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# Integrated optomechanical ultrasonic sensors with nano-Pascal-level sensitivity

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## Abstract

Ultrasonic sensors are widely used for object detection and localization in underwater and biological settings. The operational range and spatial resolution are inherently limited by sensor sensitivity, in which conventional piezoelectric transducers have been overwhelmed by advanced photonic sensors. Here, we demonstrate an optomechanical ultrasonic sensor integrated into a photonic platform, which comprises a suspended SiO<sub>2</sub> membrane embedded with a high-Q Si<sub>3</sub>N<sub>4</sub> microring resonator. By exploiting simultaneous optical and mechanical resonances, the sensor achieves a record low noise-equivalent pressure (NEP) of 218 nPa Hz<sup>-1/2</sup> at 289 kHz in air and 9.6 nPa Hz<sup>-1/2</sup> at 52 kHz in water. We demonstrate its versatility through photoacoustic gas spectroscopy in air and underwater ultrasound imaging, achieving a minimum detectable C<sub>2</sub>H<sub>2</sub> concentration of 2.9 ppm (integration time 1 s) and an imaging resolution of 1.89 mm, respectively. Our work represents a significant advancement in compact CMOS-compatible ultrasound sensing, unlocking new possibilities in biomedical imaging, environmental monitoring, industrial testing, and underwater communications.

## Introduction

Ultrasound sensing<sup>1</sup> has become a cornerstone of modern diagnostic and monitoring technologies, with crucial applications spanning medical imaging<sup>2,3</sup>, non-destructive testing<sup>4,5</sup>, underwater acoustics<sup>6</sup>, and industrial process monitoring<sup>7,8</sup>. Conventional piezoelectric transducers<sup>9–11</sup>, though widely used, face critical challenges in miniaturization, integration, and sensitivity, limiting their applications in scenarios that require compact designs or high resolutions<sup>12,13</sup>. These constraints are particularly evident in miniaturized systems such as portable medical devices<sup>14</sup> and intravascular ultrasound probes<sup>15,16</sup>, where sensor arrays are often restricted to sub-centimeter scale to meet spatial and operational requirements. Recent advances in micro-machined sensors, such as piezoelectric micromachined ultrasound transducers (PMUTs)<sup>17</sup> and capacitive micro-machined ultrasound transducers (CMUTs)<sup>18–20</sup>, have

improved array density and electronic integration. However, these devices remain susceptible to electromagnetic interference and exhibit inadequate sensitivities required for next-generation applications. For instance, their inability to operate close to high-voltage transmission lines hinders the monitoring of corona discharge noise<sup>21</sup>. Furthermore, their insufficient sensitivity poses significant challenges in applications like cranial ultrasound imaging<sup>22,23</sup>.

Optical microcavity-based ultrasonic sensors have emerged as a transformative alternative<sup>1,11,24</sup>, offering unprecedented sensitivity, immunity to electromagnetic interference, broad bandwidth, and chip-scale integration capability. Fiber-based microcavity sensors, including Fabry-Pérot cavities<sup>25–28</sup>, Bragg gratings<sup>16,29–31</sup>, and microsphere cavities<sup>32–34</sup>, have shown particular promise in endoscopy applications but face scalability challenges due to limited arraying capabilities and reliance on bulky optical components<sup>35</sup>. The growing demand for miniaturized, high-performance sensing systems has spurred significant research toward chip-scale sensing platforms<sup>36</sup>, where CMOS-compatible microring resonators show significant promise. Various microring resonator platforms, fabricated from polymer<sup>37–39</sup>, silicon<sup>40,41</sup>, and

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chalcogenide<sup>42</sup>, have been developed for ultrasound sensing. Despite their broad detection bandwidth, these integrated sensors achieved noise equivalent pressures (NEPs) only in the  $\text{mPa Hz}^{-1/2}$  range, primarily due to their modest optical quality ( $Q$ ) factors and restricted mechanical displacements caused by the thick substrates.

Optomechanical sensors offer a powerful approach for ultrasensitive ultrasound detection by harnessing the strong interaction between light and mechanical vibrations in high- $Q$  microresonators<sup>43</sup>. Suspended microdisk resonators have achieved unprecedented low NEPs at the  $\mu\text{Pa Hz}^{-1/2}$  level<sup>44–46</sup>. However, the reliance on fiber-taper coupling presents significant challenges in practical applications, including alignment complexity, reduced robustness, and difficulty working in aqueous environments. Here, we develop a novel integrated, mass-produced, ultrasensitive optomechanical ultrasonic sensor with nano-Pascal-level sensitivity and further demonstrate the sensor's versatility through photoacoustic gas spectroscopy and underwater ultrasound imaging. Beyond prototype demonstrations, this fully integrated, CMOS-compatible, highly sensitive ultrasonic sensor enables scalable production and paves the way for future multi-scene sensing applications, advancing medical diagnostics and industrial precision measurements.

## Results

### Principle

As depicted in Fig. 1a, the proposed integrated ultrasonic sensor comprises a suspended  $\text{SiO}_2$  membrane and

a high- $Q$   $\text{Si}_3\text{N}_4$  microring resonator. The  $\text{SiO}_2$  membrane is clamped at its periphery to the silicon substrate and suspended at its center to maximize mechanical displacement under ultrasound excitation. The  $\text{Si}_3\text{N}_4$  microring is embedded within the  $\text{SiO}_2$  membrane, enabling optomechanical readout by transducing the mechanical displacement of the membrane into optical resonance shifts of the ring resonator. To maximize the readout signal, we systematically optimize the geometries of the sensor, yielding final radii of  $450\ \mu\text{m}$  for the membrane and  $235\ \mu\text{m}$  for the microring resonator (see Materials and methods “Device design” Section). As shown in the right inset of Fig. 1a, by locking the frequency of an input continuous wave (CW) laser on the blue-detuned side of the optical resonance, the ultrasound-induced displacement is converted into the laser intensity modulation. The higher the optical  $Q$  factor, the steeper the slope of the transmission spectrum, and the greater the readout sensitivity.

Mechanical resonance further amplifies the displacement of the  $\text{SiO}_2$  membrane, significantly enhancing the sensitivity to acoustic signals. The bottom left inset of Fig. 1a compares the displacement response of a mechanical oscillator at different frequencies, illustrating the amplification of the mechanical resonance. Figure 1b presents a schematic comparison of integrated ultrasonic sensors with three configurations: (i) conventional non-suspended structure, suspended structure operating at (ii) off-mechanical and (iii) on-mechanical resonance frequencies, highlighting the mechanical resonance-enhanced

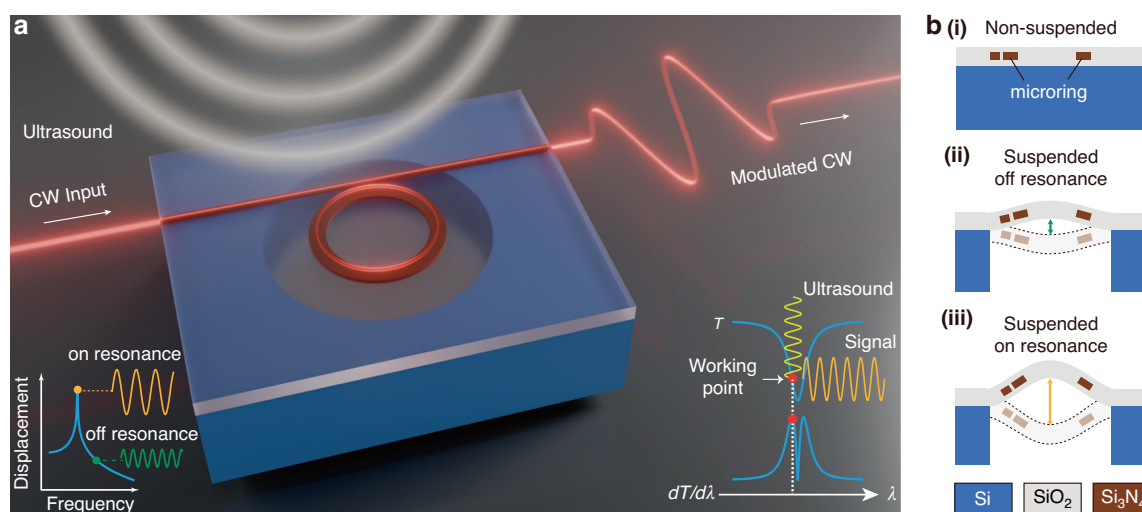


**Fig. 1** Schematic and working principle of the integrated ultrahigh-sensitivity optomechanical ultrasonic sensor. **a** Schematic of the ultrasonic sensor, where the intensity of the continuous wave (CW) input laser is modulated by the ultrasound. Left inset: Schematic of the mechanical resonance-enhanced displacement response. Right inset: Optical readout principle of the sensor, illustrating the optical resonance-enhanced sensitivity. **b** Schematic comparison of integrated microring resonator-based ultrasonic sensors with three configurations: (i) non-suspended structure, (ii) suspended structure (off mechanical resonance), and (iii) suspended structure (on mechanical resonance), illustrating the enhanced response of the suspended structure with mechanical resonance

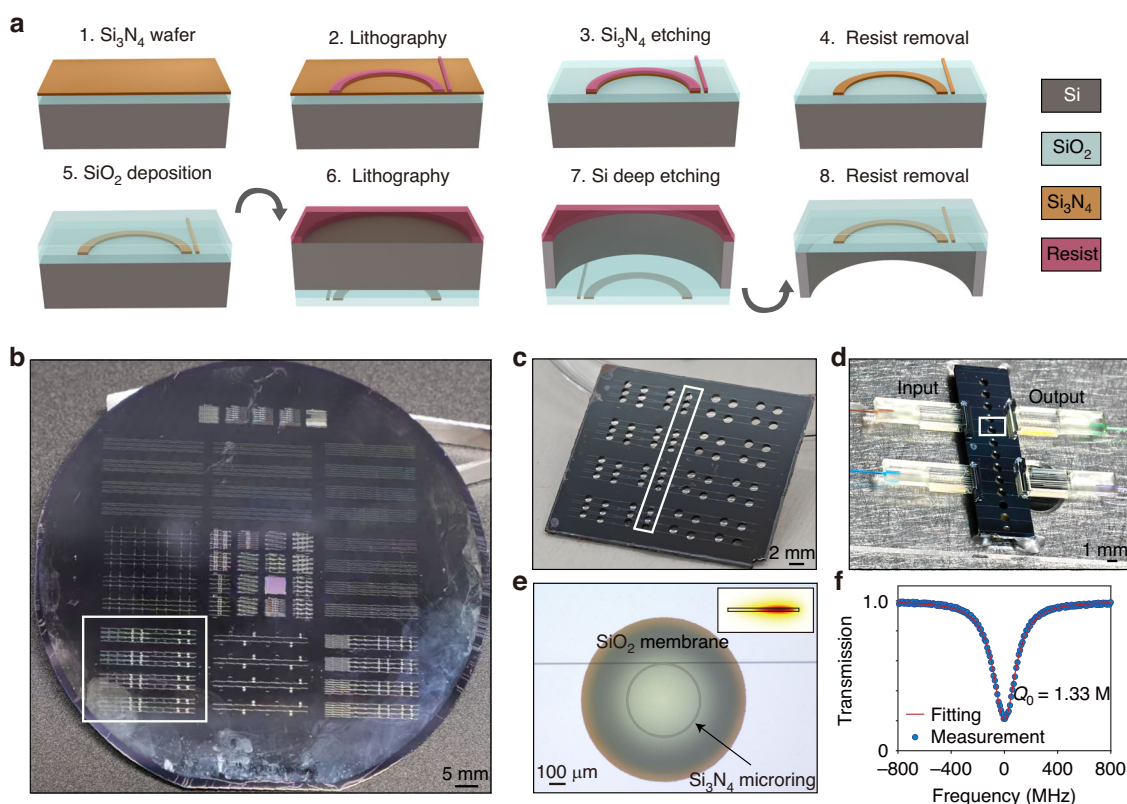


**Fig. 2** Sensor fabrication and characterization. **a** Fabrication process of the sensor. **b** Photograph of the processed 4-inch wafer. **c** Photograph of a  $3 \times 3 \text{ cm}^2$  chip after Si deep etching. **d** Photograph of the packaged sensor. **e** Optical microscopy image of the sensor, showing a  $\text{SiO}_2$  membrane with an embedded  $\text{Si}_3\text{N}_4$  microring resonator. Inset: Optical mode profile  $|E|$ , with  $E$  denoting the electric field. **f** Normalized transmission spectrum of the microring resonator measured in water

sensor response. It is worth noting that the sensor's response to ultrasonic waves is dominated by optical mode shifts caused by variations in the microring radius. Specifically, the contributions from the changes in the coupling gap or refractive index variations induced by photoelastic effect are negligible, as detailed in Supplementary Information Section IB.

### Sensor design and fabrication

We fabricate the devices at the wafer scale, with the process flow outlined in Fig. 2a. The fabrication begins with electron beam lithography (EBL) and reactive ion etching (RIE) to create the  $\text{Si}_3\text{N}_4$  microring resonator<sup>47</sup>, with both processes optimized to achieve near-vertical sidewalls. A  $\text{SiO}_2$  cladding layer is deposited on the  $\text{Si}_3\text{N}_4$  layer to preserve resonator optical performance, with Fig. 2b showing a photograph of a full 4-inch wafer after the deposition. We then flip the wafer upside down and perform photolithography and deep reactive ion etching (DRIE) processes to selectively etch the silicon underneath  $\text{Si}_3\text{N}_4$  microring resonators. Figure 2c presents a photograph of a  $3 \times 3 \text{ cm}^2$  chip after releasing the suspended  $\text{SiO}_2$  membranes. For full integration and

enhanced robustness, two mode conversion fibers are used to couple light into the microring resonator and are co-packaged with the device (Fig. 2d). This results in a compact and portable ultrasonic sensor with reliable operation in diverse environments.

### Device characterization

Figure 2e shows an optical microscopy image of the sensor, with the optical field distribution shown in the inset. The microring resonator exhibits an optical Q factor of  $1.33 \times 10^6$  in water (Fig. 2f), which is similar to that of  $1.35 \times 10^6$  in air (Fig. 7e), confirming that the packaging robustly preserves the microring optical performance in various operational environments. The high optical Q factor also ensures the sensor has sufficiently high readout sensitivity, enabling thermal noise to dominate around mechanical resonance frequencies.

The experimental set-up for sensitivity characterization is depicted in Materials and methods "Ultrasound sensitivity characterization" Section. Specifically, a 1550 nm laser is coupled into the microring resonator and detuned to the optimal slope of the optical resonance. The probe laser is locked to the blue-detuned side of the optical mode using a

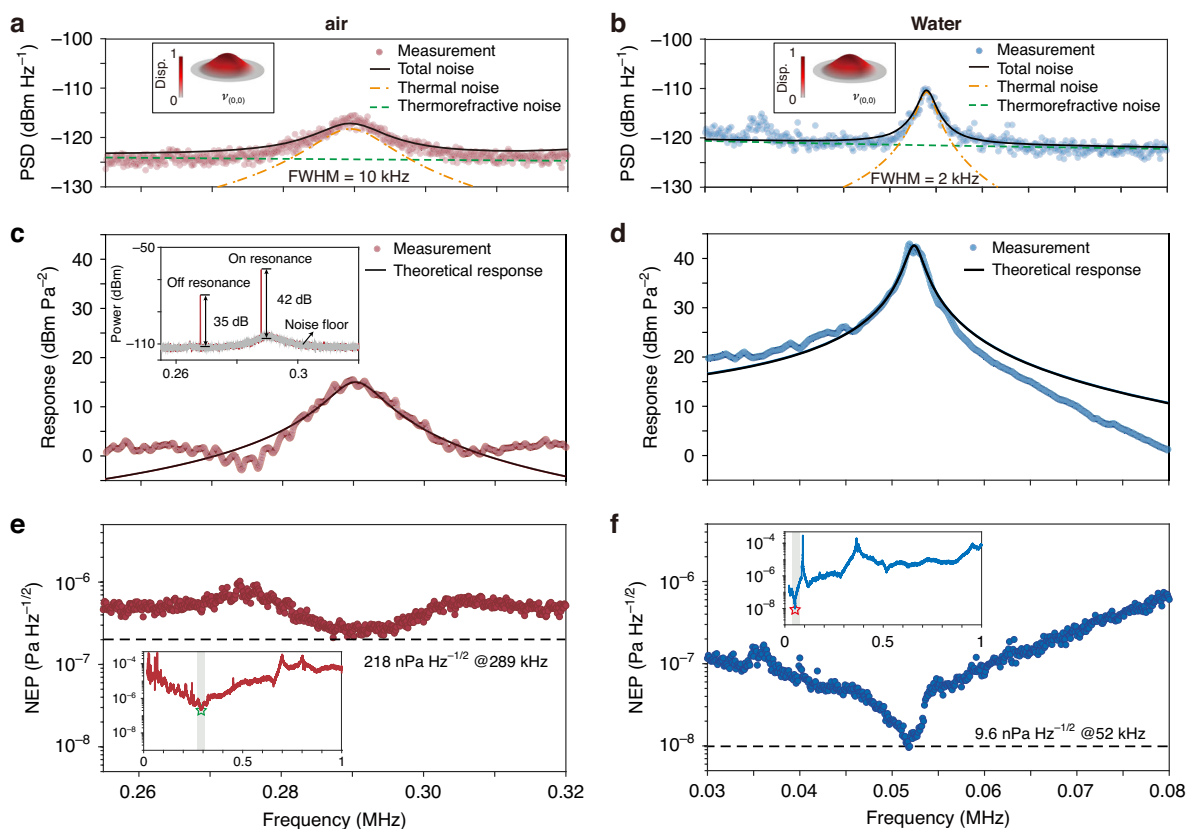


**Fig. 3 Characterization of the ultrasonic sensor.** Noise power spectral densities (PSDs) of the  $v_{(0,0)}$  mode in air (a) and water (b), with insets showing the  $v_{(0,0)}$  mode profile of the  $\text{SiO}_2$  membrane. FWHM full width at half maximum. Ultrasound response of the sensor at different frequencies in air (c) and water (d). Inset in (c): PSD of the sensor in air when the applied ultrasound frequency is on resonance (288 kHz) and off resonance (268 kHz) with the  $v_{(0,0)}$  mechanical mode, with the gray curve representing the noise floor. Noise-equivalent pressure (NEP) spectral densities around the  $v_{(0,0)}$  mode in air (e) and water (f). The achieved minimum NEPs are  $218 \text{ nPa Hz}^{-1/2}$  at 289 kHz in air and  $9.6 \text{ nPa Hz}^{-1/2}$  at 52 kHz in water, respectively. Insets: NEP spectral densities between 20 kHz and 1 MHz in air (e) and water (f)

proportional-integral-derivative (PID) controller. The stability of the device is discussed in Supplementary Information Section IIC. A pre-calibrated ultrasound transducer, driven by a vector network analyzer (VNA) and positioned above the sensor, serves as the ultrasound source. The ultrasound-induced optical resonance shift modulates the transmitted light intensity and is detected by a photo-detector. The transmission spectrum of the optical mode, noise power spectral density (PSD), and the ultrasound response of the sensor are measured by an oscilloscope, electronic spectrum analyzer (ESA), and VNA, respectively.

Noise PSD across the first-order flapping mode  $v_{(0,0)}$  is measured in both air and water, as shown in the red and blue dots in Fig. 3a, b, respectively. The orange dash-dotted, green dashed, and black solid curves represent the calculated thermal noise, thermorefractive noise (TRN), and total noise, respectively. Notably,  $v_{(0,0)}$  mode (with its mechanical profile shown in the insets of Fig. 3a, b) exhibits the greatest spatial overlap with incident ultrasound from above the sensor and therefore is expected to

have better sensitivity. Near the mechanical resonance frequency, thermal noise—arising from both surrounding molecule collisions and intrinsic damping of the mechanical resonator<sup>44</sup>—dominates as Lorentzian peaks (orange dashed curves). Resonance frequencies of the  $v_{(0,0)}$  mode in air and water are 289 kHz and 52 kHz with linewidths of 10 kHz and 2 kHz, respectively. The resonance frequency shifts downward in water compared to air due to the increased effective mass and viscosity<sup>48</sup>. Away from the mechanical resonance frequency, TRN dominates (green dashed curves), originating from temperature fluctuations induced refractive index variation via the thermo-optic effect<sup>49</sup>. TRN can be mitigated through using cavity materials with a smaller thermo-optic coefficient, increasing the cavity mode volume, or improving the thermal stability of the device, with details provided in Supplementary Information Section IA. As investigated in previous works, achieving thermal noise-limited performance is essential for enhanced sensitivity<sup>44</sup> (Supplementary Information Section IA).



The sensor's single-frequency linear dynamic range (LDR) at mechanical resonance is approximately 52 dB in both air and water (Supplementary Information Section ID), and all subsequent ultrasound pressures are maintained within this linear range. The ultrasound response of the sensor is measured in air and water by sweeping the applied frequency to the transducer using the VNA, with the results shown in red and blue dots in Fig. 3c, d, respectively. The observed spectral peak matches the mechanical mode identified in Fig. 3a and b, confirming that the mechanical mode amplifies the membrane's displacement response. As illustrated in the inset of Fig. 3c, the on-resonance case achieves a signal-to-noise ratio (SNR) of 42 dB at 30 Hz resolution bandwidth (RBW), a 7 dB improvement over the off-resonance case (35 dB). Deviations of responses in the off-resonance frequency regions (red and blue dots in Fig. 3c, d) from the theoretical mechanical response curves using mechanical susceptibility functions (black curves in Fig. 3c, d) result from interferences with other mechanical modes near the  $\nu_{(0,0)}$  mode (see Supplementary Information Section IC for details). Additionally, the observed response undulations are caused by sound waves reflections between the ultrasound transducer and the sensor chip (Supplementary Information Section IIB).

The NEPs of the sensor at different frequencies are derived from the noise spectra, single-frequency response, and response spectra, as shown in Fig. 3e (air) and 3f (water). The NEP spectra exhibit minima at mechanical resonance frequencies, consistent with theoretical predictions (Supplementary Information Sections IA and IC). Remarkably, our sensor achieves optimal NEPs of  $218 \text{ nPa Hz}^{-1/2}$  at 289 kHz in air and  $9.6 \text{ nPa Hz}^{-1/2}$  at 52 kHz in water, representing a record in microcavity-based ultrasonic sensors. The superior sensitivity in water compared to that in air likely arises from improved acoustic impedance match at the water-sensor interface. We also measured the NEPs in a larger frequency range from 20 kHz to 1 MHz, displayed in the insets of Fig. 3e, f, where the shaded areas correspond to the  $\nu_{(0,0)}$  mode, and the pentagrams mark the best sensitivities. Device reproducibility is confirmed across nine sensors from the same batch, showing 1% variation in mechanical resonance frequencies and 38% deviations in sensitivities (Supplementary Information Section IIA). Furthermore, the  $\nu_{(0,0)}$  mode demonstrates excellent directional uniformity, with a 3 dB angular bandwidth exceeding  $100^\circ$  (Supplementary Information Section IIB). Crucially, suspended-membrane sensors outperform non-suspended (without DRIE process) devices by two orders of magnitude in NEP (Fig. 9), highlighting the critical role of membrane suspension.

### Photoacoustic gas spectroscopy

Highly sensitive air-coupled ultrasonic sensors are indispensable for gas detection in applications ranging

from breath analysis and environmental monitoring to hazardous gas detection<sup>50</sup>. Among spectroscopic techniques, photoacoustic spectroscopy (PAS) has emerged as a powerful approach<sup>51,52</sup>, leveraging the photoacoustic effect where gas molecules convert intensity-modulated light into acoustic waves through periodic thermal expansion. PAS offers unique advantages, including background-free detection, high sensitivity, and broad spectral coverage<sup>53</sup>, making it particularly suitable for trace gas analysis.

We demonstrate the application of our highly sensitive integrated ultrasonic sensor in PAS of  $\text{C}_2\text{H}_2$  gas, with the experimental setup shown in Fig. 4a (details in Materials and methods "Experimental details for photoacoustic gas spectroscopy" Section). Specifically, a pump laser with its intensity modulated by an electro-optic modulator (EOM) is injected into the gas cell containing a mixture of 1%  $\text{C}_2\text{H}_2$  and 99%  $\text{N}_2$ , while the integrated ultrasonic sensor within the gas cell detects the generated acoustic waves. PA signals are measured at two modulation frequencies: 288 kHz (on mechanical resonance) and 268 kHz (off resonance), at different pump powers. As the PA signal scales linearly with pump power<sup>54</sup>, the noise-equivalent concentration (NEC) is inversely proportional to the pump power (Fig. 4b). At 350 mW pump power and 1 s integration time, NECs of 2.9 ppm (on resonance) and 13.1 ppm (off resonance) are achieved, demonstrating significant sensitivity enhancement through mechanical resonance. The inset of Fig. 4b quantifies this improvement, showing SNRs of 61 dB at 288 kHz (on resonance) versus 48 dB at 268 kHz (off resonance). The background-free nature of PAS is confirmed by comparing PA signals from  $\text{C}_2\text{H}_2/\text{N}_2$  mixture and pure  $\text{N}_2$  (Supplementary Information Section IIIA), with PA responses exclusively emerging at  $\text{C}_2\text{H}_2$  absorption wavelengths. The PA spectrum of our sensor is obtained by scanning the pump laser wavelength with a step of 0.01 nm, as shown in the blue curve of Fig. 4c, which matches well with the high-resolution transmission molecular absorption database (HITRAN) simulations (red curve). The relative residuals between the two are below 5% (black curve), validating the system's spectroscopic accuracy.

### Underwater ultrasound imaging

Ultrasound imaging technology is a powerful non-destructive technique widely used in medical diagnostics, industrial inspection, and marine acoustics. We demonstrate these capabilities through underwater ultrasound imaging using our highly sensitive sensors. As depicted in Fig. 4d, the experimental setup includes a water tank containing a 500 kHz focused ultrasound transducer, our sensor, and an acrylic sample featuring an "F" shaped groove. When immersed in water, the groove traps air,



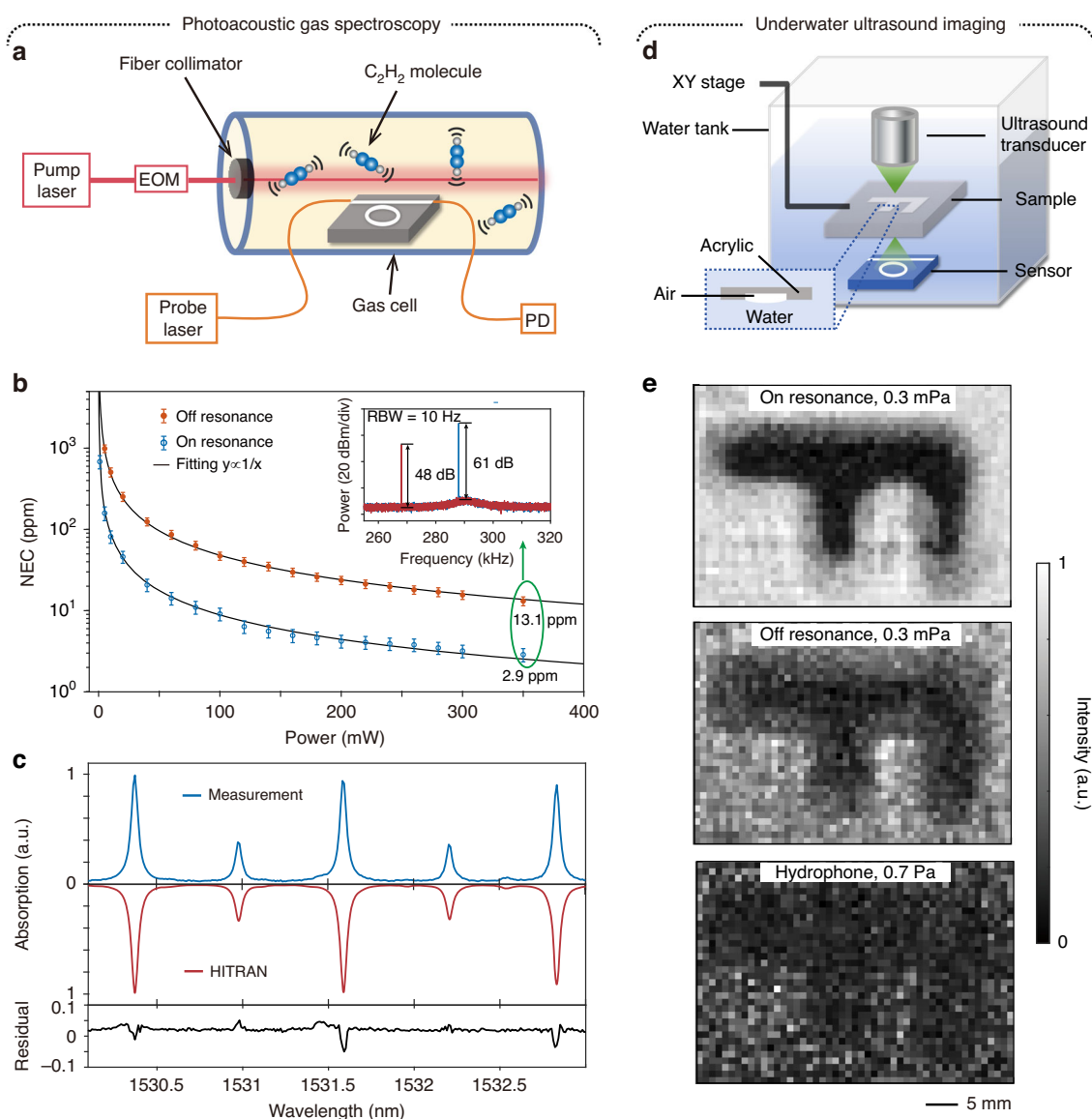


**Fig. 4** Demonstrations of photoacoustic gas spectroscopy and underwater ultrasound imaging. **a** Experimental setup for photoacoustic detection of gaseous acetylene ( $C_2H_2$ ). EOM electro-optic modulator, PD photodetector. **b** Noise-equivalent concentration (NEC) as a function of pump power. Inset: Photoacoustic signal for 1%  $C_2H_2$  mixed with 99%  $N_2$  at 350 mW pump power. **c** Measured (blue) and HITRAN-simulated (red) absorption spectra of  $C_2H_2$  from 1530 to 1533 nm. Bottom panel: Residual between experimental and simulated spectra. **d** Schematic of the underwater ultrasound transmission imaging system. Inset: Cross-sectional view of the sample structure. **e** (i, ii) Ultrasound images at 0.3 mPa pressure with an ultrasound frequency of (i) 517 kHz (on mechanical resonance) and (ii) 550 kHz (off mechanical resonance). (iii) Ultrasound image from a commercial hydrophone, with an ultrasound frequency of 517 kHz and pressure of 0.7 Pa

creating a localized acoustic impedance mismatch. The sample is raster-scanned using a two-dimensional horizontal translational stage. When the ultrasound beam generated by the ultrasound transducer is focused on the “F” region, the air-filled groove reflects the ultrasound waves, resulting in significant attenuation of the transmitted ultrasound signal. This generates a strong contrast between the groove and the surrounding homogeneous regions, enabling high-resolution imaging. For optimal

performance, we operate the sensor at the  $\nu_{(3,0)}$  mechanical mode of 517 kHz (mode profile shown in the inset of Fig. 10b), achieving higher spatial resolution than the  $\nu_{(1,0)}$  due to its shorter wavelength.

The sample is scanned with a 1 mm step size, and the sensor’s response is recorded at each position. Figure 4e(i) and (ii) represent the imaging results obtained with a driving ultrasound pressure of 0.3 mPa and driving frequencies of 517 kHz (on-resonance) and 550 kHz



(off-resonance), respectively, revealing significantly improved contrast and clarity under resonant conditions, a direct consequence of mechanical resonance-enhanced sensitivity. Further quantitative analysis (Supplementary Information Section IIIB) confirms a spatial resolution of 1.89 mm, demonstrating the potential of our sensor for high-resolution imaging and long-range target detection in underwater applications. This spatial resolution is primarily determined by the focusing geometry of the external ultrasound transducer, instead of by the sensor itself.

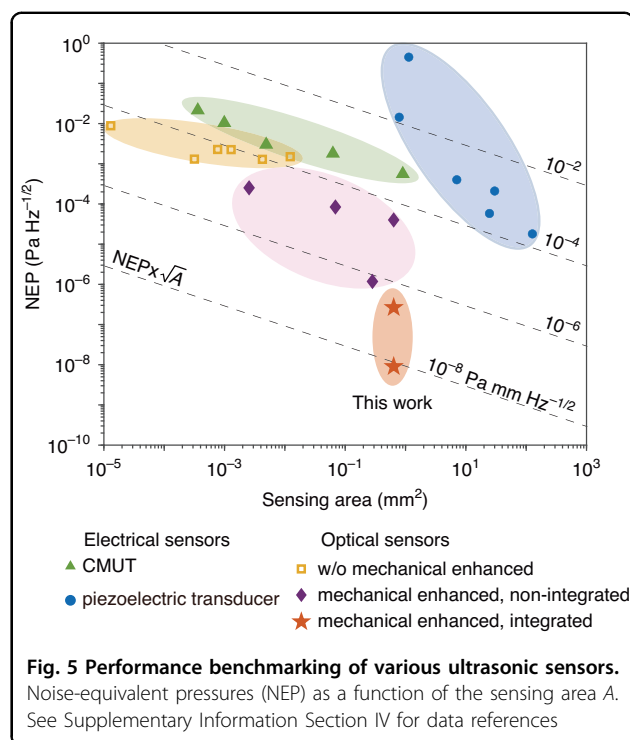
To provide a comprehensive performance benchmark, we include a comparison with a commercial hydrophone. Remarkably, our sensor outperforms a commercial hydrophone even at ultrasound pressures three orders of magnitude lower (0.3 mPa versus 0.7 Pa), as shown in Fig. 4e(iii). This demonstrates the sensor's imaging capability under extremely weak ultrasound signals. Notably, conventional hydrophones are engineered for fundamentally different operational regimes, typically prioritizing wide bandwidth and high-pressure detection capability. This comparison underscores their complementary nature, with our sensor excelling in scenarios requiring ultra-high sensitivity within specific frequency bands while conventional hydrophones remain preferred for wide bandwidth applications at higher ultrasound pressure levels.

While only transmissive imaging is demonstrated here, the compact size of the sensor facilitates easy integration with an ultrasound transmitter to achieve reflective measurement. Furthermore, adding an impedance-matching layer (such as polymers) on the membrane can further enhance the device's robustness for contact detection.

## Discussion

Figure 5 compares the NEP versus sensing area  $A$  for various ultrasonic sensors. Our sensor achieves the highest sensitivity (lowest NEP) while maintaining a compact footprint. As the NEP of optomechanical sensors scales approximately as  $1/\sqrt{A}$ , we evaluate the figure of merit  $\text{NEP} \times \sqrt{A}$  across different technologies. Remarkably, our sensor achieves  $\text{NEP} \times \sqrt{A}$  values at the  $10^{-8} \text{ Pa mm Hz}^{-1/2}$  level, surpassing conventional electric and optical sensors by several orders of magnitude, highlighting its compelling advantages in both miniaturization and sensitivity. A comprehensive performance comparison with existing ultrasonic sensors is provided in Supplementary Information Section IV. Table S2 demonstrates more comprehensively that, unlike other sensors with wide bandwidth and high operating ultrasound pressure levels, our sensor possesses a significant advantage in detecting extremely weak acoustic signals.

Several key directions will drive further development of integrated ultrasound sensing technology. First, the



sensitivity of the sensor can be further enhanced by geometric optimization. The working bandwidth can be significantly expanded by suppressing TRN to enable thermal-noise-limited sensitivity across wider frequency ranges and utilizing multiple mechanical modes for broader spectral coverage. Furthermore, external feedback control can be applied to the mechanical mode to achieve mode cooling<sup>55,56</sup>. This strategy can effectively reduce the mechanical quality factor, thereby extending the frequency range where thermal noise dominates while maintaining the sensitivity<sup>57</sup>.

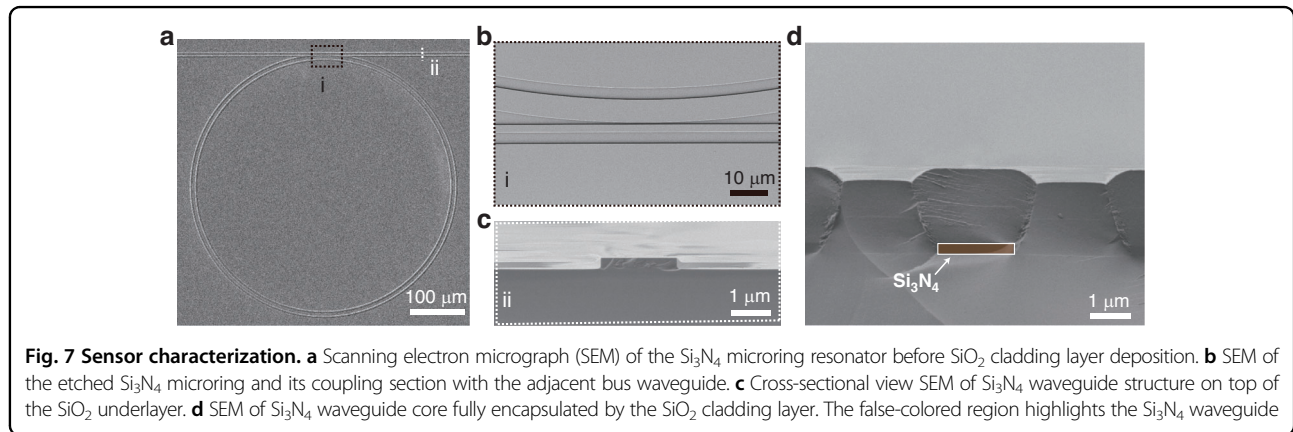
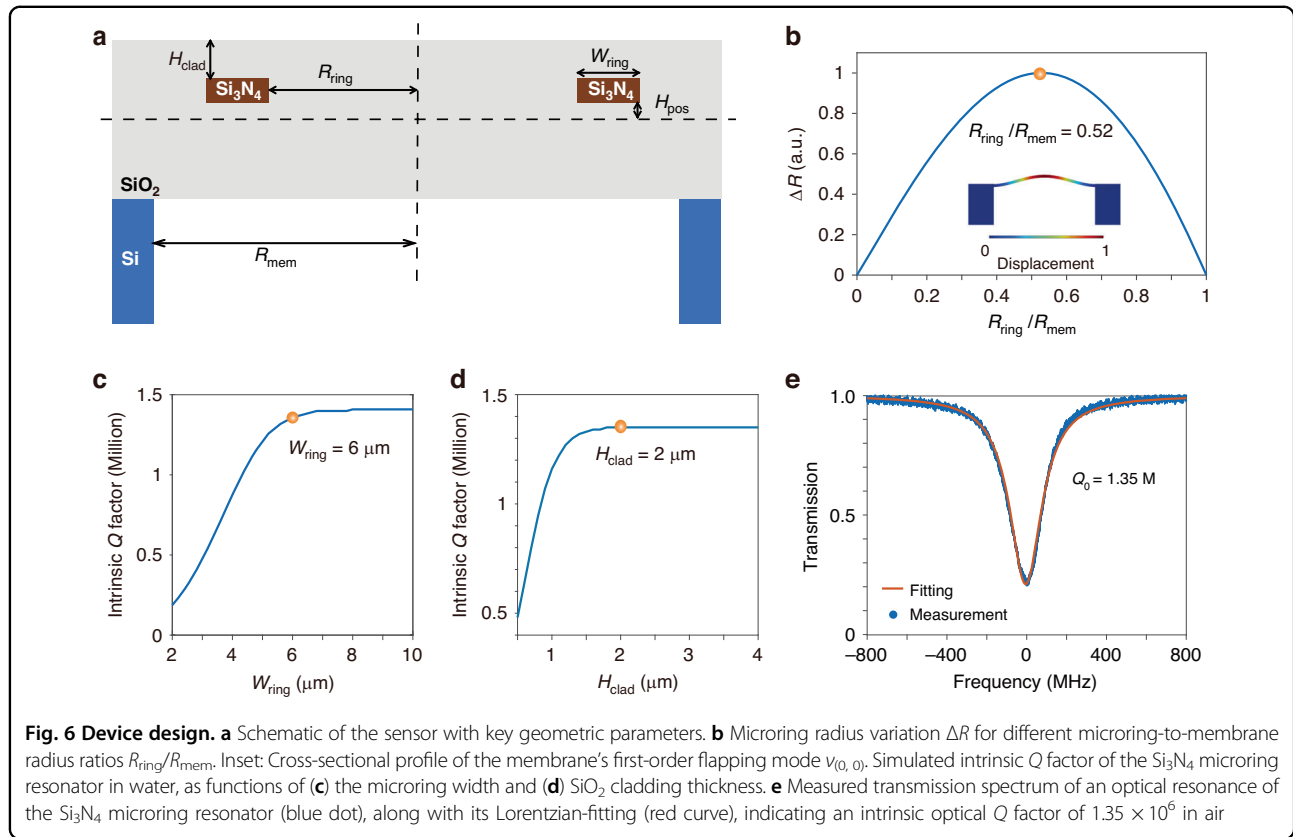
Additionally, advances in photonic integrated circuits<sup>58–60</sup> enables co-integration of complementary on-chip components, including lasers<sup>61,62</sup>, spectrometers<sup>63</sup>, and photodetectors<sup>64</sup>, facilitating fully integrated sensor systems<sup>65</sup>. Beyond device optimization, we envision practical applications of these highly sensitive sensors in multimodal sensing and wearable technologies, transitioning these sensitive platforms from laboratory prototypes to field-deployable solutions for mobile sensing<sup>66</sup>. Our work establishes a foundation for next-generation compact, high-performance sensing systems with transformative potential across diverse domains, from biomedical diagnostics and industrial monitoring to real-time environmental surveillance.

## Materials and methods

### Device design

We systematically optimize the device geometry to enhance ultrasound sensitivity (Fig. 6a). The key design





parameter is the radius ratio  $R_{\text{ring}}/R_{\text{mem}}$ , where  $R_{\text{ring}}$  and  $R_{\text{mem}}$  denote the radii of the microring and membrane, respectively. Our optimization focuses on the first-order flapping mode  $v_{(0,0)}$ , as it has the largest spatial overlap with the incident ultrasound from above the sensor. The mechanical mode profile of  $v_{(0,0)}$  is derived by modeling the  $\text{SiO}_2$  membrane as a thin plate<sup>67</sup>, yielding the normalized displacement:

$$w_{0,0} = J_0\left(\lambda_{00} \frac{R_{\text{ring}}}{R_{\text{mem}}}\right) - \frac{J_0(\lambda_{00})}{I_0(\lambda_{00})} I_0\left(\lambda_{00} \frac{R_{\text{ring}}}{R_{\text{mem}}}\right) \quad (1)$$

where  $J_0$  and  $I_0$  represent the 0-th Bessel function and the modified Bessel function of the first kind, respectively.  $\lambda_{00}$  is a dimensionless parameter which is related to the mode shape, determined from ref.<sup>68</sup>. The normalized radial change  $\Delta R$  is calculated as<sup>69</sup>:

$$\Delta R = -H_{\text{pos}} \frac{\partial w_{0,0}}{\partial R_{\text{ring}}} \quad (2)$$

where  $H_{\text{pos}}$  is the vertical position of the waveguide. Figure 6b shows  $\Delta R$  versus  $R_{\text{ring}}/R_{\text{mem}}$ , revealing a

maximum at  $R_{\text{ring}}/R_{\text{mem}} = 0.52$ . Considering an EBL write field of  $500\ \mu\text{m}$ , we implement  $R_{\text{ring}} = 235\ \mu\text{m}$  and  $R_{\text{mem}} = 450\ \mu\text{m}$ .

We systematically investigate the intrinsic  $Q$  factor of a  $\text{Si}_3\text{N}_4$  microring resonator with a radius of  $235\ \mu\text{m}$  and a thickness of  $265\ \text{nm}$ , as a function of microring width and  $\text{SiO}_2$  cladding thickness (Fig. 6c, d). The intrinsic  $Q$  factor exhibits strong geometric dependence, showing a sharp decline for microring width below  $6\ \mu\text{m}$  (Fig. 6c) while remaining high for cladding thickness above  $2\ \mu\text{m}$  (Fig. 6d). While thicker cladding enhances optical  $Q$  factor, it comprises mechanical compliance and thus decreases sensitivity. We therefore optimize the design with  $6\ \mu\text{m}$  microring width and  $2\ \mu\text{m}$  cladding thickness, achieving both intrinsic  $Q$  factors ( $1.35 \times 10^6$  in air, Fig. 6e) and environmental stability without sacrificing sensitivity.

### Device fabrication

The ultrasonic sensors are fabricated on a 4-inch wafer comprising a  $265\ \text{nm}$ -thick  $\text{Si}_3\text{N}_4$  layer on a  $4\ \mu\text{m}$ -thick buried-oxide layer above a  $500\ \mu\text{m}$ -thick silicon substrate. The fabrication begins with patterning  $\text{Si}_3\text{N}_4$  microring resonators and bus waveguides using EBL with AR-P 6200 resist, followed by reactive ion etching (RIE) employing  $\text{CHF}_3/\text{O}_2$  plasma. A protective  $2\text{-}\mu\text{m}$ -thick  $\text{SiO}_2$  cladding layer is subsequently deposited via inductively coupled plasma chemical vapor deposition (ICP-CVD) to shield the microring resonator from environmental contamination (Fig. 7). To ensure precise concentric alignment between the  $\text{Si}_3\text{N}_4$  microring and  $\text{SiO}_2$  membrane structures, we perform double-sided aligned photolithography after flipping the wafer upside down. The  $\text{SiO}_2$  membrane release is achieved through a Bosch process utilizing alternating  $\text{C}_4\text{F}_8$  passivation and  $\text{SF}_6$  etching cycles to remove the underlying silicon substrate. Finally, the completed devices are diced and packaged with mode conversion fibers to create compact, portable sensor units.

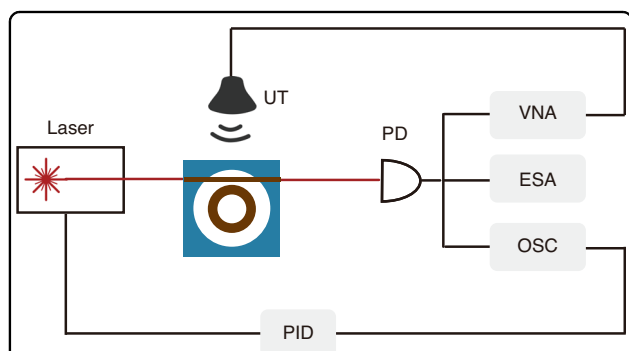


**Fig. 8 Experimental setup for ultrasound sensitivity measurements.** UT ultrasound transducer, PD photodetector, OSC oscilloscope, ESA electronic spectrum analyzer, VNA vector network analyzer, PID proportional-integral-derivative

### Ultrasound sensitivity characterization

The experimental setup for ultrasound sensitivity measurement is shown in Fig. 8. A  $1550\ \text{nm}$  laser (TOPTICA CTL1500) is coupled into the microring resonator, with its frequency detuned relative to the optical mode to maintain 25% transmission power for optimal sensitivity<sup>70</sup>. The transmitted optical signal is detected using a photodetector (KEYANG KY-PRM-10M-1-FC) and recorded via an oscilloscope. A PID controller (Moku:Pro) is used to lock the probe laser frequency on the blue-detuned side of the optical mode.

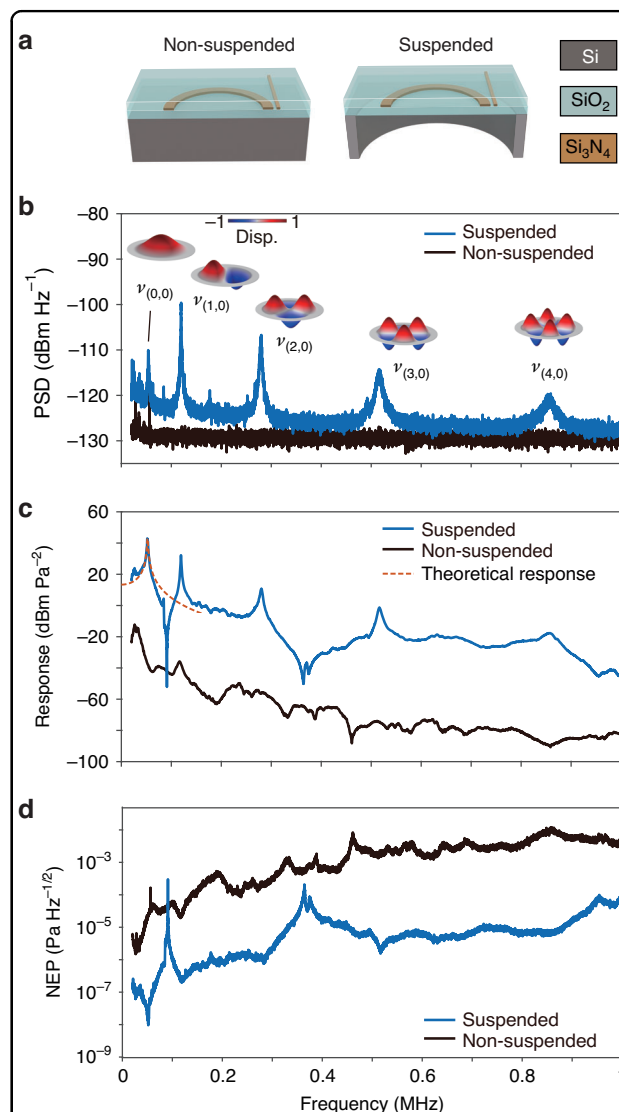
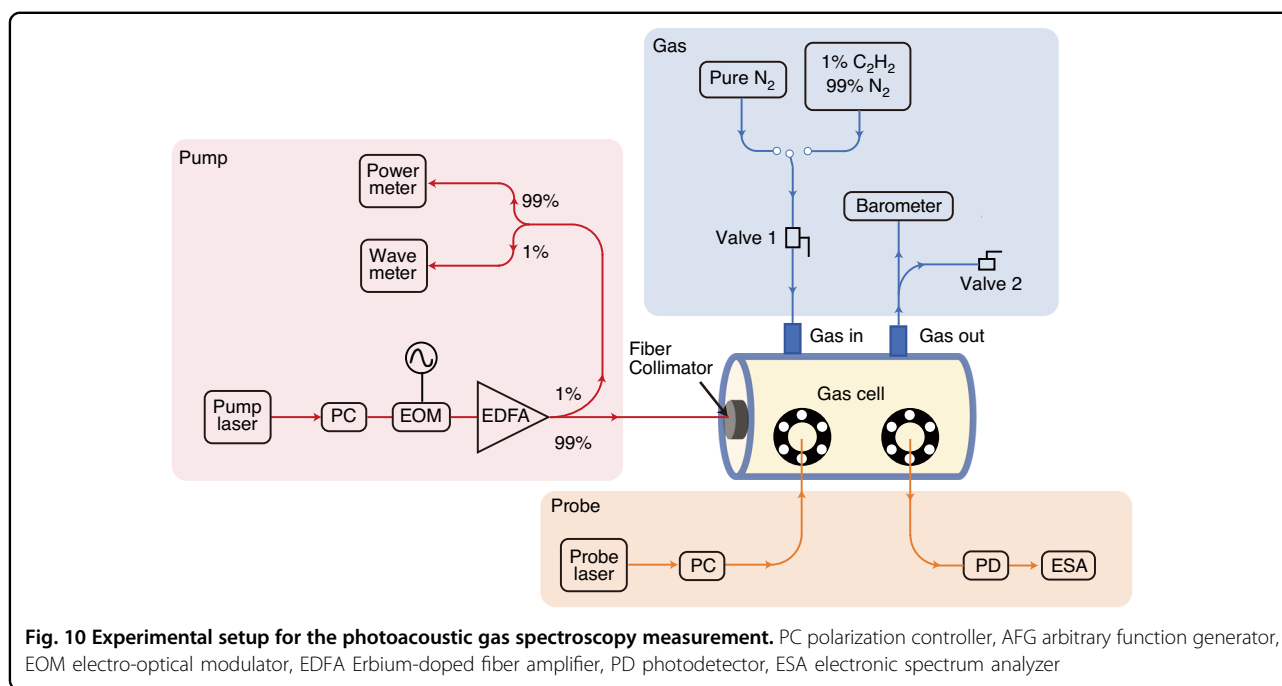


**Fig. 9 Performance comparison of non-suspended and suspended sensor structures in water.** **a** Schematic illustrations of the non-suspended (left) and suspended (right) sensor configurations. Comparative results of noise power spectra densities (PSDs) (**b**), frequency-dependent ultrasound responses (**c**), and noise-equivalent pressures (NEPs) (**d**) of the sensors with both non-suspended (black curves) and suspended (blue curves) structures



This allows the probe laser to follow the optical resonance frequency during the measurement. Ultrasound waves are generated by a pre-calibrated ultrasound transducer<sup>45</sup>, positioned above the sensor. Ultrasound pressure calibration at the sensor location is performed using: (i) a needle hydrophone (Onda HNR-1000) for aqueous measurements, and (ii) a scanning laser vibrometer (Sunnyinnovation optical Intelligence LV-SC400) for measurements in air. For single-frequency response characterization, the ultrasound transducer is driven by a sinusoidal signal generated by an arbitrary function generator, with the sensor response recorded using an electrical spectrum analyzer (ROHDE&SCHWARZ FPS4). Broadband frequency response of the sensor is measured via a vector network analyzer (ROHDE&SCHWARZ ZNL3), which sweeps ultrasound frequencies while recording the corresponding sensor response. The same experimental setup is employed for measuring non-suspended devices (Fig. 9).

#### Experimental details for photoacoustic gas spectroscopy

Photoacoustic gas spectroscopy measurements are performed at room temperature and atmospheric pressure using a sealed gas cell with fiber-optic feedthroughs for optical access (Fig. 10). The cell features a gas inlet port connected to acetylene ( $C_2H_2$ ) and nitrogen ( $N_2$ ) cylinders for precise gas mixing, and an outlet port connected with a built-in barometer for pressure monitoring. Before measurements, the gas cell is purged with 1%  $C_2H_2$  mixed with 99%  $N_2$ , by opening the  $C_2H_2$  supply valve (valve 1) and vent valve (valve 2) to ensure complete air

displacement and concentration accuracy. During measurements, both valves are closed to maintain a static gas environment, with identical procedures followed for  $N_2$  measurements. A CW pump laser (TOPTICA CTL1550) is used to excite the PA signals of the gas, with its intensity modulated by an electro-optic modulator (Exail MX-LN-10). The modulation frequency is controlled by a function generator, while the modulation depth is adjusted via a DC voltage. Before entering the gas cell, the pump power is amplified by an erbium-doped fiber amplifier (Amonics AEDFA-37-R-FA) and converted to a free-space beam with a 1–2 mm diameter spot using a fiber collimator. The integrated ultrasonic sensor, positioned below the optical path, detects the resulting PA signal of the gas.

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Experiments were conceived and designed by X.C., H.Y., M.W., B.-B.L., and Q.-F.Y. Measurements and data analysis were performed by X.C., H.Y., and M.W., with assistance from Z.-G.H. and Z.-L. W. Devices were designed by X.C., H.Y., and M.W. Devices were fabricated by X.C., H.Y., and M.W., with assistance from Y.W. and C.L. The project was supervised by B.-B.L. and Q.-F.Y. All authors participated in preparing the manuscript.

#### Data availability

The data that support the findings of this study are available from the corresponding author upon request.

#### Conflict of interest

The authors declare no competing interests.

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