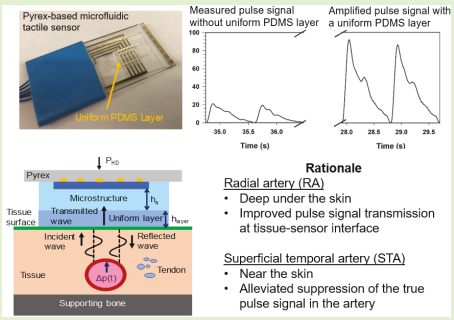
Arterial Pulse Signal Amplification by Adding a Uniform PDMS Layer to a Pyrex-Based Microfluidic Tactile Sensor

Zhili Hao and Dan Wang

***Abstract*—Various flexible tactile sensors based on micro/nano-fabrication technology have been developed to amplify a measured pulse signal for accuracy. Yet, these sensors suffer from complicated configurations and fabrication complexity. This work is aimed to investigate the feasibility of amplifying a measured pulse signal by adding a uniform polydimethylsiloxane (PDMS) layer to a Pyrex-based microfluidic tactile sensor. The amplifying mechanism of the proposed approach is revealed by theories on sensor-artery interaction. The pulse signals at the radial artery (RA) deep under the skin and the superficial temporal artery (STA) near the skin of one subject are measured by the sensor first with no uniform layer and then with a set of uniform layers with different mixing ratios of PDMS and thickness. Arterial parameters: elasticity, viscosity and radius, are estimated from the measured pulse signals. As compared to those measured with no uniform layer, a uniform layer generates a pulse signal at transmural pressure (P_T) near zero, greatly amplifies the measured pulse signal at both arteries, causes a moderate increase in estimated arterial elasticity, and has negligible effect on estimated arterial viscosity and radius. Due to their anatomical difference, pulse signal amplification is attributed to improved pulse transmission at tissue-sensor interface at the RA and alleviated suppression of the true pulse signal at the STA. The effect of overlying tissue and a uniform layer on estimated arterial parameters is further discussed. The proposed solution offers a low-cost solution to acquiring an amplified pulse signal at P_T near zero for CV health assessment.**



***Index Terms*—Tactile sensor, arterial pulse signals, sensor-artery interaction, pulse transmission, arterial parameters, hold-down pressure, transmural pressure.**

I. Introduction

ARTERIAL Pulse signal manifests the cardiovascular (CV) system condition and is measured for estimation of various arterial indices to infer diagnostic information on the CV system [1]. Since arterial indices are estimated from key features of a measured arterial pulse waveform, acquiring the pulse signal in an artery with minimum distortion is critical for accuracy in estimated arterial indices. Photoplethysmogram (PPG) sensors and applanation tonometry are the two big categories of non-invasive medical instruments for arterial pulse signal measurements in clinical studies [2], [3]. A PPG sensor measures the blood volume change in an artery through optical transduction, and is extremely sensitive to motion artifacts and leads to inaccuracy in estimated arterial indices [3], [4]. An applanation tonometer is essentially a tactile sensor and measures the pulsatile pressure signal in an artery. Its operation demands a high degree of skill for achieving measurement accuracy, because alignment of the single transducer of a tonometer at the center of an artery is extremely challenging, due to small size of an artery [5].

In recent years, the advancement of micro/nano-fabrication technology has rendered development of flexible tactile sensors in various forms for arterial pulse signal measurements [6]–[11]. Achieving the conformity of a tactile sensor to the artery has been pursued for amplifying a measured pulse signal (or high signal-to-noise ratio). For instance, micro-hairy structures built on a flexible supporting layer achieved improved sensor-artery conformity and thus amplified the measured pulse signal [9]. Bioinspired composite microfibers improved both conformity and skin adhesion for amplifying a measured pulse signal [10]. Ultra-flexible sensor arrays were also built from Graphene Nanoplatelet Networks (GNN) patterned on polydimethylsiloxane (PDMS) and soft PDMS coated micro-lines were further attached to the sensor array to improve conformity and skin adhesion so as to amplify a

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The authors are with the Department of Mechanical and Aerospace Engineering, Old Dominion University, Norfolk, VA 23529 USA (e-mail: zlhao@odu.edu; dwang009@odu.edu).

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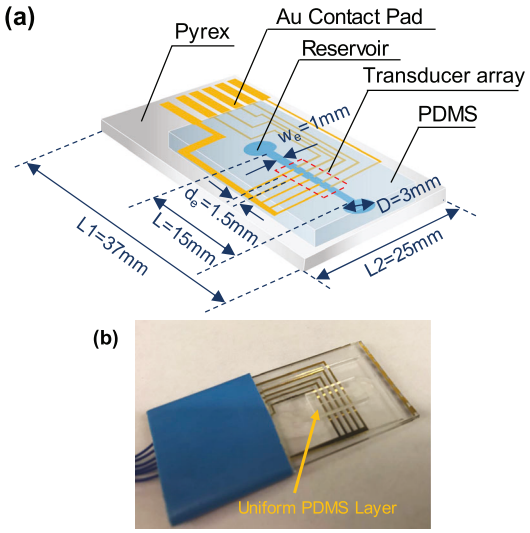


Fig. 1. A Pyrex-based microfluidic tactile sensor (a) schematic with labeled key design parameters and (b) picture with a uniform PDMS layer on the top of microstructure portion above the transducer array.

measured pulse signal [11]. The physical essence of achieving sensor-artery conformity and skin adhesion is improving sensor-artery interaction so that the true pulse signal in an artery is maximally transmitted to the sensor at tissue-sensor interface. Due to their complicated configurations of multiple stacked layers, these tactile sensors entail great fabrication complexity. Interfacial stress between different layers experienced by these sensors during operation also causes performance degradation and undermines long-term reliability of these sensors. Furthermore, theories on sensor-artery interaction for pulse signal amplification are not explored in these experimental studies.

To overcome the technical issues associated with the above-mentioned flexible tactile sensors, this work proposes to add a uniform PDMS layer to a Pyrex-based microfluidic tactile sensor for arterial pulse signal amplification. The sensor entails a PDMS microstructure embedded with a 5×1 resistive transducer array on a Pyrex slide. The standard fabrication process for microfluidic devices is employed for the fabrication of this sensor, thus featuring great fabrication simplicity [12], [13]. The sensor retains performance robustness and long-term reliability, since the Pyrex slide and electrodes on the Pyrex slide in the sensor do not bend during operation. Previously, we demonstrated the feasibility of this sensor for arterial pulse signal measurements [14]. In this work, we explore theories on sensor-artery interaction for pulse signal amplification by relating relevant engineering principles to arterial anatomy and investigate the feasibility of the proposed approach by conducting pulse signal measurements on one subject.

II. Microfluidic Tactile Sensor

As illustrated in Fig. 1, built upon a Pyrex slide, the sensor entails a PDMS microstructure embedded with an electrolyte-filled microchannel underneath and a set of Au/Cr electrode pairs distributed along the microchannel length.

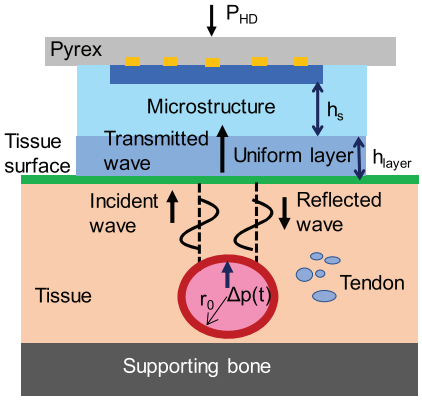


Fig. 2. Physical essence of sensor–artery interaction in arterial pulse signal measurements using the microfluidic tactile sensor and a uniform layer.

The electrolyte across each electrode pair functions as a resistive transducer, whose resistance is a function of the deflection of the microstructure above it. The contact pads for electrical connections are routed at one side for easy use of the sensor. To measure a pulse signal, the transducer array is aligned perpendicular to an artery so that at least one transducer can be easily aligned at an artery site by a layperson, as illustrated in Fig. 2 and Fig. 3. Upon a hold down pressure, P_{HD}, exerted on the Pyrex side of the sensor, the pulse signal reaching the tissue surface passes through the tissue-sensor interface, deflects the PDMS microstructure, and registers as a resistance change by the transducer at the artery. The details about the sensor fabrication [12], [13] and its acquisition of arterial pulse signals can be found in our previous work [15]. To improve sensor-artery interaction, a uniform PDMS layer is placed on the top of microstructure portion above the transducer array, as shown in Fig. 1(b).

III. Related Theories on Sensor-Artery Interaction for Arterial Pulse Signal Amplification

As shown in Fig. 2, pertaining to pulse signal measurements, the key design parameter of the sensor is its microstructure thickness, h_s, referred to as sensor thickness. A uniform layer of thickness, h_{layer}, is added between the sensor and the tissue surface. We assume that the sensor and the uniform layer are perfectly bonded together, upon exertion of P_{HD} in a pulse measurement. Thus, addition of the uniform layer is equivalent to increasing the sensor thickness, adding flexibility to the sensor design. From the perspective of elastic wave propagation, the physical essence of sensor-artery interaction in a pulse measurement is pulse signal transmission at the tissue-sensor interface.

Similar to an earthquake [16], the pulse signal in an artery acts as an excitation source and excites an elastic wave propagating in the overlying tissue above the artery and is incident on the tissue surface. When the tissue surface above the artery is free of a sensor, the pulse signal is completely reflected back into the overlying tissue [17], [18]. When a sensor is at the tissue surface for measuring the pulse signal incident on the tissue surface, the effect of P_{HD} on

a measured pulse signal is twofold. On the one hand, P_{HD} is needed for entailing conformal contact at the tissue-sensor interface so that a portion of the incident wave transmits to the sensor and registers as the measured pulse signal by the sensor. Ideally, a high P_{HD} translates to improved conformal contact at the tissue-sensor interface and thus improved pulse signal transmission or an amplified measured pulse signal [18]. On the other hand, excessive P_{HD} may occlude the blood flow and suppress the true pulse signal in an artery. In other words, different from an earthquake, as the excitation source, the true pulse signal in an artery may be suppressed by excessive P_{HD} [19].

Sensor-artery interaction is a function of P_{HD}, the overlying and adjacent tissue, and the sensor thickness. The radial artery (RA) is deeply embedded under the skin and its adjacent tissue contains tendon [20]. Even under high P_{HD}, a thin sensor may not deform enough to generate conformal contact at the tissue-sensor interface for desirable pulse signal transmission, since high P_{HD} may be shouldered by the adjacent rigid tissue. Conversely, the superficial temporal artery (STA) is very near the skin and no rigid tissue is in its adjacent region. Then, even moderate P_{HD} on a thin sensor will result in conformal contact at the tissue-sensor interface but may suppress the true pulse signal in the artery.

With the anatomical difference between the RA and the STA in mind, a PDMS layer is added in pulse signal measurements at the two arteries for different functions. At the RA, a PDMS layer is added to accommodate the thick overlying tissue and the adjacent rigid tissue for generating the needed deformation in the sensor to achieve conformal contact at the tissue-sensor interface. At the STA, a PDMS layer is added to accommodate the thin overlying tissue for alleviating suppression of the true pulse signal in the artery even under moderate P_{HD}.

IV. Arterial Pulse Signal Measurements

The protocols and informed consent documents for arterial pulse measurements were approved by the Institutional Review Board (IRB) at Old Dominion University. The consent form was obtained from a 28yr-old male healthy subject with body mass index (BMI) of 27.9, prior to pulse measurements.

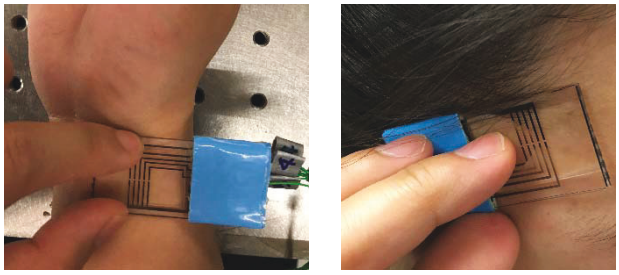
A. Preparation of Uniform PDMS Layers

A set of uniform PDMS layers were prepared using Sylgard 184 silicone elastomer kit (Dow Corning Corp.). The uniformity of a layer was defined as being made of PDMS of the same mixing ratio and having the same thickness across the whole layer. The same procedure was used to prepare all the uniform PDMS layers. Briefly, a mixture of a chosen mixing ratio of curing agent to elastomer base was poured into a container and was kept at room temperature 48hrs for curing. Then, the cured mixture was peeled off from the container and cut into an approximately 15mm $\times$ 10mm in-plane dimension. The thickness of a PDMS layer was controlled by the amount of a mixture poured into the container.

In order to examine the effect of the viscoelasticity and the thickness of a uniform PDMS layer on the measured results, nine uniform PDMS layers with three mixing ratios: 1:10, 1:20, and 1:30, and three thicknesses: 1mm, 2mm, and 3.5mm, were prepared for the RA. The three 1mm-thick uniform layers of three mixing ratios: 1:10, 1:20, and 1:30, were also utilized for the superficial temporal artery (STA). All the layers had the same in-plane dimension.

TABLE I
Elastic Modulus of PDMS Materials at Different Mixing Ratios of Curing Agent to Elastomer Base [21]

Mixing ratio	Elastic Modulus (MPa)
1:10	1.545
1:20	0.445
1:30	0.17



(a) (b)

Fig. 3. Pictures of pulse signal measurements (a) at the RA (b) at the STA.

Elastic modulus of PDMS at the three chosen mixing ratios has been reported in the literature [21] and is summarized in Table I. The viscosity of PDMS has been found to increase with its mixing ratio, but the metric for PDMS viscosity has not been well established [22], [23] and thus is not included in the table.

B. Pulse Measurements at the RA and the STA

The RA was chosen for its convenience and the STA was chosen for its suitability in special situations [24]. The pulse signal was first measured by the sensor with no PDMS layer, as a reference for evaluating the effect of a PDMS layer on a measured pulse signal. The details about the experimental setup and apparatus for pulse signal measurements using the sensor can be found in our previous work [15]. To measure a pulse signal, a PDMS layer was placed at the artery and then the sensor was put on top of the uniform layer. Afterward, two fingers were used to hold the sensor and maintain the stillness of the sensor as shown in Fig. 3. Furthermore, the holding-strength of two fingers was utilized to adjust P_{HD} on the sensor for achieving conformal contact or alleviating suppression of the true pulse signal through real-time monitoring the pulse signal for its maximum amplitude in a LabVIEW program.

Such manual control of P_{HD} was chosen for two reasons. First, this approach is of practical utility in real life, because the sensor can be easily held by the subject or a caregiver. Owing to respiratory-related motion artifacts and non-respiratory-related motion artifacts (e.g., body shifting of the subject and finger jittering), P_{HD} fluctuates over time in a pulse measurement. A control mechanism with a feedback

loop on quantitatively controlling P_{HD} may introduce another source of motion artifact to P_{HD} and consequently contaminate a measured pulse signal.

The same sensor with $h_s = 1$mm was used for all the pulse measurements with the PDMS layers. Each pulse signal was recorded for a 40-60sec time duration with a sampling rate of 1kHz. Owing to the variation in alignment between the transducer array and an artery, the transducer capturing the pulse signal with the largest amplitude varied among measurements. Only the pulse signal with the largest amplitude in a measurement was chosen for estimation of arterial parameters through pulse waveform analysis.

C. Data-Processing for Pulse Waveform and Hold-Down Pressure

The originally recorded pulse signal is the resistance signal of a transducer in the sensor, $R(t)$. Before aligning the sensor with/without a uniform layer on an artery, the sensor was free of loading, the resistance of a transducer was defined as its original resistance, R_0. Owing to fabrication variation, the original resistance varied among the transducers, although the transducers were designed identical. In a pulse measurement, P_{HD} caused a deflection in the sensor and thus a change in the original resistance of a transducer. The resistance of a transducer under P_{HD} was defined as its initial resistance, $R_0'(t)$. Therefore, the resistance change relative to the initial resistance represented a measured pulse signal:

$$\Delta R(t) = R(t) = R_0'(t) \tag{1}$$

The peak of each pulse cycle, $\Delta R(t)$, was the pulse amplitude and was denoted as u_{r0}.

Unavoidable motion artifacts caused fluctuation in P_{HD} and consequently variation of this initial resistance with time. The initial resistance became the baseline for a measured pulse signal. The related signal-processing algorithm was used to estimate the initial resistance, $R_0'(t)$, from the originally recorded resistance signal, $R(t)$. The details about the algorithm can be found in the literature [25]. The difference between the original resistance and the initial resistance normalized to the original resistance was indicative of P_{HD}:

$$P_{HD} \propto \frac{R_0'(t) - R_0}{R_0} \tag{2}$$

D. Pulse Waveform Analysis for Estimation of Arterial Parameters

Owing to great similarity between the pulsatile pressure waveform and the radius waveform at an artery [26], the measured arterial pulse waveform, $\Delta R(t)$, from the subject was utilized to represent the arterial radius waveform, $u_r(t)$, with its amplitude being denoted as u_{r0}. Owing to time-harmonic nature of a pulse signal, the arterial wall was modeled as a unit-mass dynamic system with its spring stiffness, K, and damping coefficient, D, being expressed as:

$$D = \frac{\eta}{\rho_w r_0^2}, \quad K = \frac{E}{\rho_w r_0^2} \tag{3}$$

where ρ_w is the arterial wall density and r_0 is the arterial radius at diastolic blood pressure (DBP), as shown in Fig. 2. The values of K and D were estimated by the key features in a measured pulse waveform:

$$K \propto \frac{a_{\max} - a_{\min}}{u_{r0}} \cdot \frac{\Delta t}{T}, \quad D \propto \frac{a_{\max} - a_{\min}}{v_{\max}} \cdot \frac{\Delta t}{T} \tag{4}$$

where T is time duration of a pulse cycle; a_{max} and a_{min} are the maximum and minimum amplitudes of the second-order derivative of a pulse waveform, respectively; Δt is time duration between a_{max} and a_{min}; andv_{max} is the maximum of the first-order derivative of a pulse waveform. Finally, three arterial parameters: elasticity E, viscosity η and radius r_0 of the arterial wall were estimated as [14], [27]:

$$E \propto \frac{K}{\sqrt{D}}, \quad \eta \propto \sqrt{D}, \; r_0 \propto 1/\sqrt[4]{D} \tag{5}$$

Note that the above expressions are relative estimation of the three arterial parameters from a scaling analysis [14]. Thus, the estimated values of three arterial parameters do not have units and do not represent their true values.

V. Measured Results and Discussion

Figures 4 and 5 show the effect of a uniform PDMS layer on the measured pulse amplitude at the RA and the STA, respectively. Tables II(a) and II(b) summarize the estimated values of the relevant parameters from the measured pulse signals at the RA and the STA, respectively. The mean value of P_{HD} and the mean value and standard deviation of K, D, E, η and r_0 were estimated from five pulse cycles in each measured pulse signal. Overall, the measured pulse amplitude at the two arteries was amplified by addition of a uniform PDMS layer, regardless of its thickness and viscoelasticity. As compared to the sensor with no uniform layer, addition of a uniform layer caused a moderate increase in the estimated K and negligible change in the estimated D. According to Eq. (5), the estimated E also registered a moderate increase and the estimated η and r_0 revealed negligible change, as compared to their counterparts measured with no PDMS layer.

A. Effect of Transmural Pressure on Arterial Elasticity and its Variation Among Measurements

Other than affecting pulse transmission at tissue-sensor interface and the true pulse signal in an artery in sensor-artery interaction, P_{HD} is also critical in influencing transmural pressure, P_T, in the arterial wall:

$$P_T = DBP - P_{ext} \tag{6}$$

where P_{ext} is the external pressure outside the arterial wall, as depicted in Fig. 6(a). P_T has a notorious influence on E, since E shows a sigmoid relation with P_T: when $P_T \cong 0$, E is the lowest and the measured pulse signal reaches its maximum amplitude; when P_T is away from zero (either negative or positive), E increases dramatically and the measured pulse amplitude decreases [28]–[30]. To the best knowledge of the authors, the influence of P_T on η has not been reported in the literature.

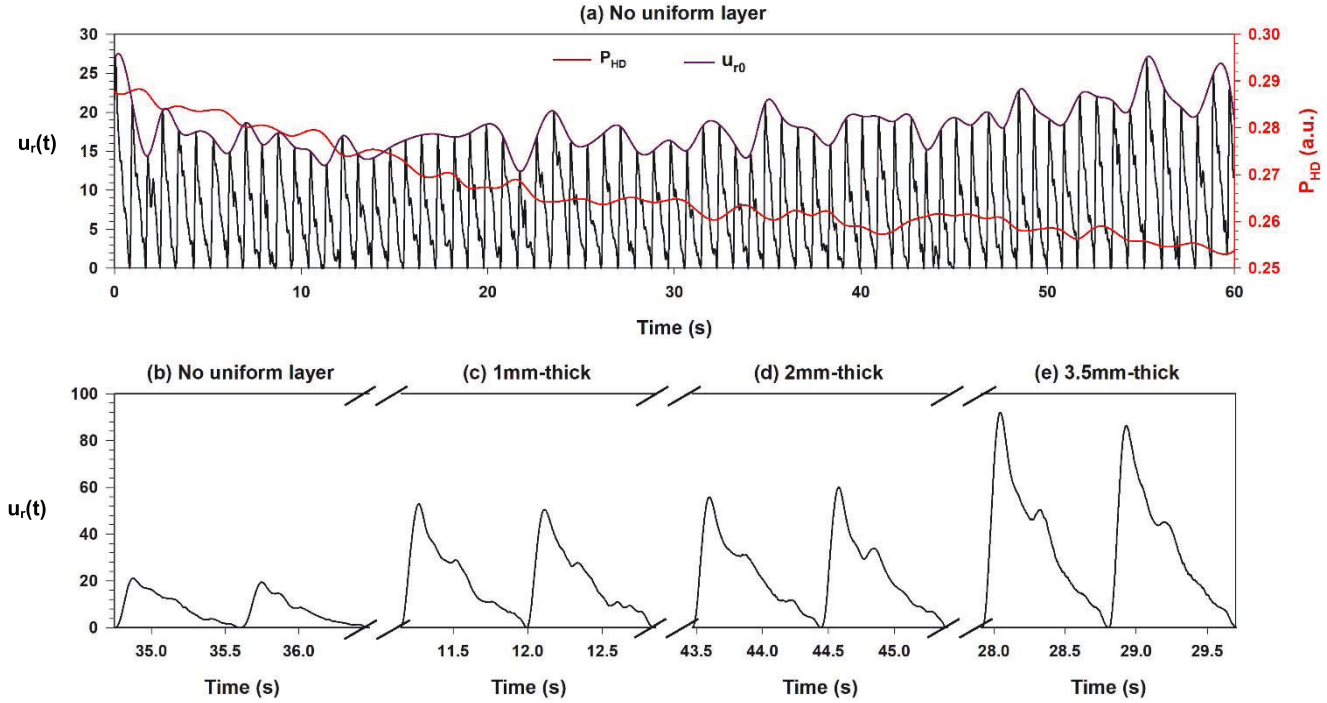


Fig. 4. Measured pulse signals (a) at the RA using the microfluidic tactile sensor (b) with no uniform layer and with a uniform PDMS layer with 1:10 mixing ratio and (c) 1mm thickness (d) 2mm thickness (e) 3.5mm thickness.

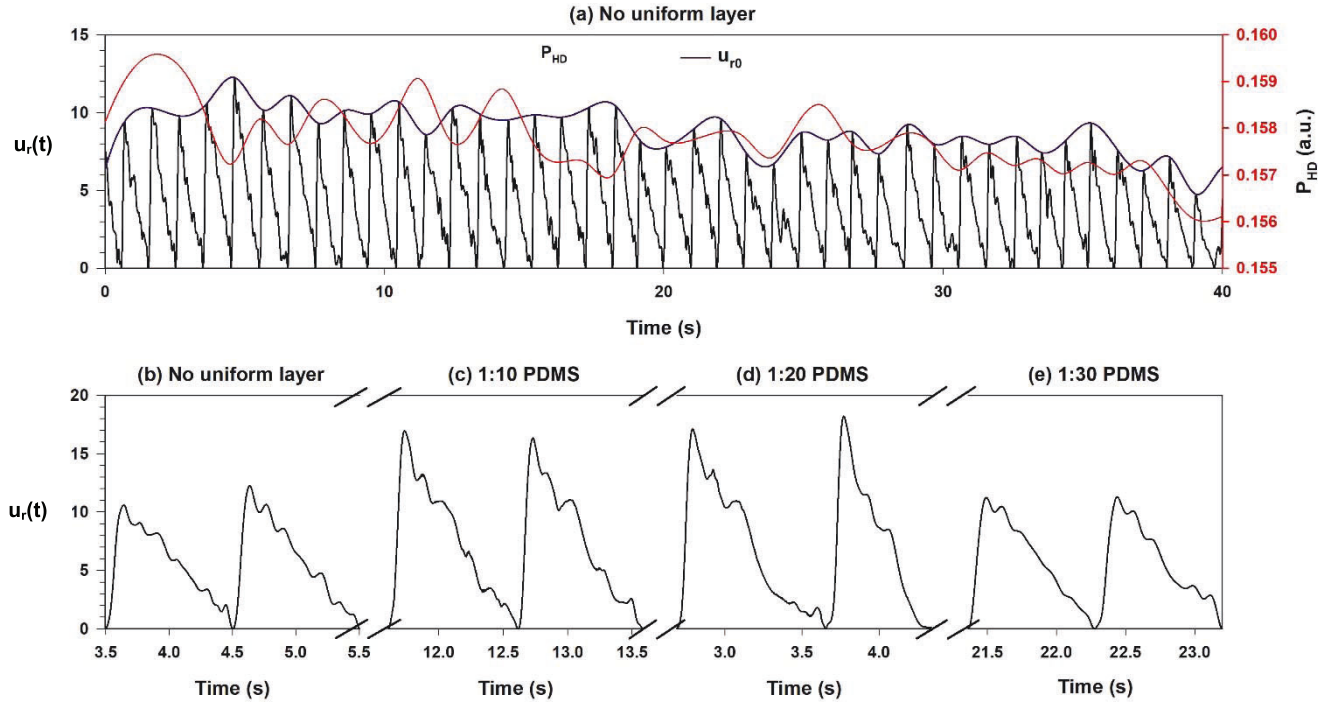


Fig. 5. Measured pulse signals (a) at the STA using the microfluidic tactile sensor (b) with no uniform layer and with a uniform PDMS layer with 1m thickness and (c) 1:10 mixing ratio (d) 1:20 mixing ratio (e) 1:30 mixing ratio.

Variation in P_T exists in the same pulse measurement. P_{HD} affects P_T through affecting P_{ext}. Since the position of the sensor relative to an artery is fixed in a measurement, P_{HD} is indicative of P_T in the arterial wall. As shown in Figs. 4(a) and 5(a), P_{HD} varies randomly over time, because unavoidable motion artifacts cause fluctuation in sensor-artery interaction. Yet, there is no correlation between u_{r0} with P_{HD} in the same pulse measurement, possibly because the dynamic nature of motion artifacts causes random time-varying variation in P_T and arterial parameters.

Variation in P_T also exists among measurements at an artery. However, neither the estimated P_{HD} nor the estimated u_{r0} are suitable for comparison of P_T among measurements.

TABLE II
ESTIMATED VALUES OF RELEVANT PARAMETERS FROM MEASURED ARTERIAL PULSE SIGNALS WITH AND WITHOUT A UNIFORM PDMS LAYER AT THE RA AND THE STA

(A) RADIAL ARTERY (RA)

PDMS layer	Thickness	P_{HD}	u_{r0}	HR(bmp)	E	η	r_0	K ($1/s^2$)	D (1/s)
No layer	--	0.2593	20.08±1.53	70±2	72.83±4.49	3.00±0.45	0.581±0.04	218.8±38.2	9.15±2.87
	1mm	0.2132	50.10±2.53	70±1	82.27±3.80	3.20±0.25	0.560±0.022	263.5±28.0	10.28±1.59
1:10	2 mm	0.2169	62.58±2.30	65±2	75.53±3.91	2.96±0.15	0.582±0.015	223.7±23.0	8.76±0.90
	3.5 mm	0.2608	87.66±2.74	68±1	79.13±3.66	3.04±0.20	0.574±0.018	241.4 ±27.1	9.30±1.26
	1 mm	0.2352	61.46±3.39	68±3	76.02±3.60	3.03±0.25	0.576±0.024	230.6±29.7	9.20±1.54
1:20	2 mm	0.2756	59.94±2.22	65±1	74.89±1.39	2.99±0.16	0.579±0.015	223.8±13.3	8.95±0.93
	3.5 mm	0.3057	33.98±2.33	73±6	86.83±12.23	2.74±0.39	0.608±0.043	241.7±66.6	7.64±2.14
	1 mm	0.2399	59.71±1.85	66±3	76.60±2.77	2.84±0.10	0.593±0.010	217.9±14.2	8.09±0.55
1:30	2 mm	0.2730	65.19±1.36	66±2	78.83±1.31	3.08±0.06	0.570±0.006	242.7±7.7	9.48±0.38
	3.5 mm	0.3261	41.27±1.32	65±2	77.57±4.67	3.21±0.14	0.558±0.013	249.4±21.8	10.33±0.91

(B) SUPERFICIAL TEMPORAL ARTERY (STA)

PDMS layer	Thickness	P_{HD}	u_{r0}	HR(bmp)	E	η	r_0	K ($1/s^2$)	D (1/s)
No layer	--	0.1583	10.58±0.96	63±3	70.40±9.25	2.60±0.28	0.623±0.035	184.4±39.5	6.81±1.43
1:10	1 mm	0.2188	18.02±1.21	60±1	81.28±5.30	2.33±0.06	0.655±0.008	189.6±9.0	5.45±0.26
1:20	1 mm	0.2025	16.57±0.62	71±11	98.24±6.05	2.48±0.11	0.636±0.015	243.8±23.7	6.15±0.55
1:30	1 mm	0.2082	14.14±1.04	67±2	70.82±5.91	2.77±0.16	0.602±0.017	196.6±24.6	7.68±0.86

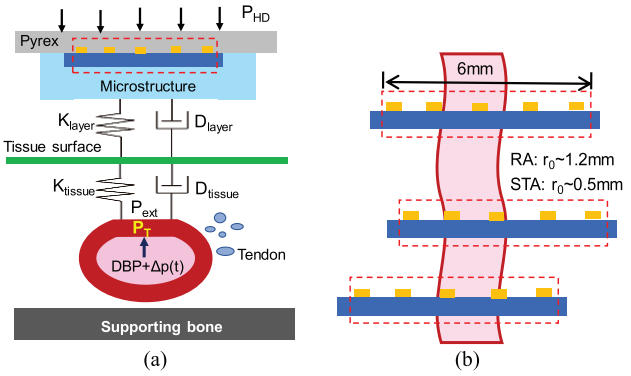


Fig. 6. Schematics of (a) transmural pressure, P_T, in the arterial wall and an equivalent lumped-element model of a uniform PDM layer and overlying tissue (b) variation of the sensor position relative to an artery.

The estimated P_{HD} is the pressure at the location of the transducer whose output is used for estimation of arterial parameters. As shown in Fig. 6(b), it is impossible to align the same transducer exactly at the center of an artery among measurements. Both the transducer aligned closest to an artery for obtaining the maximum pulse amplitude and to what extent the closest-aligned transducer is off the center of an artery vary among measurements. Thus, P_T in the arterial wall is believed to vary randomly among measurements. Then, variation in P_T among measurements might result in no correlation between P_{HD} and u_{r0}, as shown in Table II.

B. Function of a Uniform PDMS Layer

Despite variation in P_T among measurements, the measured pulse signal was indeed amplified by addition of a PDMS layer. Since the estimated arterial parameters do not show dependence on the mixing ratio (or viscoelasticity) and thickness of a uniform PDMS layer, all the measurements with different PDMS layers are combined together and are compared with the measurement with no PDMS layer, as shown in Table III. In the table, coefficient of variation (CV: standard deviation/mean) is used to evaluate repeatability of the measurements using different PDMS layers and percent change is defined as change relative to the measurement with no PDMS layer to quantify the effect of a PDMS layer on a measured pulse signal.

At the RA, the arterial wall without P_{HD} is in highly tensile state ($P_T >> 0$). In a measurement with no PDMS layer, the thick overlying tissue and the adjacent rigid tissue make it difficult to achieve conformal contact at the tissue-sensor interface, even under high P_{HD}. Consequently, pulse transmission at tissue-surface interface is low and the measured pulse signal is small. Meanwhile, the arterial wall is only slightly compressed and then is still highly tensile ($P_T >> 0$). In contrast, addition of a PDMS layer improves conformal contact at the tissue-sensor surface so that pulse transmission at tissue-sensor interface is high and gives rise to a large pulse signal, and compression of the arterial wall increases and the arterial wall is in a low tensile or compressive state ($P_T \rightarrow 0$). Given the effect of P_T on E, the estimated E is expected to be lower than that measured with no PDMS layer.

The STA is smaller than the RA and very near the skin. In a measurement with no PDMS layer under even moderate P_{HD}, the arterial wall is severely compressed (or occluded) and is in highly compressive state (or $P_T << 0$) and the true pulse signal is severely suppressed. Addition of a PDMS layer cushions the effect of P_{HD} on the artery, and thus alleviates compression of the arterial wall so that the arterial wall is in a low compressive or tensile state ($P_T \rightarrow 0$), and the true pulse signal is less suppressed and gives rise to a large measured pulse signal. Then, the estimated E with addition of a PDMS layer is expected to be lower than that measured with no PDMS layer.

TABLE III
EVALUATION OF THE EFFECT OF ADDITION OF A PDMS LAYER AT THE RA AND THE STA

(a) RADIAL ARTERY (RA)

	u_{r0}	HR (bmp)	E	η	r_0	K ($1/s^2$)	D (1/s)	Relation to P_T
With no uniform PDMS layer	20.08	70	72.83	3	0.581	218.8	9.15	$P_T>>0$
Measurements with nine uniform PDMS layers								
Mean	57.99	67.33	78.63	3.01	0.578	235.83	9.03	$P_T \rightarrow 0$
Standard deviation	15.37	2.74	3.82	0.15	0.016	14.29	0.94	
CV	26.5%	4.1%	4.9%	5.1%	2.7%	6.1%	10.4%	
Percent change	188.8%	-3.8%	8.0%	0.3%	-0.6%	7.8%	-1.3%	

(b) SUPERFICIAL TEMPORAL ARTERY (STA)

	u_{r0}	HR (bmp)	E	η	r_0	K ($1/s^2$)	D (1/s)	Relation to P_T
With no uniform PDMS layer	10.58	63	70.4	2.6	0.623	184.4	6.81	$P_T<<0$
Measurements with three uniform PDMS layers								
Mean	16.24	66.0	83.45	2.53	0.631	209.25	6.39	$P_T \rightarrow 0$
Standard deviation	1.96	5.6	13.84	0.22	0.027	29.27	1.12	
CV	12.1%	8.4%	16.6%	8.9%	4.3%	14.0%	17.5%	
Percent change	53.5%	4.8%	18.5%	-2.8%	1.3%	13.5%	-6.2%	

Contradictory to the effect P_T on E, addition of a PDMS layer causes a moderate increase in the estimated Eat both arteries, as shown in Table III. Now, we explore the reason underlying this contradiction. Since the PDMS materials used are viscoelastic, a PDMS layer functions as an elastic spring, K_{layer}, and a damper, D_{layer}, in the pulse transmission from an artery to the sensor, as shown in Fig. 6(a). Similarly, the overlying tissue also functions as K_{tissue} and D_{tissue}. Then, a moderate increase in the estimated E by addition of a PDMS layer might be attributed to K_{layer}.

The expressions of K and D in Eq. (3) represent the spring stiffness and damping coefficient per unit mass of an artery. As compared with the STA, the RA has a larger radius and a lower viscoelasticity [31]. Thus, K and D at the RA are expected to be smaller than their counterparts at the STA. Yet, the estimated K and D in Table II revealed the opposite. Since difference in the estimated K between the two arteries is relatively large, variation in P_T is ruled out as the cause. Given that the overlying tissue at the RA is much thicker than that at the STA, the overlying tissue is believed to make non-negligible contribution to the estimated K and D. Consequently, contribution of K_{layer} is believed to be the cause of a moderate increase in the estimated E. Since the estimated D is not affected by addition of a PDMS layer, the effect of D_{layer} on the estimated η and r_0 is negligible, possibly implying that the overlying tissue is more viscous than the PDMS materials used. Given their small difference in the estimated K and D between measurements using different PDMS layers, their independence of the mixing ratio (or viscoelasticity) and thickness might be caused by variation in P_T. Taken together, the overlying tissue has influence on the estimated K and D to a much larger extent than a PDMS layer. Yet, the influence of a PDMS layer on the estimated E overpasses the influence of variation in P_T on the estimated E. This explains the reason why the estimated E at $P_T \rightarrow 0$ is higher than that at $P_T >> 0$ or $P_T << 0$. The estimated E does not show dependence on the mixing ratio and thickness of a PDMS layer, possibly because variation in the viscoelasticity and thickness of a PDMS layer is negligible, as compared with variation in P_T.

C. Advantages of a Uniform PDMS Layer

For the purpose of acquiring a pulse signal with minimum distortion and improving accuracy in estimated arterial parameters, addition of a uniform PDMS layer has several advantages over the measurement with no PDMS layer. In clinical studies, E is commonly estimated from pulse signals measured at $P_T \rightarrow 0$ and is used for detection and diagnosis of CV diseases [28]–[30]. As discussed above, the measurement with no PDMS layer results in a small pulse signal at $P_T >> 0$ or $P_T << 0$. In contrast, addition of a PDMS layer allows acquiring an amplified pulse signal at $P_T \rightarrow 0$.

From the perspective of measurement accuracy, high pulse transmission at tissue-sensor interface and low suppression of the true pulse signal in an artery will reduce distortion in the measured pulse signal, and a high signal-to-noise ratio will reduce the effect of noise on the measured pulse signal. Then, the estimated arterial parameters from the measurement with no PDMS layer may carry larger measurement errors. From the perspective of measurement repeatability, a uniform PDMS layer alleviates the effect of P_{HD} on P_T, and then variation in P_T is relatively moderate among the measurements using different PDMS layers at an artery. As shown in Table III, measurement repeatability of the estimated E at the RA is 4.9%, which is comparable to that on arterial stiffness using applanation tonometry [32]. Measurement repeatability at the STA is relatively high, possibly because the effect of P_{HD} on P_T is relatively high, due to its small size and closeness to the skin. Note that there are no studies on the STA in the literature for comparing its measurement repeatability. It is worth mentioning that the effect of P_{HD} or P_T on the measured results was not examined in the studies on arterial pulse signal amplification using flexible tactile sensors [8]–[11].

As explained earlier, the overlying tissue has non-negligible influence on the estimated E. Given the same instrument, the estimated E needs to be interpreted at the individual level, simply because the overlying tissue varies among subjects with different characteristics (e.g., ages and BMI) [20], [33]. Therefore, interpretation of estimated arterial parameters needs to factor in the instrument design and subject-specificity [14], [33]–[35]. This explains the current trend on subject-specific calibration and modeling of CV parameters [34], [35]. Despite causing a moderate increase in the estimated E, addition of a PDMS layer allows obtaining a pulse signal at P_T near zero with reduced distortion.

D. Study Limitations

Currently, the three arterial parameters are only measured at the carotid artery (CA) using applanation tonometry and an imaging instrument [14] in clinical studies. Applanation tonometry is commonly used at the RA to measure its pulse signal and derive arterial indices indicative of E. PPG sensors have been used at the STA to measure its pulse signal and derive only its blood pressure, and no arterial indices indicative of E have been obtained yet [24]. As such, the estimated arterial parameters in this work cannot be quantitatively compared with the related data in the literature.

Since this work is aimed to validate the feasibility of amplifying a measured pulse signal by addition of a uniform PDMS layer to a microfluidic tactile sensor, only one subject was measured. Despite lacking in statistical significance, the measured results with different PDMS layers collectively revealed amplified pulse signals, as compared to the measurement with no PDMS layer. Further validation of the effect of a uniform PDMS layer and overlying tissue on the estimated arterial parameters with statistical significance will be conducted in the future. The rationale of the proposed approach is expected to be applicable for subjects with different characteristics. Evidently, since overlying tissue, artery size, and true pulse signal amplitude all vary among subjects [20], thickness of a PDMS layer may need to be adjusted accordingly for desired sensor-artery interaction.

VI. Conclusion

In this work, addition of a uniform PDMS layer to a Pyrex-based microfluidic tactile sensor is investigated for arterial pulse signal amplification. Pulse signal measurements at the RA and the STA of a subject are conducted with the same sensor and a set of uniform PDMS layers with different mixing ratios and thickness. As compared with those measured with no PDMS layer, the measured pulse signals at the two arteries are amplified by addition of a PDMS layer. The amplified measured pulse signal at the RA is attributed to improved conformal contact at the tissue-sensor interface, due to its thick overlying tissue and adjacent rigid tissue. In contrast, the amplified measured pulse signal at the STA is attributed to alleviated suppression of the true pulse signal in the artery, due to its small size and closeness to the skin. Addition of a uniform layer causes a moderate increase in estimated arterial elasticity and does not affect estimated arterial viscosity and radius. The related theories and the measured results are combined to examine the effect of overlying tissue and a PDMS layer on estimated arterial parameters. As compared with those flexible tactile sensors, the propose approach offers a low-cost solution to arterial pulse signal amplification for CV health assessment.

References

[1] W. B. Kannel, C. Kannel, R. S. Paffenbarger, Jr., and L. A. Cupples, "Heart rate and cardiovascular mortality: The framingham study," *Amer. Heart J.*, vol. 113, no. 6, pp. 1489–1494, Jun. 1987.

[2] G. M. Drzewiecki, J. Melbin, and A. Noordergraaf, "Arterial tonometry: Review and analysis," *J. Biomech.*, vol. 16, no. 2, pp. 141–152, 1983.

[3] J. Allen, "Photoplethysmography and its application in clinical physiological measurement," *Physiol. Meas.*, vol. 28, no. 3, p. R1, Feb. 2007.

[4] J. F. Rojano and C. V. Isaza, "Singular value decomposition of the time-frequency distribution of PPG signals for motion artifact reduction," *Int. J. Signal Process. Syst.*, vol. 4, no. 6, pp. 475–482, Dec. 2016.

[5] K. Matthys and P. Verdonck, "Development and modelling of arterial applanation tonometry: A review," *Technol. Health Care*, vol. 10, no. 1, pp. 65–76, Jan. 2002.

[6] C. Pang *et al.*, "A flexible and highly sensitive strain-gauge sensor using reversible interlocking of nanofibres," *Nature Mater.*, vol. 11, no. 9, pp. 795–801, 2012.

[7] Y. Zang, F. Zhang, C.-A. Di, and D. Zhu, "Advances of flexible pressure sensors toward artificial intelligence and health care applications," *Mater. Horizons*, vol. 2, no. 2, pp. 140–156, 2015.

[8] C. Dagdeviren *et al.*, "Conformable amplified lead zirconate titanate sensors with enhanced piezoelectric response for cutaneous pressure monitoring," *Nature Commun.*, vol. 5, p. 4496, Aug. 2014.

[9] C. Pang *et al.*, "Highly skin-conformal microhairy sensor for pulse signal amplification," *Adv. Mater.*, vol. 27, no. 4, pp. 634–640, Jan. 2015.

[10] D. M. Drotlef, M. Amjadi, M. Yunusa, and M. Sitti, "Bioinspired composite microfibers for skin adhesion and signal amplification of wearable sensors," *Adv. Mater.*, vol. 29, no. 28, Jul. 2017, Art. no. 1701353.

[11] Y. Park *et al.*, "Microtopography-guided conductive patterns of liquid-driven graphene nanoplatelet networks for stretchable and skin-conformal sensor array," *Adv. Mater.*, vol. 29, no. 21, Jun. 2017, Art. no. 1606453.

[12] W. Gu *et al.*, "Detection of distributed static and dynamic loads with electrolyte-enabled distributed transducers in a polymer-based microfluidic device," *J. Micromech. Microeng.*, vol. 23, no. 3, 2013, Art. no. 035015.

[13] P. Cheng, W. Gu, J. Shen, A. Ghosh, A. Beskok, and Z. Hao, "Performance study of a PDMS-based microfluidic device for the detection of continuous distributed static and dynamic loads," *J. Micromech. Microeng.*, vol. 23, no. 8, p. 085007, 2013.

[14] D. Wang, L. Reynolds, T. Alberts, L. Vahala, and Z. Hao, "Model-based analysis of arterial pulse signals for tracking changes in arterial wall parameters: A pilot study," *Biomech. Model. Mechan.*, vol. 18, no. 6, pp. 1629–1638, Dec. 2019.

[15] D. Wang, J. Shen, L. Mei, S. Qian, J. Li, and Z. Hao, "Performance Investigation of a Wearable Distributed-Deflection Sensor in Arterial Pulse Waveform Measurement," *IEEE Sensors J.*, vol. 17, no. 13, pp. 3994–4004, Jul. 1, 2017.

[16] Z. Hao, A. Erbil, and F. Ayazi, "An analytical model for support loss in micromachined beam resonators with in-plane flexural vibrations," *Sens. Actuators A, Phys.*, vol. 109, pp. 156–164, Dec. 2003.

[17] B. Drinkwater, R. Dwyer-Joyce, and P. Cawley, "A study of the interaction between ultrasound and a partially contacting solid—Solid interface," *Proc. Roy. Soc. London. Ser. A: Math., Phys. Eng. Sci.*, vol. 452, no. 1955, pp. 2613–2628, Dec. 1996.

[18] S. Biwa, S. Nakajima, and N. Ohno, "On the acoustic nonlinearity of solid-solid contact with pressure-dependent interface stiffness," *J. Appl. Mech.*, vol. 71, no. 4, pp. 508–515, Jul. 2004.

[19] J. Lee and K. C. Nam, "Tonometric vascular function assessment," in *Biomedical Engineering*. London, U.K.: IntechOpen, 2009.

[20] P. Singh, M. I. Choudhury, S. Roy, and A. Prasad, "Computational study to investigate effect of tonometer geometry and patient-specific variability on radial artery tonometry," *J. Biomech.*, vol. 58, pp. 105–113, Jun. 2017.

[21] A. Sharfeddin, A. A. Volinsky, G. Mohan, and N. D. Gallant, "Comparison of the macroscale and microscale tests for measuring elastic properties of polydimethylsiloxane," *J. Appl. Polym. Sci.*, vol. 132, no. 42, p. 42680, Nov. 2015.

[22] Z. Hajjarian, and S.K., Nadkarni, "Evaluating the viscoelastic properties of tissue from laser speckle fluctuations," *Sci. Rep.*, vol. 2, p. 316., Mar. 2012.
[23] M. Fincan, "Assessing viscoelastic properties of polydimethylsiloxane (PDMS) using loading and unloading of the macroscopic compression test," M.S. thesis, Dept. Chem. Eng., Univ. South Florida, Tampa, FL, USA, 2015.
[24] J. Canning *et al.*, "Noninvasive and continuous blood pressure measurement via superficial temporal artery tonometry," in *Proc. 38th Annu. Int. Conf. IEEE Eng. Med. Biol. Soc. (EMBC)*, Orlando, FL, USA, Aug. 2016, pp. 3382–3385.
[25] D. Wang, D. Krusienski, and Z. Hao, "A flexible PET-based wearable sensor for arterial pulse waveform measurement," in *Proc. Int. Joint Conf. Biomed. Eng. Sys. Tech. (BIOSTEC)*, Rome, Italy, Feb. 2016, pp. 66–75.
[26] D. Wang, L. Vahala, and Z. Hao, "Radial and longitudinal motion of the arterial wall: Their relation to pulsatile pressure and flow in the artery," *Phys. Rev. E, Stat. Phys. Plasmas Fluids Relat. Interdiscip. Top.*, vol. 98, no. 3, Sep. 2018, Art. no. 032402.
[27] D. Wang, L. Vahala, T. Alberts, and Z. Hao, "Arterial wall motion and its dynamic modeling for arterial stiffness and damping," in *Proc. ASME Int. Mech. Eng. Congr. Expo. (IMECE)*, Pittsburg, PA, USA, Nov. 2018, Art. no. V003T04A072.
[28] A. Grabovskis, Z. Marcinkevics, O. Rubenis, U. Rubins, and V. Lusa, "Photoplethysmography system for blood pulsation detection in unloaded artery conditions," *Proc. SPIE, Biophoton, Photon. Solutions Better Health Care III*, vol. 8427, no. L-1, May 2012, Art. no. 84270L.
[29] X. F. Teng and Y. T. Zhang, "Theoretical study on the effect of sensor contact force on pulse transit time," *IEEE Trans. Biomed. Eng.*, vol. 54, no. 8, pp. 1490–1498, Aug. 2007.
[30] M. Guerrisi, I. Vannucci, and N. Toschi, "Differential response of peripheral arterial compliance-related indices to a vasoconstrictive stimulus," *Physiol. Meas.*, vol. 30, no. 1, pp. 81–100, Jan. 2009.
[31] P. Reymond, Y. Bohraus, F. Perren, F. Lazeyras, and N. Stergiopulos, "Validation of a patient-specific one-dimensional model of the systemic arterial tree," *Amer. J. Physiol.-Heart Circulatory Physiol.*, vol. 301, no. 3, pp. H1173–H1182, May 2011.
[32] J. D. Suever *et al.*, "Reproducibility of pulse wave velocity measurements with phase contrast magnetic resonance and applanation tonometry," *Int. J. Cardiovascular Imag.*, vol. 28, no. 5, pp. 1141–1146, Jun. 2012.
[33] G. Greiwe *et al.*, "Is applanation tonometry a reliable method for monitoring blood pressure in morbidly obese patients undergoing bariatric surgery," *BJA Brit. J. Anaesthesia*, vol. 116, no. 6, pp. 790–796, Apr. 2016.
[34] J. Poleszczuk, M. Debowska, W. Dabrowski, A. Wojcik-Zaluska, W. Zaluska, and J. Waniewski, "Subject-specific pulse wave propagation modeling: Towards enhancement of cardiovascular assessment methods," *PloS ONE*, vol. 13, no. 1, Jan. 2018, Art. no. e0190972.
[35] J. Joseph, P. M. Nabeel, M. I. Shah, and M. Sivaprakasam, "Arterial compliance probe for cuffless evaluation of carotid pulse pressure," *PloS ONE*, vol. 13, no. 8, Aug. 2018, Art. no. e0202480.

Zhili Hao received the B.S. and M.S. degrees in mechanical engineering from Shanghai Jiao Tong University, Shanghai, China, and the Ph.D. degree from the Department of Mechanical, Materials and Aerospace Engineering, University of Central Florida. She worked in industry for two years. From 2003 to 2006, she was a Post-doctoral Researcher on micro-sensors with the School of Electrical and Computer Engineering, Georgia Tech. In July 2006, she joined the Department of Mechanical and Aerospace Engineering, Old Dominion University, as an Assistant Professor and is currently an Associate Professor. Her current research interests include biomechanics, development of microfluidics tactile sensors and investigation of biomedical applications of these sensors, including arterial pulse signal measurements, tumor detection and differentiation, and mechanical characterization of soft biological tissues.

Dan Wang received the B.S. degree in electronic and information engineering from the Southwest University of Science and Technology, Mianyang, China, in 2010, the M.S. degree in control engineering from the University of Electronic Science and Technology of China, Chengdu, China, in 2013, and the Ph.D. degree from the Department of Mechanical and Aerospace Engineering, Old Dominion University in 2019. Her current research interests include design, fabrication, and testing of polymer-based microfluidic sensors, as well as applications of these sensors, such as arterial pulse measurement and signal analyzing.