be retrieved. When these cardiac vibrations are processed using the Hilbert-Huang frequency-time transform, beat-to-beat blood pressure transients can be derived by matching the frequency-time signal to the Wiggers diagram [16]. The multi-point NCS method enables position-specific sensing, in which multiple antennas provide a wealth of diversified information to analyze complex heartbeat motion dynamics, including both systemic and pulmonary circulation when the subjects sit still and control their breathing rhythm. The new multi-point NCS system in this paper is the first long-term wearable technology that can provide such detailed central blood pressure transients and heartbeat dynamics, and can offer unparalleled monitoring capabilities under the given measurement condition for clinical and at-home cardiovascular diagnostics.

**II. THE MULTI-POINT NCS SYSTEM SETUP**

The NCS principle states that when a dielectric boundary moves in the near-field region of a transmitting or backscattering antenna, its motion becomes part of the antenna characteristics that can be evaluated after carrier demodulation and baseband filtering [4], [17]. Both carrier and baseband injection in NCS contributes to channel isolation and eventual multiplexing, and the near-field coupling provides the sensing locality. Dielectric boundary motion relative to the antenna will be mainly captured by the received NCS magnitude, from the interferometer-like structure as a differential mode, while boundary motion occurring together with the antenna will be captured by the NCS phase as a common mode [4]. The NCS system couples significant RF energy in the ultra-high frequency (UHF) band (300 MHz – 3 GHz) into the entire heart region, allowing for the complex dielectric boundary motion of the heart to be directly modulated onto our received signal magnitude and phase. The radiation level is very low and well under the safety standard prescribed by OSHA (Occupational Safety and Health Administration) [18]. To study the complex heartbeat dynamics from different angles, multiple observation points are helpful, as the received signal at each point is dominated by the local dielectric boundary motion.

The new multi-point NCS system is shown in Fig. 1(a), which consisted of two software defined radios (SDR, National Instrument Ettus Research B210), denoted as SDR1 and SDR2. The two SDRs are synchronized by an external local oscillator (LO) with 10 MHz reference and 1 PPS (pulse per second) baseband synchronization (BG7TBL-GPSDO). The SDRs were connected to a host computer through US b cables. The control software was implemented in LabVIEW.

Each port of the NCS system consists of one RF transmitter (Tx) and one receiver (Rx). For Port 1 (Po1) in SDR1, the first NCS sensing antenna pair is connected to Tx1 (blue triangle) and Rx1 (red triangle). The field programmable gate array (FPGA) prepares the baseband signal to be fed to the Tx1 chain, and the received RF signal modulated by the heartbeat is fed to Rx1 and demodulated by the FPGA. The other three ports (Po2 to Po4) are similarly configured with Tx2 and Rx2 to Po2 in SDR1, Tx3 and Rx3 to Po3 in SDR2, and Tx4 and Rx4 to Po4 in SDR2. The RF signal at each Tx can be multiplexed with time-division multiple access (TDMA) [19], code division multiple access (CDMA) [20], [21] or, in the case of this work, frequency division multiple access (FDMA) [17], [22]. The carrier frequency is 1.82 GHz with 0.71 MHz, 1.22 MHz, 1.71 MHz and 2.33 MHz baseband offsets for Tx1 – Tx4, respectively. The four sensing antenna pairs were placed at Points 1 – 4 in Fig. 1(b), which were chosen to mimic the conventional auscultatory stethoscope for listening to the aortic, mitral, pulmonary and tricuspid valves. Point 0 was chosen as Erb’s point, a central area of the chest used as a reference NCS position for timing comparison. The electrodes of the electrocardiogram (ECG, NeuroSky BMD101) were pasted on the torso at Points LL (left leg) and RA (right arm) and connected to the host computer by USB. The ECG signal was used as a timing reference to highlight the feature points in the cardiac cycle. A digital stethoscope (Thinklabs One) was placed at Point ST to record the S1 and S2 heart sounds. The SDR sampling rates of the data converter were set as 5 MSps, which is above the Nyquist rate of 2 times of the highest
FIGURE 1. The multi-point RF system for near-field coherent sensing (NCS). (a) Electronic hardware: The cardio reader constructed by two SDRs and an external reference clock source for synchronization. (b) Body placement: The location of the sensing points: Points 0 to 4 are for NCS, ST is the stethoscope drum placement, and the ECG electrodes are pasted at LL and RA. The letters “L” and “R” in the deltoid area indicate body left and right. FPGA denotes the field-programmable gate array; Tx the transmitter; Rx the receiver.

baseband offset at 2.33 MHz. After channel demodulation, the NCS signal was down-sampled to 50 kSps, which is slightly higher than 44.1 kSps of the digital stethoscope so that NCS can contain all information from the audible signal. The NCS, ECG, and stethoscope signals were synchronized in LabVIEW. A detailed system schematic is shown in Supplementary Fig. 1.

The picture of the fully coherent system is shown in Fig. 2(a). In Figs. 2(b) and (c), two kinds of NCS antenna pairs are shown. In Fig. 2(b), two monopole antennas (black whips, Taoglas TG.19.0112) are mounted on a 3D-printed holder. The overall dimensions are 69 × 17 × 11 mm (W × L × D). The operating frequency is around 1.82 GHz with a bandwidth of 40 MHz. A smaller NCS sensing antenna pair by two ceramic patch antennas (Taoglas WLP.12C) is shown in Fig. 2(c). The overall dimension is reduced to 32 × 14 × 5 mm. The operating frequencies are around 900 MHz and 5 GHz. For the commercial antennas used here, the operational bands are chosen by the reflection coefficient $S_{11}$ when the NCS antenna pair is placed on the human chest area [23]. In general, the lower frequency provides better penetration into the body and stronger signal coupling. The carrier frequency selection should also comply with the government regulation in the future when custom antennas are deployed.

To extend the system to full multiple-input-multiple-output (MIMO) capability, we configured the dual SDR system to implement cross-point channels. This enabled, for example, Tx1 to be received by Rx1, Rx2, Rx3 and Rx4, for greater signal diversity and measurement opportunities. The ECG electrodes and the stethoscope drum require direct skin contact to acquire clear signal, yet the NCS sensing antennas can be placed outside the clothing as shown in Fig. 2(d) due to the effective RF penetration in the UHF band [4], [22]. All NCS, ECG and stethoscope recordings were synchronized in the host computer. An illustration of the MIMO NCS coverage is shown in Supplementary Fig. 2.

III. SYNCHRONIZED NCS, ECG AND STETHOSCOPE

The timing relation among the synchronized ECG, stethoscope and single-channel NCS signals is shown in Fig. 3(a), where the NCS antenna pair was put on Point 0 (Erb’s point) in Fig. 1(b). The demodulated NCS raw recording is shown as the green curve in Fig. 3(a), and its zero-phase bandpass filtered (7 – 20 Hz) waveform is the purple curve to highlight the high-frequency vibration components. The synchronized ECG and stethoscope waveforms are shown as the blue and pink curves. The first heart sound (S1) happens around the R-S period of the ECG caused by the closure turbulence or claps of the atrioventricular (AV) valves, and the second heart sound (S2) is at the end of the T wave generated by that of the semilunar (SL) valves. These timing features will be aligned according to the physiological Wiggers diagram [1], [16] in later analyses.

As the NCS signal is modulated by the mechanical motion of the dielectric boundaries in its proximity, the received raw waveform is dominated by the strong fundamental heartbeat tone. In addition to this tone, the details of the internal motion of the atria, ventricles, valves and artery are also coupled to the NCS signal and each motion has its frequency features and timing. Because all physiological frequencies are much lower than the impinging RF carrier, they will all be captured in the demodulated NCS signal, constrained only by the signal-to-noise ratio (SNR). Similar to audible sound analysis, higher frequency components will have lower amplitude for the same spectral energy. The high-frequency information of the heartbeat is thus visually overwhelmed by the low-frequency component in the time domain. After proper bandpass
filtering is applied to the raw NCS signal, the high-frequency features can be more clearly viewed in the systole and diastole phases. Notice that some NCS features are not included in either ECG or stethoscope. ECG measures the electrical activity started from the sinoatrial node, which does not carry the direct information of the mechanical motion. Stethoscope, on the other hand, can only capture vibration in the audible range as determined by the acoustic impedance matching of the drum, and will likely attenuate low-frequency parts of the auditory signal. From Fig. 3(a), we can clearly observe that no clear stethoscope signal is present between S1 and S2, but the bandpass filtered NCS signal still contains many features, which are likely from atrial and ventricular chamber movement with spectral features lower than the audible frequencies in stethoscopes. For a direct comparison, the NCS signal after bandpass filtering of 16 – 120 Hz is compared to the synchronized waveforms in the acoustic stethoscope [23], as illustrated in the Supplementary Fig. 3 and Supplementary Movie 1.

We will now use the four-point NCS system shown as Points 1 – 4 in Fig. 1(b). In Fig. 3(b), the backscattering channels of Tx1-to-Rx1 ($C_{11}$), Tx2-to-Rx2 ($C_{22}$), Tx3-to-Rx3 ($C_{33}$) and Tx4-to-Rx4 ($C_{44}$) after the same zero-phase bandpass filtering (7 – 20 Hz) are presented as red, yellow, green and pink curves, respectively. The synchronized ECG in the blue curve is added for timing reference. The system setups of $C_{11}$ to $C_{44}$ are the same, but each antenna pair demonstrates its unique weighted coupling to the different parts of the heartbeat. $C_{11}$ and $C_{33}$ have stronger signals just...
FIGURE 4. BP analyses based on the NCS backscattering channel during the systole phase. (a) The cross section of the heart ejecting blood from left and right ventricles. (b) The Wiggers diagram shows the expected timing relation between ECG (blue) and the aortic pressure (red). (c) The HHT frequency-time analysis of \( C_{11} \) when the person sat on the floor. The synchronized ECG signal (blue) indicates the timing. (d) \( C_{11} \) HHT when the person took a standing posture. (e) \( C_{33} \) HHT for the pulmonary cycle when the person held breath after maximum exhalation. (f) \( C_{33} \) HHT when the person held breath after maximum inhalation.

After the QRS complex when the heart ejects the blood from the ventricles to the artery through the SL valves. Analogous to the auscultation position, \( C_{11} \) will have more emphasis on the aortic valve and artery, while \( C_{33} \) will emphasize the pulmonary valve and artery. Similarly, \( C_{22} \) contains more mitral valve motion during the P-R period, and \( C_{44} \) contains more tricuspid valve information although it also couples in the strong left ventricular motion during the T wave. The aortic circulation usually has larger motion and higher pressure than the pulmonary circulation due to the higher vascular resistance. This cannot be directly observed from Fig. 3, as the NCS magnitude is normalized to its respective highest peak during the recording to visualize each channel more clearly.

**IV. BLOOD PRESSURE EXTRACTION**

During the systole phase, the \( C_{11} \) waveform is dominated by the aortic motion and can be used for BP derivation through the vessel vibration characteristics. Notice that \( C_{11} \) is related to the central BP of the aortic artery instead of the branchial BP given by the arm cuff, although the two values are often reasonably correlated when the cuff is at the same height of the heart [1]. At the end of the diastole phase, left and right ventricles are at their largest volume filled with blood. Then the ventricular contraction is triggered by the QRS signals into the systole phase. Fig. 4(a) shows the cross section of the heart ejecting blood from both ventricles into the aorta and pulmonary arteries. Fig. 4(b) shows the corresponding part of the Wiggers diagram [16] for the timing relation between ECG (blue) and the aortic pressure (red). The systole period is between the black dashed lines, where the tricuspid and the mitral valves are closed to inhibit regurgitation and thus the atria have no significant motion during systole. When the BP in the ventricles are higher than those in aorta and pulmonary arteries, the aortic and pulmonary valves will open, which is indicated by the magenta arrow in Fig. 4(b). The ventricular contraction makes the aortic pressure reach its peak value, which is defined as the central systolic BP indicated by the orange arrow around the T wave. Afterwards the aortic pressure begins to drop due to blood flow to branchial vessels until the aortic and left ventricle pressures are equal, then the aortic valve closes to form a small cusp in the aortic pressure generating the S2 heart sound, as indicated by the cyan arrow. The aortic pressure keeps dropping in the next diastole phase when the AV valves are open for blood to flow from atria to ventricles. Immediately after the QRS complex, the aortic pressure reaches its minimum value, which is defined as the central diastolic BP indicated by the green arrow, where the AV valves close to make the S1 heart sound. As the fluid pressure is positively correlated to the vibration characteristics of the containing viscoelastic vessel and valve [9], [24], the high-frequency...
components in $C_{11}$ during systole would contain the aorta vibration features, which can be used to estimate the central systolic and diastolic BP.

Therefore, NCS BP derivation requires high resolution in both frequency and time with a broad base bandwidth. Frequency and time are the Fourier transform pair, and hence frequency and time resolutions are limited by the Uncertainty Principle model [25], [26]. A simplest way to observe this limitation is by the short-time Fourier transform (STFT) [27] where a time window is applied to compute the spectrum of an infinite time series. To obtain high time resolution in STFT, we need to reduce the time window length, which causes the spectrum within the time window to spread out, and thus high frequency resolution cannot be simultaneously achieved by Fourier transform. Other methods such as wavelet transform and windowed Fourier transform [28] can mitigate this deficiency to some extent, but the tradeoff between the time and frequency resolutions remains.

To minimize the frequency-time resolution contradiction, we opted to use the Hilbert-Huang transform (HHT) to obtain the frequency-time spectrum [29], [30], where the instantaneous frequency is calculated at each sampling time point for the nonstationary oscillation analysis. The time resolution depends on the sampling rate in the time domain, which can be easily above $10^6$ samples per second (Sps) by the analog-to-digital converter (ADC) in the chosen SDR. The high ADC sampling rate can also spread the noise over a larger spectrum to reduce the noise floor and increase SNR. This frequency resolution was chosen to facilitate reasonable computation time. $C_{11}$ represents observation from Point 1, which mimics the stethoscope position for the aortic valve murmur. Fig. 4(c) is the HHT frequency-time analysis of a single heartbeat obtained from $C_{11}$ when the person under test sat on the floor. The blue curve is the synchronized ECG signal to show timing, and the ventricular ejection period is between the dashed lines. The sampling rate for the waveform is $10^3$ Sps, producing a 1 ms time resolution. The highest frequency response is half of the sampling rate at 500 Hz. The frequency resolution is set to 0.125 Hz, 1/8000 of the sampling rate.

All post processing was performed in MATLAB. All filtering used zero-phase FIR (finite impulse response) filters to cancel the phase shift and maintain the linear phase response. The down-sampled signal was further band-pass-filtered (1.5–16 Hz) to attenuate the fundamental heartbeat tone, high-frequency heart sounds, and noises. The filter frequencies are selected based on the signal characteristics [23]. The resulting waveform $x(t)$ was then processed by empirical mode decomposition (EMD) [29] as a part of the HHT algorithm, where several intrinsic mode functions (IMF) $c_j(t)$ represent the simple nonstationary oscillatory modes and can be extracted as:

$$x(t) = \sum_{j=1}^{n} c_j(t) + r_n$$

where the final residue $r_n$ is either the mean trend or a constant. We extracted the first seven IMF $c_1(t)$ to $c_7(t)$ for $x(t) = C_{11}$ in Supplementary Fig. 4, where we can see most features are captured in $c_1(t)$ and $c_2(t)$. The signal is further processed by the Hilbert spectral analysis (HSA).

$$y_j(t) = H[c_j(t)] = \frac{1}{\pi} PV \int_{-\infty}^{\infty} \frac{c_j(\tau)}{t - \tau} d\tau$$

where $H[\cdot]$ and $PV$ represent the Hilbert transform and the Cauchy principal value. Then the analytic signal $z_j(t)$, which comprises the original signal and its Hilbert transform, can be defined as

$$z_j(t) = c_j(t) + iy_j(t) = a_j(t) \exp[i\theta(t)]$$

where $i$ is the imaginary unit, $a(t)$ is the instantaneous amplitude, and $\theta(t)$ is the phase function. The analytic signal $z_j(t)$ is a complex-valued function that has no negative frequency component, and can be considered as a generalized time-varying phasor. The instantaneous frequency can then be extracted as

$$\omega_j = \frac{d\theta_j}{dt}$$

We only kept the first two IMFs with the highest magnitude here. The IMF distribution is shown in Supplementary Fig. 4 for illustration. This procedure can be considered as an additional filtering which is not based on the pass-stop band, but by the oscillatory modes. The real part ($\Re\{\cdot\}$) of the processed signal can be expressed as

$$x'(t) = \Re\{\sum_{j=1}^{2} a_j(t) \exp[i \int \omega_j(t) dt]\}$$

In Fig. 4(c), each point represents the instantaneous frequency $\omega_1$ at the corresponding time, and its color shows the instantaneous amplitude in the dB scale normalized by the respective maximum value. The horizontal and vertical axes are the time and frequency, respectively. During the systole phase, the frequency-time points of $\omega_1$ form a red-pink “curve” with a similar shape of the aortic pressure curve in the Wiggers diagram. The maximum frequency point at 9.427 Hz, denoted as the systolic pressure frequency (SPF), is well aligned with the systolic BP point. The minimum at 4.676 Hz, denoted as the diastolic pressure frequency (DPF), is aligned with the diastolic BP point at the beginning the systole phase. The concurrent brachial BPs measured by an arm-cuff monitor (Omron BP760N) are 131 mmHg and 81 mmHg. We then repeated the experiment when the person stood up. The frequency-time analysis is shown in Fig. 4(d). The SPF and DPF in this case are 10.88 Hz and 5.126 Hz, and the corresponding arm-cuff BPs are 143 mmHg and 94 mmHg. Two-point linear interpolation can transform the two SPF to systolic BP, and similarly the two DPFs to diastolic BP in Figs. 4(c)(d). This interpolation scheme enable the NCS system to estimate continuous blood pressure transients. The frequency resolution is 0.125 Hz, corresponding to 1.0 mmHg resolution for the systolic BP and 3.6 mmHg...
for the diastolic BP. The frequency resolution can be further improved by adjusting the HHT parameters at the cost of computational time. A measurement from the second person is shown in Supplementary Fig. 5. Due to the chosen HHT calculation, spurious traces of $\omega_2$ with lower frequency were also observed in Figs. 4(c)(d), which can be observed as the faint yellow curves. This ambiguity can be eliminated with the narrower choice of HHT bandwidth or can be simply ignored from its lower magnitude as usually the first IMF $c_1(t)$ is much stronger than the second IMF $c_2(t)$. Notice that the aorta region is the main nonstationary oscillation in $C_{11}$ HHT only for the systole period. During the diastole phase, there can be other vibration in the vicinity of Point 1 such as the right atrium. However, we do not have a direct method to confirm the physiological meaning of diastole $C_{11}$ at the present time. Analogous to the stethoscope recording, the heart sound S2 originated from the aorta region can be listened from Point 1 over the right atrium which does not have significant vibration during the systole phase. The collected heart sound by SL valve closure was mostly in the frequency range of 16 - 120 Hz, which was not included in our HHT.

After calibration, we tested the extracted systemic BP in a longer period. The measurements in Fig. 5(a) were taken when the person sat in a chair. The pink and cyan curves are the systolic and diastolic BP from 300 continuous heartbeats monitored by $C_{11}$ HHT. The dashed curves are the BP moving average with the window of 10 heartbeats. The orange stars are the measurements from the arm-cuff BP monitor. The measurements in Fig. 5(b) were taken after the person did some mild exercises and stood up, where the BP relaxation was traced accurately.

$C_{33}$ at Point 3 in Fig. 1(b) corresponds to the auscultatory position to acquire the pulmonary valve murmurs. Similar to the aortic circulation, the oscillatory characteristics of the pulmonary artery are related to the right ventricular and pulmonary BP [9]. As the pulmonary BP is more affected by the lung volume due to transpulmonary pressure instead of the posture [31]–[33], Fig. 4(e) shows the $C_{33}$ HHT analysis with the synchronized ECG when the person held breath after maximal exhalation. The SPF (orange arrow) and DPF (green arrow) were extracted at 7.752 Hz and 1.875 Hz. We then took the measurement in Fig. 4(f) when the person held breath after maximal inhalation with SPF and DPF now at 11.38 Hz and 4.376 Hz. Both SPF and DPF increased due to the higher transpulmonary pressure. In Figs. 4(e)(f), only one IMF curve was observed. Due to a lack of access to cardiac catheterization, the SPF and DPF frequencies cannot be quantitatively mapped to BP, but are independently useful as they provide a non-invasive method of monitoring pulmonary artery physiology, enabling improved cardiac diagnostics and overall cardiopulmonary health analysis. We analyzed 20 heartbeats for each of the two scenarios and repeated the measurements three times, where the calculated SPF (pink) and DPF (cyan) for the pulmonary cycle are shown in Fig. 5(c). In comparison with the PTT scheme which assumes a fixed ratio between the systolic and diastolic pressures, the calibration method in this work utilizes individual SPF and DPF to interpret the two pressures independently to give a more accurate model.

V. MULTI-STATIC MIMO NCS

The NCS system can observe heartbeat dynamics not only from each sensing antenna pair with collocated Tx and Rx by backscattering, but also from the multi-static MIMO channels. Four cross-point signals are presented in Fig. 6(a). The purple waveform ($C_{21}$) is the signal transmitted from Tx1 and received by Rx2. The naming order of Tx-Rx ports follows the convention of the multi-port scattering matrix in microwave component testing [34]. The pink ($C_{12}$), light blue ($C_{13}$) and green ($C_{34}$) curves are the signals from Tx2 to Rx1, Tx3 to Rx4 and Tx4 to Rx3, respectively. The ECG waveform (blue) remains as the timing reference. The real-time recording is shown in Supplementary Movie 2 with descriptions in Supplementary Fig. 6. Because the collocated...
backscattering channels are more sensitive to the respective local areas, the purpose of the cross-point observation is to expand the sensing path so that extra content of the heartbeat motion can be collected. Notice that this observation mode is not applicable in the passive stethoscope, but is an option for ultrasound imaging. The MIMO observation is also difficult for the conventional RF systems [12], [13], because of the signal reflection on the body surface and low spatial resolution. Each multi-static channel records clear cyclic information. The dynamic-time warping (DTW) [35] algorithm is employed to search for the optimal match among waveforms in each cycle. The Euclidean distance $a_{m,n}$ between a pair of points in the template $\Gamma$ and the heartbeat $h$ signals is defined as:

$$a_{m,n} = \frac{|\Gamma_m - h_n|}{e + f}$$

(6)

where $e$ is the length of the discretized $\Gamma$ and $f$ is the length of the discretized $h$. The indices $m$ and $n$ are the time stamps in the waveforms of $\Gamma$ and $h$. The distance $\sigma$ in the warp path $P$ can be calculated as:

$$\sigma(P) = \sum_{k,l} \sigma(p_k, p_l)$$

(7)

For a legitimate warp path $P$, the monotonically increasing $k$ must match with one or more monotonically increasing $l$, and vice versa. The first and last indices of $k$ must match with the first and last indices of $l$, respectively. The minimal $\sigma(P)$ from all legitimate paths of $P$ is then defined as the DTW distance, which measures the similarity between the two waveforms. It is noteworthy that the DTW calculation can be performed efficiently by dynamic programming [36], and is often implemented as an on-line method.

With the warping adjustment that gives the minimal $\sigma(P)$, we overlaid the signals of each heartbeat and calculate the averaged waveform as the template of the specific channel. A box-whisker plot displaying the DTW distances of the cross-point waveforms to the corresponding template is shown in Fig. 6(b), and the overlaid waveforms from the 4 channels are shown in Figs. 6(c) to (f). During data collection, the person under test remained as still as possible and controlled their breathing. DTW analysis was performed on recordings during maximum inhalation breath hold. The algorithm first extracted beat segments by identifying peaks and peak-to-peak intervals. Then the averaged peak-to-peak interval was set as the segment length which is around 703 sampling points under the sampling rate of 1 kSps. We see from the DTW analyses that the recorded waveforms are repeatable on the waveform details after periodicity adjustment, and DTW distances stay relatively low for these cross-point signals. The multi-static waveform results from
modulation of the dielectric boundary motion along the cross-
point path, which has a nonlinear coupling with stronger
weighting towards Tx according to the near-field approxima-
tion [37]. \( C_{21} \) and \( C_{12} \) monitor the heart region in a longer
path and are further away from the strong left ventricle and
aortic valve than \( C_{43} \) and \( C_{34} \). Hence, \( C_{21} \) and \( C_{12} \) have their
largest component during the atrial systole phase around the
P-R interval, while \( C_{43} \) and \( C_{34} \) have larger signals during the
isovolumetric contraction around the R-S interval. Cross-
point waveforms can be further used to extract the timing and
magnitude of detailed heart dynamics by least-square fitting
in the future. Similar analyses are conducted with another
individual, and the result is shown in Supplementary Fig. 7.

VI. DISCUSSION
The described multi-point NCS system provides a conve-
nient method of acquiring rich cardiac dynamics, including
estimation of the central BP. In comparison with the PTT
method, which derives both systolic and diastolic BP from
one variable, the NCS method can provide SPF and DPF to
derive the systolic and diastolic BP independently. Unlike a
traditional arm-cuff monitor, the multi-point NCS method
enables SPF and DPF to be calculated on each heartbeat.
This allows for beat-to-beat BP changes to be estimated; a
metric that is unattainable with an arm-cuff. Other approaches
analyze the shape of the time-domain waveforms by optical
or ultrasound methods to indirectly interpret BP [38], [39],
but are often inconsistent due to operation variations. Another
advantage of the NCS system is the continuous trace of
pressure dynamics instead of just two BP points. Although
our present estimation of systemic BP is only derived from
\( C_{11} \) HHT during the systole phase, multiple channels can
provide more reliable parameter extraction for continuous
dynamics as well as more features of the heartbeat in the
future. As the S1 and S2 heart sounds can be recovered from
NCS [23] to provide timing reference, the synchronized ECG
is not essential. However, similar to other indirect BP meth-
ods, NCS presently needs personal calibration. BP variation
between adjacent heartbeats may result from measurement
inaccuracy or may have physiological origins similar to heart
rate variation. This will need to be investigated in the future.
The sensing antennas also need to be carefully deployed at
the corresponding positions. The signal should be recorded
according to the given instructions in this paper in order to
achieve the high-quality data needed for further HHT and
DTW analyses.

Prior to this work, critical pulmonary circulation dynamics
have few viable measurement methods. The gold standard
is cardiac catheterization [40], [41], which requires signifi-
cant setup and highly trained professionals as well as posing
considerable risks. Another method is Doppler echocardiog-
raphy to derive the pulmonary blood pressure through the
velocity of the blood flow [4], [42], similar to our derivation.
However, the ultrasound system size is significant, and the
broad-spectrum acoustic impedance matching often demands
gel application, disabling wearable fitting or long-term usage.

The lower sampling rate in Doppler imaging can also reduce
its accuracy. On the other hand, NCS can work over clothing
and take advantage of the low cost and compact size in
modern RF devices such as the smart phone. For various
application scenarios, the NCS system can be adapted to the
RFID system [4], wearable devices [4], [43], and dedicated
in-clinic wired setup [17], [22], [44]. The standard digital
filters in the above analyses can be easily realized on various
platforms without heavy computational loads. Because HHT
is an adaptive algorithm, the computation time depends on the
signal complexity and required resolution. Recent heteroge-
neous computing design [45], [46] can potentially offer on-
line responses.

The multi-point NCS system is the first RF technology
achieving rich cardiovascular features and profusion of obser-
vation channels. Conventional RF-based heartbeat sensing
systems [12], [13] are limited by low SNR, so that the detailed
heartbeat dynamics cannot be distinguished. Spatial resolu-
tion is also severely limited due to the broad beam cover-
age. Our demonstrated system has significant mechanical-RF
coupling to improve SNR, and the locality nature of NCS
enhances the spatial resolution, which provides the signal
from different parts of the heart. The information from the
MIMO NCS system provides broader application possibil-
ities, including physiological and pathological diagnostic,
bioelectric recognition, and remote health care. However,
the present hardware system and understanding of the signal
characteristics are still in the early phase. More efforts on
variation studies, signal processing and calibration methods
are still needed.

VII. CONCLUSION
In this paper we present a multi-point NCS system to inves-
tigate the detailed BP transient and heartbeat motion. Based
on the different observation channels, the systolic and dia-
stolic BP in the systemic and pulmonary circulation can be
evaluated from HHT analyses. Personal parameter extraction
is still needed, but continuous values of independent systolic
and diastolic pressures for each heartbeat can be evaluated
after calibration. The full MIMO observation from backscat-
tering and cross-channels gives rich information on the vari-
ous parts of heartbeats. Our proposed approach provides a
new diagnostic tool for not only cardiology research but also
at-home long-term monitoring.

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