Self-Powered Piezoelectric Biosensing Harvester for Intracardiac Monitoring

Noora Almarri[1], Fawaz Nahab[2], Jinke Chang[2], Dai Jiang[1], Wenhui Song[2], Andreas Demosthenous[1]
[1]Department of Electronic and Electrical Engineering, University College London, Torrington Place, London WC1E 7JE, UK
[2]Department of Surgical Biotechnology, University College London, Royal Free Hospital, London, NW3 2PS UK
e-mail: uceenal@ucl.ac.uk, a.demosthenous@ucl.ac.uk

***Abstract*— This paper presents a novel, biocompatible, in-body energy harvesting piezoelectric biosensing device coupled to sensors for the detection of cardiac allograft vasculopathy. The computational modelling, implementation and measured results of the device are presented. The system includes an ultra-low power readout circuit that operates solely from power harvested from heartbeats. With a natural frequency of 1.52 Hz, the nonlinear piezoelectric harvester generates 56 µW in *ex-vivo* testing. The device has a 4 × 6 array of fabricated MEMS sensors, which provides over 80% accuracy in detection of heart geometry changes.**

Keywords—Cardiac allograft vasculopathy, COMSOL multiphysics, energy harvesting, heart, implantable medical devices, piezoelectric, readout circuits.

I. Introduction

Cardiovascular allograft vasculopathy (CAV) is a condition that affects transplanted hearts, causing the blood vessels that supply the heart muscle to narrow and eventually block. It is unclear what causes CAV but there are potential leading causes including viral infections and the use of immunosuppressive drugs. In an advanced CAV, re-transplantation is the only viable solution; however, early diagnosis could reduce the need for re-transplantation through personalized management and prevention strategies, allowing better clinical outcome. Better understanding of potential treatments could be achieved by an intracardiac pressure sensor in a transplanted heart for an extended period. Energy harvesting from the body (e.g. in this case from the heart muscle) can increase the longevity of an implant by providing continuous power, particularly where battery replacement is unfeasible [1].

Harvesting of vibrational energy is of significant interest in the field of bioelectronics [2]. Piezoelectric harvesters have simple configurations and potentially a high conversion efficiency compared to other in-body energy harvesting approaches. A basic model of piezoelectric energy harvester (PEH) consists of a current source, I_p, in parallel with piezoelectric capacitor, C_p, and resistor, R_p, with V_p denoting the voltage generated by the PEH. However, this is not sufficient to fully simulate the behaviour of the PEH. Multidisciplinary solutions are required to combine electrical and mechanical analyses to optimize power harvesting and sensing methods within the body.

Several previous studies have investigated the possibility of using brittle Piezoelectric Lead Zirconate Titanate (PZT) ceramics to harvest energy from heartbeats, as demonstrated by various proof-of-concept designs [1], [3]. Its disadvantage is that it is not biocompatible, and it requires encapsulation that limits harvested power. The proposed design shown in Fig. 1 uses Poly(vinylidene fluoride) barium titanate (PVDF-$BaTiO_3$) nanofiber composites with a high degree of flexibility and biocompatibility to replace the toxic PZT commonly used in implantable medical devices [4].

This research was supported by doctoral studentship number 201968267 to N. Almarri by UAE Ministry of Education.

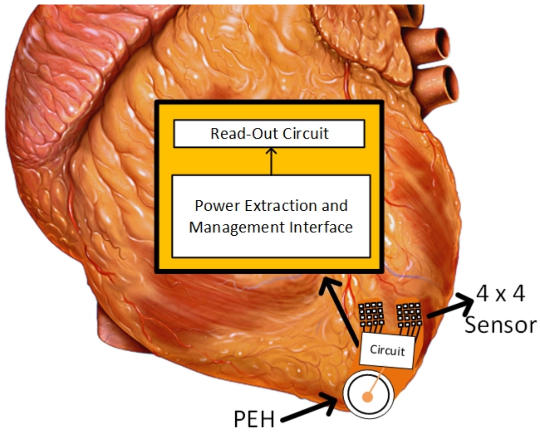

Fig. 1. Envisaged placement location of battery-less piezoelectric biosensing harvester for post heart transplanatation with piezoelectric sensors and harvester (PEH).*

This paper describes the design of a heartbeat harvester to provide power to fabricated sensors which measure the changes in the heart geometry for early detection of CAV. The sensor array size and distribution are arranged to optimize the sensitivity to changes in the environment. The implant would be placed at the lower left ventricle, as this location yields the highest power output [2]. Furthermore, the constriction of blood vessels frequently manifests in the context of smaller vasculature, thereby requiring invasive means to detect and monitor.

The rest of the paper is organised as follows. Section II explains the design principles and constraints. Section III outlines the computational modelling of the piezoelectric device with stress, voltage and power analysis. Section IV details the system architecture and readout circuit. Section V outlines the fabrication of the sensor. Section VI presents measurement results and discussion. Conclusions are drawn in Section VII.

II. Design Principles and Constraints

The proposed device addresses several challenges: 1) Dimension optimization to adjust the resonance of the piezoelectric harvester to match the heartbeat frequency of 0.6-1.80 Hz. When damping effects are small [5], the derived resonant frequency f_o of the system is

$$f_o = \frac{1}{2\pi}\sqrt{\frac{3E\cdot T^3/4L^3\cdot(1-\upsilon^2)}{m_{eff}+C_P\cdot C_e}} \quad (1)$$

where $E, T, L, \upsilon, m_{eff}, C_P, C_e$ are elastic modulus, thickness, length, Poisson's ratio of the piezoelectric material, its effective mass, material capacitance, and electrical capacitance of circuitry, respectively. T and L are adjusted to provide optimal power excitation. 2) Optimum transmission of the harvested power by matching the internal capacitance

* Heart cartoon in the background is modified from: https://coastfieldguides.com/my-medical-illustrations/

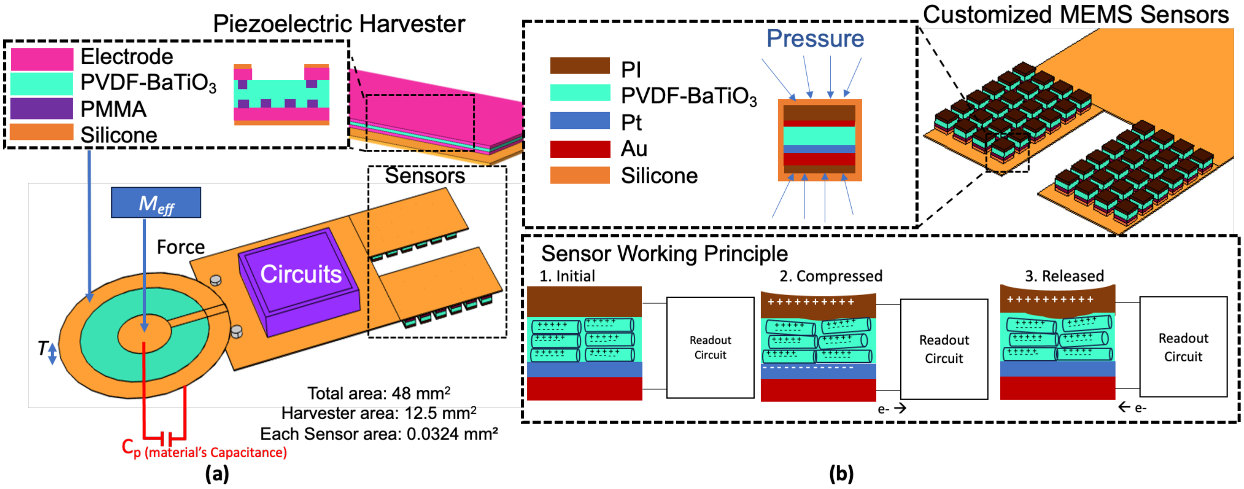

Fig. 2. (a) Proposed batteryless biosensor. **(b)** The sensor design based on MEMS sensors to detect the change of heart geometry.

of the piezoelectric harvester with the external receiver circuit [1], [6] by using ultra-low power electronics to ensure maximum power transfer at low power consumption. 3) Monitoring the geometry of the vessels using an array of PEH for early detection of CAV. For the device to be fully powered by the harvester, the readout circuit must operate at nW levels. 4) Heartbeat is an unstable low frequency power source requiring circuits that can handle the variability while providing a stable power output. 5) Miniaturization and good flexibility are crucial as the implant is placed on the surface of the heart.

III. Modelling Analysis of Piezoelectric Devices

The design was simulated with COMSOL Multiphysics 6.0 (COMSOL Inc.). Solid mechanics and pressure acoustics boundary conditions with a fine mesh of 0.002 mm were used. Placement consideration of circuits for performance optimization were taken into consideration.

When stress is applied to a PEH material it generates a voltage through a process known as 'direct effect'. The coupling relationship between the strain and the electric field can be determined as follows

$$S = S_E T + d^T \boldsymbol{E} \tag{2}$$

$$D = dT + \varepsilon_T \boldsymbol{E} \tag{3}$$

where S is the zero-to-peak strain, S_E is the material compliance, T is the stress, $\boldsymbol{E}$ is the electric field, D is the electric displacement field, d is the coupling properties, and ε_T is the permittivity. The equations of solid mechanics and electrostatics are used to analyze this problem. The equations (2) and (3) can be rearranged into the stress-charge form, which relates the material stresses to the electric field. In the frequency domain, the behavior is modelled using material properties with complex values, in the absence of the loss mechanism, as shown below

$$S = \tilde{S}_E T + \tilde{d}^T \boldsymbol{E} \tag{4}$$

$$D = \tilde{d} T + \tilde{\varepsilon}_T \boldsymbol{E} \tag{5}$$

where $\tilde{S}_E$, $\tilde{d}$, $\tilde{\varepsilon}$ are complex matrices including an imaginary part used to define the dissipative function of the material. The real and imaginary parts of the material are defined to account for the symmetrical properties of the modelled material. Mechanical damping was introduced in the system

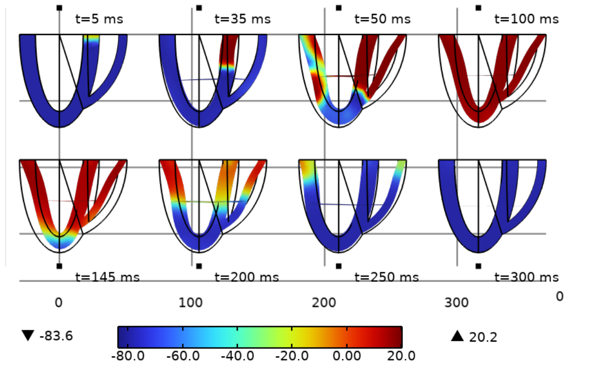

Fig. 3. Representation of the heart shape and simulated heart potential change and movement due to application of the harvester for 300 ms.

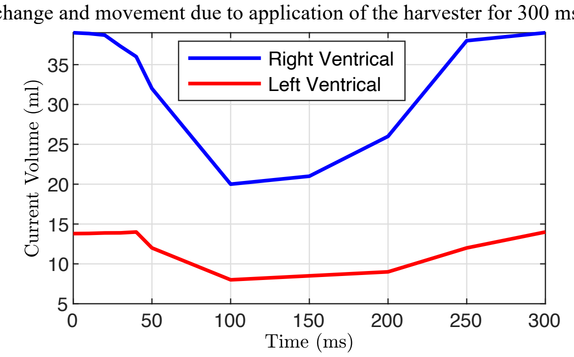

Fig. 4. Simulated active stress generated by the deformation of the cardiac walls due to application of the harvester.

by computing several eigenfrequencies, using different sets of boundary conditions. All the eigenfrequencies must be positive imaginary to represent a damped motion, while considering mechanical, coupling, and conduction losses within the system. The overall system is shown in Fig. 2. In the harvester design (Fig. 2(a)) both the triboelectric and piezoelectric effects of the material are utilized. The triboelectric effect can produce large voltage peaks, which allows for miniaturization of the device. The interdigital electrode design captures triboelectricity by allowing the center electrode to float from the material. The harvester was made with PVDF-BaTiO$_3$ with gold plated electrodes and a 0.01 mm of silicone covering layer. The use of PMMA enables piezoelectric design to operate at a low natural frequency and within an area of 12 mm × 4 mm. The harvester consists of a primary beam of PVDF-BaTiO$_3$, and secondary array beam with PMMA. Several PMMA

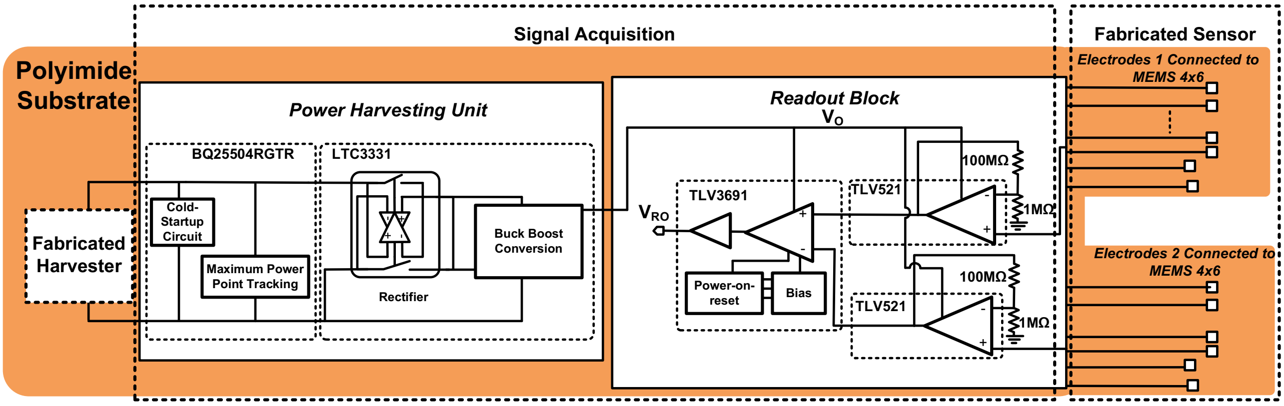

Fig. 5. Circuit block diagram of the proposed harvesting biosensor and MEMS sensors.

piezoelectric nanofibres beams are placed on the PVDF-BaTiO3 beam as shown in Fig. 2(a). The design consists of two natural frequencies (1.02 Hz and 1.52 Hz) in a cut out electrode beam, which are combined to increase the overall power produced. This model causes the stiffness to be halved while operating at conditions governed by the dynamic equations.

In the piezoelectric MEMS sensors shown in Fig. 2(b), the pressure interface at the acoustic-structure boundary is modelled in COMSOL. When the device is subjected to pressure, the generated power is dependent on piezoelectric layer thickness. The optimum power, P_{rms}, is

$$P_{rms} = \frac{\omega T (V_{amp} d_{31} s E_p)^2}{2\varepsilon_{33}^S} \tag{6}$$

where ω is the angular frequency of vibration, T is the thickness of piezoelectric, V_{amp} is the amplitude of vibration, d_{31} is the piezoelectric constant, s is the side of the piezoelectric square, E_p is Young's Modulus, and ε_{33}^S is the permittivity constant [7]. In (6) the side for which the power can be optimized is found to be s =180 μm. The electrodes connecting the sensors to the circuits were also modelled to find the resonance frequency, generated peak voltage, and the average power as functions of the piezoelectric thickness.

The placement of the device was carefully chosen to minimize its impact on the heart's function when applied to the surface. Figs. 3 and 4 demonstrate the modelled device presence does not affect the modelled heart's movement and it is within normal parameters during the cardiac cycle.

IV. System Design

To test the performance of the fabricated harvester and sensor, a power harvesting unit and readout block were placed on a flexible polyimide (PI) substrate as shown in Fig. 5. The commercial integrated circuits were selected for their small size and feasibility for validating the system.

A. *Power Harvesting Unit*

A commercial maximum power point tracking (MPPT) control circuit (BQ25504RGTR) and energy harvesting power management unit (LTC3331) were used due to their feasibility for piezoelectric harvesters. The overall power consumption is 2 μW.

B. *Readout Block*

The readout consists of two (TLV521) NanoPower operational amplifiers with power consumption of 595 nW and noise less than 300 nV/$\sqrt{\text{Hz}}$. The outputs of the two operational amplifiers are fed to a hysteresis comparator (TLV3691) to detect the heart geometry from two sensing locations. The power consumption of the comparator is 255 nW. The readout block is powered by the rectified voltage output, V_O , of the power harvesting unit. The readout interfaces to the fabricated MEMs sensors producing voltage output, V_{RO}.

V. Design Fabrication

The sensor's piezoelectric fibrious membrane PVDF-BaTiO3 was fabricated using an electrospinning technique, where deposition thickness depends on electrospinning duration, producing poling effects to enhance piezoelectricity of nm-fibres shown in Fig. 6. The size of the fabricated sensing electrode was optimized to achieve a low input referred noise (IRN)

$$IRN = 20\ \log_{10}\left(\frac{MDP}{20\ \mu Pa}\right) \tag{7}$$

where MDP is the minimal detectable pressure [5]. The design was optimized to remove pressure cancellation due to diffraction to reduce sensitivity. The IRN and sensitivity are dependent on the size of MEMS electrode, the material properties, readout circuitry, and sensor's environmental conditions. The optimized size targeted IRN was less than 25 (dB) as shown in Fig. 7. The sensitivity of the sensor was calculated through bending analysis with a natural frequency of 1.33 Hz and a displacement of 4.5 nm.

VI. Measurements and Discussion

Fig. 8 shows the *ex-vivo* measurement setup with a fabricated device on sheep's heart to mimic the physiological condition of a beating heart. The heart was perfused at 1 Hz with an aqueous-glycerol solution containing 10% glycerol, which had a viscosity of 5 centipoise (cP), using a perfusion pump. This solution was chosen due to its similarity in viscosity to blood. The pump cycle causes heart movement mimicking the natural heartbeat.

Table I. provides comparison with the state-of-the-art cardiac harvesters. Fig. 9 shows measured and modelled power output for a 1 MΩ resistive load over a range of heartbeat frequencies. The bandwidth of the harvester is widened by reducing its sensitivity to vibration in the excitation by using a nonlinear energy harvester which avoids a shift in the excitation frequency from the resonance frequency. The power levels can vary at given excitation parameters but remain in a controllable range due to the power harvesting circuit. Fig. 10 shows the measured output

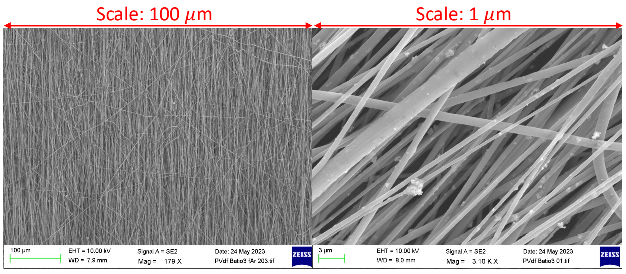

Fig. 6. Scanning Electron Microscopy (SEM) images of piezoelectric fibers.

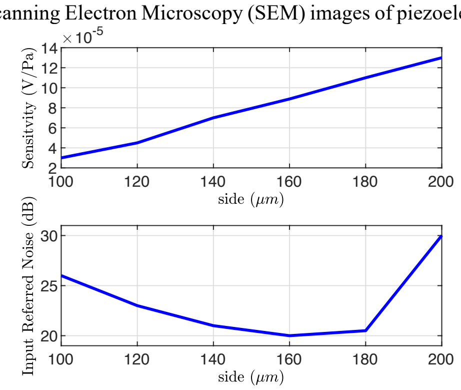

Fig. 7. Sensitivity and IRN for sizing MEMS electrode.

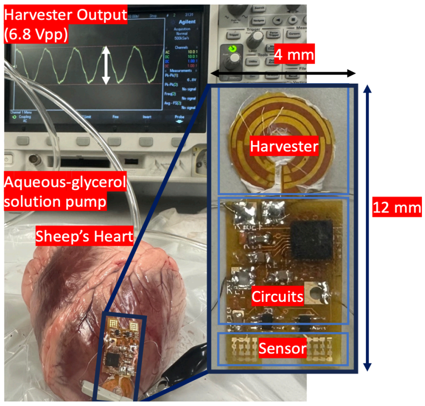

Fig. 8. *Ex-Vivo* measurement of fabricated device on sheep's heart.

voltage waveform at 1.02 Hz produced by the harvester and the rectified voltage, V_O. The readout block voltage output, V_{RO}, is a square waveform as because of pressure applied to the MEMS as shown in Fig. 11.

VII. Conclusion

A novel self-powered piezoelectric biosensing harvester has been presented for early detection of CAV for post heart transplantation patients. The fabricated device operates at heartbeat frequencies with an optimum power harvesting and sensing. The device can provide 56 µW at a heartbeat rhythm of 1.52 Hz. It is sufficient for powering MEMS sensors detecting CAV, and potentially a sub-nW communication module [9, 10].

TABLE I. Comparision of State-of-the-Art Cardiac Harvester

	[1]	[3]	[8]	***This Work***
Testing Setup	In-Vitro	In-Vivo	In-Vivo	**Ex-Vivo**
$V_{O\,(Peak)}$	4 V	1.5 V	3 V	**7 V**
R_{load}	NA	10.8 MΩ	NA	**1 MΩ**
$P_{out,Max}$	32 µW	1 µW	NA	**56 µW**
MPPT	No	No	No	**Yes**
Biocompatible	No	No	No	**Yes**

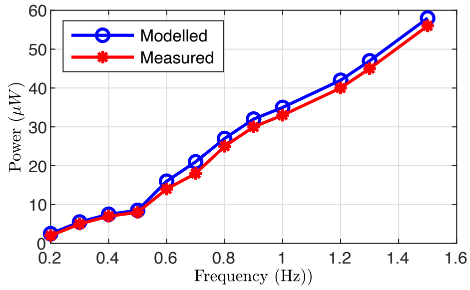

Fig. 9 Modelled and measured harvester output power with 1 MΩ load.

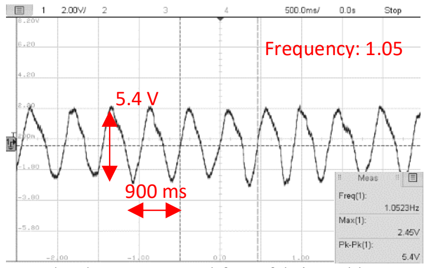

Fig. 10. Measured voltage generated from fabricated harvester with 1MΩ load.

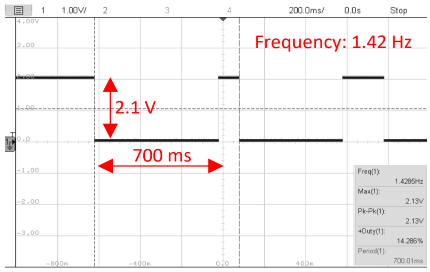

Fig. 11. Measured voltage response of harvester-powered readout circuit.

References

[1] N. Li, *et al.* "Direct powering a real cardiac pacemaker by natural energy of heartbeat," *ACS Nano*, Feb. 2019.

[2] Z. Chen, *et al.* "Piezoelectric energy-harvesting interface using split-phase flipping-capacitor rectifier with capacitor reuse for input power adaptation, " *IEEE J. Solid-State Circuits*, vol. 55, no. 8, pp. 2106-2117, Aug. 2020

[3] Lu, B., *et al.* "Ultra-flexible piezoelectric devices integrated with heart to harvest the biomechanical energy," *Nat. Sci. Rep.,* 2015.

[4] Y. Yu, *et al.* "Biocompatibility and in vivo operation of implantable mesoporous PVDF-based nanogenerators," *ACS Nano Energy*, 2016.

[5] C. Chen *et al.* "A method for quantitatively separating the piezoelectric component from the as-received 'Piezoelectric' signal," *Nat. Commun.*, vol. 13, no. 1, 2022.

[6] N. Almarri, *et al.* "In-body energy harvesting power management interface for post heart transplantation monitoring," *2022 29th IEEE Int. Conf. Electronics, Circuits Systems (ICECS)*, Glasgow, U.K., 2022.

[7] X. Ma, *et al.* "Efficient energy harvesting using piezoelectric compliant mechanisms: Theory and experiment," *J. Vib. Acoust.*, vol. 138, no. 2, 2016, Art. no. 021005.

[8] G.-T. Hwang, *et al.* "Self-Powered cardiac pacemaker enabled by flexible single crystalline PMN-PT piezoelectric energy harvester," Adv. Mater., vol. 26, no. 27, pp. 4880-4887, Jul. 2014.

[9] De la Paz, *et al.* "A self-powered ingestible wireless biosensing system for real-time in situ monitoring of gastrointestinal tract metabolites," *Nat. Commun.*, vol. 13, p. 7405, 2022.

[10] M. Schormans, *et al.* "Short-range quality-factor modulation (SQuirM) for low Power high speed inductive data transfer," *IEEE Trans. Circuits and Systems I: Regular Papers*, vol. 66, no. 9, pp. 3254-3265, Sept. 2019.