Silicon coupled-resonator optical-waveguide-based biosensors using light-scattering pattern recognition with pixelized mode-field-intensity distributions

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Due to the increasing demand of healthcare, various chip-scale, label-free optical biochemical sensing technologies have been proposed and studied over the past two decades\textsuperscript{1-5}. Specifically, optical microresonator-based biochemical sensors have been attracting significant attention over the past decade. Conventional biochemical sensing techniques using optical microresonators typically employ two ways to quantitatively derive real-time information of the analyte on the microresonator surface. One is to monitor ultra-high-quality (ultra-high-Q) cavity resonance wavelength shifts in the transmission spectrum through scanning the input laser wavelength in the proximity of the resonance\textsuperscript{6-14}. The other is to monitor the transmission intensity change around a cavity resonance at a fixed wavelength\textsuperscript{7}. However, both approaches working in the spectral domain typically require a precision spectrum scanning system such as a wavelength-tunable diode laser.

Previously, our research group has proposed an alternative microresonator-based biochemical sensing scheme working in the spatial domain by using a coupled-resonator optical waveguide (CROW) excited at a fixed wavelength and monitoring the analyte-induced discrete modulations of the pixelized light-scattering intensity patterns among the CROW eigenstates\textsuperscript{15,16}. Such a sensing scheme only requires a relatively simple optical readout system including a fixed-wavelength laser and a camera. The simultaneous imaging of the spatially distributed coupled microresonators allows such a scheme to be more immune to the equipment noise that equally affects each microresonator but does not change the relative intensity distribution. Other researchers have also recently studied CROWs through imaging the out-of-plane elastic light scattering intensity patterns in the far field\textsuperscript{17-19}. Nonetheless, our initial proposal fell short in measuring only discrete modulations in refractive index applied on the CROW surface, and only considered an ideal CROW structure.
Here, we propose and demonstrate as a proof of concept an improved CROW-based biochemical sensing scheme working in the spatial domain using albeit imperfect coupled microring resonators on the silicon-on-insulator (SOI) platform. The choice of the SOI platform in 1550 nm wavelengths is primarily motivated by the maturing complementary metal-oxide-semiconductor (CMOS)-compatible SOI technology available to silicon photonics. We devise a correlation analysis of the pixelized mode-field-intensity distributions to extract from a library of calibrated correlation coefficients a refractive index change, $\Delta n$, applied upon the CROW surface from a known cladding refractive index, $n_0$. We model the CROW sensor assuming an imperfect CROW with fabrication-imperfection-induced randomly disordered coupled microresonators. Our experiments using a SOI 8-microring CROW in 1550 nm wavelengths reveal a $\Delta n$ of $\sim 1.5 \times 10^{-4}$ refractive index unit (RIU). Upon a specific probe wavelength, we demonstrate a sensitivity in terms of correlation coefficient change per unit RIU of $\sim 752$ RIU$^{-1}$ and a noise-equivalent detection limit (NEDL) of $\sim 6 \times 10^{-4}$ RIU.

**Results**

**Principle.** Fig. 1 illustrates the proposed CROW-based biochemical sensor. Fig. 1(a) shows the device schematic of a SOI CROW sensor comprising eight identically designed coupled microring resonators symmetrically coupled to input and output bus waveguides in an add-drop filter configuration. The out-of-plane elastic light scattering of the CROW is imaged by a top-view objective lens into an infrared (IR) camera in the far field. The sensor is integrated with a microfluidic channel on the top.

In the case that the dimensional disorders are small and the identically designed coupled microresonators are singlemode, the number of CROW eigenstates within each transmission band equals to the number of microresonators. A perfect CROW without dimensional disorders only exhibits at half of its complete set of eigenstates (within half of the transmission band) distinctive mode-field intensity distributions. The pair of symmetric and anti-symmetric intercavity-coupling-induced split modes have identical mode-field intensity distributions but distinctive mode-field amplitude distributions and eigenfrequencies. In practice, each coupled microresonator displays certain deviations from the design due to fabrication imperfection. This breaks the symmetry between the pair of symmetric and anti-symmetric split modes. An imperfect CROW thus exhibits distinctive mode-field amplitude and intensity distributions among its complete set of eigenstates. Fig. 1(b) illustrates for an imperfect 8-element CROW the inhomogeneously broadened transmission bands upon $n_0$ and $n_0 + \Delta n$, and the complete set of distinctive eigenstate pixelized mode-field intensity distributions upon $n_0$ denoted as $\{A_i\}$.

We integrate the mode-field intensity of each microring to form a pixelized one-dimensional (1D) pattern for the ease of analysis. Trading-off some detailed features of the distributions makes the pattern-recognition analysis of the mode-field-intensity distributions computationally efficient. Any mode-field amplitude profile at an arbitrary wavelength, $\lambda_p$, within the CROW transmission band upon $n_0$ can be given by a linear superposition of the complete set of the eigenstate mode-field amplitude distributions upon $n_0$. Therefore, it is conceivable to uniquely identify by a correlation analysis any pixelized mode-field intensity profile, $B(\lambda_p)$, as shown in inset (i), with $\{A_i\}$. Likewise, assuming a weak perturbation, we can uniquely identify by the correlation analysis any pixelized mode-field-intensity distribution upon a small global $\Delta n$ in the cladding, $B'(\lambda_p)$, as shown in inset (ii), with $\{A_i\}$.

**Correlation analysis and the sensing algorithm.** A unique feature in our correlation analysis is the use of the CROW eigenstate mode-field intensity distributions as intrinsic references. We adopt the Pearson’s correlation coefficient, $\rho$, in order to quantify the correlation between a pixelated pattern at an arbitrary wavelength $\lambda_p$, $B(\lambda_p)$, and the eigenstate pixelized patterns at the eigenstate wavelengths $\lambda_i$, $A(\lambda_i)$. For an $N$-element CROW, we define the correlation coefficient as follows:

$$\rho(\lambda_p) = \frac{1}{\sqrt{\sum_{i=1}^{N} (A_i(\lambda_p) - \bar{A}(\lambda_p))^2 \sum_{i=1}^{N} (B_i(\lambda_p) - \bar{B}(\lambda_p))^2}}$$

![Figure 1](https://www.nature.com/scientificreports/)

**Figure 1 | Working principle of CROW-based biochemical sensors in the spatial domain.** (a) Schematic of a SOI CROW sensor comprising eight coupled microring resonators in an add-drop filter configuration. (b) Illustration of an imperfect eight-element CROW, including the inhomogeneously broadened transmission bands upon $n_0$, $\lambda_p$, and the complete set of eigenstate normalized pixelized mode-field-intensity distributions upon $n_0$, $\{A_i\}$. Insets: (i) Pixelized mode-field-intensity distribution upon $n_0$, $B(\lambda_p)$. (ii) Pixelized mode-field intensity distribution upon a small $\Delta n$, $B'(\lambda_p)$.
where \( j = 1, 2, \ldots, N \) is the eigenstate number, \( i = 1, 2, \ldots, N \) is the cavity or pixel number, the pixel values \( A_i \) and \( B_i \) are normalized respectively to the total intensity of the entire patterns, the bar sign denotes the mean of the pixelized pattern over the number of pixels.

Previously\(^{20} \), Pearson’s product-moment correlation approach has been used to describe the dependence of a measured optical field on any other field. Here we detail our sensing algorithm. We first generate a library of \( \{ p_i(\lambda_0) \} \) at a wavelength \( \lambda_0 \) centered at the CROW transverse-electric (TE) mode. Fig. 2(b) shows the modeled transmission spectra of an imperfect CROW. The imperfect CROW comprises eight non-identical microring resonators. The measured waveguide width and coupling gap spacing of each microresonator were determined by finite-element method (FEM), the fraction of the optical mode into the water is 14.4%, which is much higher than the value of 5.0% in the transverse-electric (TE) mode. Fig. 2(c) shows the cross-sectional view of the numerically calculated transverse-electric (TE)-polarized waveguide mode-field amplitude profile. We choose the TM polarization mode in order to obtain a good mode-field exposure into the analyte on the CROW top surface. We assume \( n_0 = 1.318 \) (water cladding). According to our numerical modelling using finite-element method (FEM), the fraction of the optical mode into the water is 14.4%, which is much higher than the value of 5.0% in the transverse-electric (TE) mode.

\[ \Delta n = \text{RIU} \]

In order to uniquely identify \( \{ p_i(\lambda_0) \} \) from the library, we find from our modeling that it is sufficient to use only the principal component, \( p^1 \), and the second-principal component, \( p^2 \), of \( \{ p_i(\lambda_0) \} \), for \( N \) up to at least 28 (See Methods and Supplementary Information S1–S3). This streamlines the algorithm linearly by a factor of \( 2/N \).

**Modeling results.** We use transfer-matrix method to model the imperfect SOI CROW (see Methods and Supplementary Information S1). Fig. 2(a) shows the schematic of the imperfect SOI CROW. Inset shows the cross-sectional view of the numerically calculated transverse-electric (TE)-polarized waveguide mode-field amplitude profile. We choose the TM polarization mode in order to obtain a good mode-field exposure into the analyte on the CROW top surface. We assume \( n_0 = 1.318 \) (water cladding). According to our numerical modelling using finite-element method (FEM), the fraction of the optical mode into the water is 14.4%, which is much higher than the value of 5.0% in the transverse-electric (TE) mode. Fig. 2(b) shows the modeled transmission spectra of an imperfect CROW. The imperfect CROW comprises eight non-identical microring resonators. The measured waveguide width and coupling gap spacing of each microresonator vary following Gaussian distributions (see Methods and Supplementary Information S2). Fig. 2(c) shows the modeled eigenstate pixelized patterns. Fig. 2(d) shows the calculated \( \{ p_i(\lambda_0) \} \) as a function of \( \Delta n \), with \( \Delta n = 6.204 \times 10^{-5} \) RIU and \( \Delta n = 1.2 \times 10^{-4} \) RIU. Fig. 2(e) shows the calculated differential correlation coefficients per unit \( \Delta n \), given as (d(\( p_i^1(\lambda_0) \))/d(\( \Delta n \)))/p^2, in the opposite direction to \( \Delta n \). We define the sensitivity at an arbitrary probe wavelength \( \lambda_p \) as the larger (d(\( p_i^1(\lambda_0) \))/d(\( \Delta n \)))/p^2, in the opposite direction to \( \Delta n \).
Therefore, we can directly extract the sensitivity for $\lambda_p$ from the library of differential correlation coefficients at $\Delta n_B$ in Fig. 2(e). Fig. 2(f) shows a highly non-uniform distribution of the modeled sensitivity as a function of $\lambda_p$. The sensitivity spans a range of 1.6 RIU$^{-1}$ and 715 RIU$^{-1}$ over the spectral 3 dB-bandwidth with an average sensitivity of $\sim 279$ RIU$^{-1}$. It is highly dependent on the choice of $\lambda_p$. For modeling the sensing in the spatial domain, we first arbitrarily choose a fixed probe wavelength $\lambda_p$ at 1556.66 nm near the center of the CROW transmission band (Fig. 2(b)). The sensitivity at $\lambda_p$ is $\sim 306$ RIU$^{-1}$.

Fig. 3 illustrates the modeling of the CROW-based sensing using the correlation analysis. Fig. 3(a) shows the modeled pixelized patterns at $\lambda_p$ without (buffer) and with (test) applying a $\Delta n$ that is arbitrarily chosen as $2.68 \times 10^{-3}$ RIU. Fig. 3(b) shows the two sets of modeled correlation coefficients of the two pixelized patterns without and with $\Delta n$. The $\rho^p$ and $\rho^s$ without $\Delta n$ are $\rho_{5s}$ and $\rho_{4s}$, respectively. The $\rho^p$ and $\rho^s$ with $\Delta n$ are $\rho_{5s}$ and $\rho_{4s}$, respectively. Fig. 3(c) shows a zoom-in-view of the calculated library of $\rho_{5s}$ and $\rho_{4s}$ as a function of $\Delta n$ and mapping of $\rho_{5s}$ and $\rho_{4s}$. We extract from the library $\Delta n = \Delta n_B - \Delta n_B = 2.68 \times 10^{-3}$ RIU, which agrees with the arbitrarily chosen $\Delta n$ value.

**Calibration of the CROW sensor.** Fig. 4(a) shows the scanning-electron microscope (SEM) picture of the fabricated 8-element microring-based CROW. The racetrack microring comprises two half circles with a radius of 6.5 $\mu$m and two straight waveguides with an interaction length of 3.5 $\mu$m and a designed coupling gap spacing of 100 nm. We design the inter-cavity coupling in the strong-coupling regime in order to obtain a wide inhomogeneously broadened transmission band. This enables a large sensing dynamic range $\Delta n_B$ and a wide spectral range for choosing an arbitrary probe wavelength. Fig. 4(b) shows a representative zoom-in-view image of the inter-cavity coupling region.

Fig. 4(c) schematically shows the cross-sectional view of the optofluidic chip. The fabricated SOI chip is bonded with a polydimethylsiloxane (PDMS) layer, with a microfluidic channel of 50 $\mu$m-height and 1 mm-wide encompassing the CROW sensor. Fig. 4(d) schematically shows the experimental setup (see Methods).
Fig. 5(a) shows the measured TM-polarized transmission spectra of the 8-element CROW device with DI water as the upper-cladding. Green dashed-line: reference wavelength $\lambda_0$ of 1562.72 nm. Red dashed-lines: probe wavelengths, $\lambda_{p1}$ (1563.50 nm) and $\lambda_{p2}$ (1565.56 nm). (b) Measured infrared light-scattering images of the CROW with DI water upper-cladding at eigenstates I-VIII. The white-line box indicates the integration window for the pixelized patterns. (c) Pixelized mode-field intensity patterns at eigenstates I–VIII. (d) Calibrated library of the correlation coefficients $r_1$–$r_8$ as a function of $\Delta n$. White dashed-lines indicate the $\Delta n_p$ values at $\lambda_{p1}$ and $\lambda_{p2}$. (e) Calculated differential correlation coefficients as a function of $\Delta n$. (f) Calculated sensitivity as a function of probe wavelength. Red dashed-lines indicate a sensitivity of 43 RIU$^{-1}$ at $\lambda_{p1}$ and 752 RIU$^{-1}$ at $\lambda_{p2}$. (g) Calculated noise-equivalent detection limit as a function of probe wavelength. Red dashed-lines indicate a NEDL of $5.4 \times 10^{-5}$ RIU at $\lambda_{p1}$ and $6 \times 10^{-6}$ RIU at $\lambda_{p2}$.

Fig. 5(a) shows the measured TM-polarized transmission spectra of the CROW sensor covered by deionized (DI) water as the upper cladding. The CROW sensor exhibits an inhomogeneously broadened transmission band with a 3 dB bandwidth of $\approx 8.1$ nm. The measured free spectral range of 11.2 nm is consistent with the microring circumference. We discern the eight eigenstates as the eight resonance dips in the throughput spectrum (labelled by I to VIII).

Fig. 5(b) shows the measured light-scattering images of the CROW with DI water upper cladding at eigenstates I–VIII. We observe from the images highly non-uniform light-scattering profiles across each microring. We attribute such a non-uniformity to random variations of the surface roughness on each microring (see Supplementary Information S4). Such surface-roughness-induced scattering modulates the out-of-plane elastic light scattering patterns from the original mode-field distributions.

We integrate the out-of-plane scattering light intensity from each microring within a fixed integration window to form a pixel (Fig. 5(b)). The integration window covers both arcs of a microring, but excludes the two coupling regions in order to minimize the cross-talk between adjacent microring cavities. We correct the integrated patterns by normalizing with the estimated contributions of the surface-roughness-induced scattering (see Supplementary Information S4). Fig. 5(c) shows the corrected pixelized mode-field intensity patterns at the eight eigenstates, which are clearly distinguishable. Fig. 5(d) shows the measured library of the calibrated correlation coefficients as a function of $\Delta n$. We calibrate by scanning the input laser wavelength by $\pm \Delta \lambda$ about the center of the CROW transmission band upon a fixed buffer solution (DI water) with minimum wavelength interval of 0.02 nm. This wavelength interval corresponds to a $\Delta n$, of $1.87 \times 10^{-4}$ RIU, based on the calibrated linear spectral sensitivity of $\sim 106.82$ nm/RIU of the CROW sensor (see...
Supplementary Information S5). We convert $\Delta \lambda$ to $\Delta n$ using the calibrated linear spectral sensitivity. We choose $\Delta \lambda = 4.7$ nm such that the corresponding $\Delta n$ ranging from $-4.40 \times 10^{-3}$ RIU to $4.40 \times 10^{-2}$ RIU ($\Delta n_2 = 8.80 \times 10^{-2}$ RIU), which sequentially yields a unity value for $\rho_1$ to $\rho_n$.

Fig. 5(e) shows the calculated differential correlation coefficients, $d \rho / d \Delta n$, as a function of $\Delta n$. Fig. 5(f) shows the calculated sensitivity as a function of $\lambda_p$. The calculated sensitivity shows a highly non-uniform profile, ranging from 5 to 1412 RIU, with an average value of 199 RIU$^{-1}$ over the 3 dB bandwidth of the CROW transmission band.

We define the NEDL at $\lambda_p$ as the uncertainty of extracted $\Delta n$ (see Methods). We extract the NEDL from the measured uncertainty of each $\rho^p$ and $\rho'$ of the library {$p, (\rho_{n})$,} at $\Delta n_p$. Fig. 5(g) shows the calculated NEDL as a function of $\lambda_p$. NEDL is highly dependent on the choice of $\lambda_p$. We show an average NEDL over the entire transmission band as $\sim 6 \times 10^{-3}$ RIU.

**Sensing in the spatial domain with correlation analysis.** We first implement a blind test with an arbitrarily set probe wavelength $\lambda_p$ (1563.50 nm) near the center of the CROW transmission band. The sensitivity at $\lambda_p$ is only $\sim 43$ RIU$^{-1}$ (see Fig. 5(f)). The NEDL at $\lambda_p$ is $5.4 \times 10^{-3}$ RIU (see Fig. 5(g)). We prepare one buffer solution (DI water) and two NaCl solutions, X and Y, with mass concentrations close to the $\lambda_p$.

Fig. 6 shows the experimental sensing results at $\lambda_p$. Fig. 6(a) shows the measured light-scattering images of the CROW upon the buffer solution and the test solutions at $\lambda_p$. Fig. 6(b) shows the corresponding pixelized patterns. Fig. 6(c) shows the corresponding calculated correlation coefficients upon the buffer solution and solutions X and Y. Dotted-line boxes: $\rho'$, dashed-line boxes: $\rho$. (d)–(f) Mapping of $\rho^p$ and $\rho'$ with the library to extract $\Delta n$. (d) Upon the buffer solution. (e) Upon solution X. (f) Upon solution Y.

For solution X (Fig. 6(e)), we observe a significant pattern change from the buffer solution. We obtain $\rho'$ as $\rho_1$ (0.857 ± 0.004) and $\rho'$ as $\rho_3$ (0.24 ± 0.04). By mapping $\rho_1$ and $\rho_3$ to the library, we uniquely identify $\Delta n_B$ as (1.11 ± 0.12) $\times 10^{-3}$ RIU. Thus, we acquire for solution X a $\Delta n = \Delta n_B - \Delta n_B = (8.41 \pm 0.18) \times 10^{-3}$ RIU, corresponding to a mass concentration of (4.67 ± 0.10) %. This agrees with the prepared concentration of solution X (4.80 ± 0.01) %.

For solution Y (Fig. 6(f)), we find $\rho'$ as $\rho_4$ (0.969 ± 0.002) and $\rho'$ as $\rho_3$ (0.650 ± 0.006). These values are close to those observed from the buffer solution, suggesting a very small $\Delta n$. By mapping $\rho_4$ and $\rho_3$ to the library, we uniquely identify $\Delta n_B = (7.15 \pm 0.10) \times 10^{-3}$ RIU.

We implement another blind test using the same solutions X and Y at a specifically chosen wavelength $\lambda_p$ (1565.56 nm) within the transmission band. At $\lambda_p$, we obtain a higher sensitivity of $\sim 752$ RIU$^{-1}$ (see Fig. 5(f)) and a lower NEDL of $6 \times 10^{-5}$ RIU (see Fig. 5(g)) compared with those obtained at $\lambda_p$.
The choice of the SOI platform, along with the use of a 1550 nm laser, amplifier and an expensive InGaAs camera in our experimental setup, is, however, not practical for point-of-care optical biosensing applications. A practical sensing system should be low cost, and the
sensing window is likely to be in the visible to near-infrared wavelengths. Thus, a future direction for practically implementing our sensing scheme is to adopt a silicon-based material platform that is transparent to the visible and near-infrared wavelengths, is compatible with CMOS processes, and offers a sufficiently high refractive index for a small device footprint. A possible choice is silicon nitride (SiN). The operational wavelength can then be below ~1000 nm, which allows the use of a low-power diode laser source and a standard silicon charge-coupled device (CCD) or CMOS camera to image the light scattering. The relatively low refractive index contrast between the SiN waveguide and the analyte also enables a better exposure of the TM mode to interact with the analyte.

In summary, we report a paradigm-shift biochemical sensing scheme in the spatial domain using chip-scale, microresonator-based CROWs on a SOI chip. Instead of using narrowband microresonator resonances, we use an inhomogeneously broadened transmission band of an imperfect CROW and the out-of-plane elastic light scattering patterns to attain a good average sensitivity (~199 RIU⁻¹) and a low detection limit (~9 × 10⁻⁷–~2 × 10⁻⁷ RIU) over a large refractive index range (8.8 × 10⁻² RIU). The correlation analysis employed takes into account of the whole mode-field-intensity distribution across the CROW. This sensing scheme is immune to the common external noise affecting all the coupled cavities, and thus enabling an improved tolerance to the equipment noise. Our blind tests using a SOI 8-microring CROW at fixed probe wavelengths and NaCl solutions with different mass concentrations showed the detection of an added refractive index change of 1.5 × 10⁻⁶ RIU. We showed a noise-equivalent detection limit of ~6 × 10⁻⁶ RIU at a specific fixed probe wavelength. We therefore envision that our demonstrated CROW-based sensing scheme using light-scattering pattern recognition and the correlation analysis can potentially offer an alternative route toward a high-performance, reliable and relatively compact integrated label-free optical biochemical sensor.

**Methods**

**Transfer-matrix modeling on an imperfect CROW.** We model an imperfect microring CROW using transfer-matrix method with empirical and numerical inputs (see Supplementary Information S1). We accumulate statistics of the measured waveguide widths and coupling gap widths from SEM characterization over the nine coupling regions of a representative 8-microring CROW (see Supplementary Information S2). For each coupling region, we sample six waveguide widths and three coupling gap spacings. The statistics of the waveguide widths and the coupling gap widths approximately follow Gaussian distributions. We obtain the mean values and the standard deviations of fabricated waveguide widths and coupling gap widths. We assumed that Gaussian distributions of the waveguide widths and the coupling gap widths are independent. We use the Gaussian number generator in Matlab to generate a set of randomized waveguide widths and coupling gap widths distributed along an imperfect CROW. We study 200 sets of such randomly generated parameters for the 8-microring CROW (see Supplementary Information S7).

We calculate using FEM (COMSOL RF module) the waveguide effective refractive index, \( n_{\text{eff}} \), of a SOI waveguide with a water upper-cladding as a function of the waveguide width, at a fixed waveguide height of 240 nm. The mean value of the calculated \( n_{\text{eff}} \) is 1.839 ± 0.002. We calculate using two-dimensional (2D) finite-difference time-domain (FDTD) simulations the coupling coefficient in each coupling region as a function of the coupling gap width, with the waveguide width fixed at its mean value of 470 nm. The mean value of the calculated coupling coefficient is 0.910 ± 0.004. In the simulations, we choose the TM polarization mode. Based on our experiments, we estimate the waveguide propagation loss to be relatively high at 22 dB/cm, which we attribute primarily to surface-roughness-induced scattering losses. We assume that each microring follows the designed racetrack arc radius of 22 dB/cm, which we attribute primarily to surface-roughness-induced scattering. Across the CROW, this sensing scheme is immune to the relatively high pump pressure.

**Experimental setup.** The wavelength-tunable laser light in the 1550 nm wavelength range is coupled into a 33 dB-gain erbium-doped fiber amplifier. The amplified laser light is polarization-controlled with the TM polarization aligned and coupled into a tapered silicon waveguide through a singlemode polarization-maintaining fiber. We prepare NaCl solutions with mass concentrations from 1% to 5% (in steps of 1%) in order to calibrate the CROW transmission band spectral shifts upon a refractive index change from the cladding solution. All the measurements are repeated three times, with DI water as the reference cladding solution. Between two measurements, we rinse the chip by DI water for one time. We determine the resonance spectral shifts by fitting the throughput- and drop-transmission spectra with a sum of multiple Lorentzian lineshapes, with each Lorentzian lineshape centered at the CROW eigenstate wavelength. The overall transmission band shift is taken as the average value of the spectral shifts of eigenstates I–VIII.

**Imaging of elastic light scattering.** We use a long-working-distance microscope objective lens (30X Mitutoyo Plan Apo, NA = 0.55) and an InGaAs camera (Hamamatsu C10633-23) with 320 × 256 pixels (a 30 µm pixel size) to image the light scattering patterns from the top. The camera has a high responsivity at the 1000–1500 nm wavelength range with 14 bit analog-to-digital conversion in dark readout. We set the camera exposure time as 6 ms with a calibrated gamma factor of ~1. Each microring is imaged onto an array of 44 × 25 pixels. The integration window for each microring includes 44 × 19 pixels. We set the probe wavelength far away from the CROW transmission band and obtain a background image for background subtraction.

In order to acquire the calibrated correlation coefficients, we scan the laser wavelength and record 8 successive images for each wavelength step over a time interval of 0.8 s and take average for reducing the equipment noise contribution. In the blind sensing tests, we record and take average over 100 successive images during a time period of 30 s at a fixed probe wavelength after the buffer or the test solution is injected and the scattering pattern is stabilized.

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Author contributions
A.W.P. initiated and supervised all of the work. T.L. designed the device and conducted the initial experimental work. J.W. and T.L. carried out the device fabrication. J.W. conducted the experiment and the data analysis. Z.Y. and J.W. performed the numerical modeling. A.W.P., J.W. and Z.Y. wrote the manuscript.

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