

# Reflector-less refractive index sensors based on etched-tapered high-scattering single-mode optical fibers

Madina Shaimerdenova<sup>a,b</sup>, Carlo Molardi<sup>a</sup>, Wilfried Blanc<sup>c</sup>, Daniele Tosi<sup>a,b,\*</sup>

<sup>a</sup> Nazarbayev University, School of Engineering and Digital Sciences, Astana, Kazakhstan

<sup>b</sup> National Laboratory Astana, Laboratory of Biosensors and Bioinstruments, Astana, Kazakhstan

<sup>c</sup> Université Côte d'Azur, INPHYNI, CNRS UMR7010, Avenue Joseph Vallot, 06108 Nice, France

## ARTICLE INFO

### Keywords:

Optical fiber sensors  
Distributed Sensors  
Rayleigh scattering  
Optical backscattering reflectometry  
Refractive index sensors  
Reflector-less sensors

## ABSTRACT

Reflector-less refractive index (RI) sensors employ simple optical fiber strictures for the accurate detection of RI changes. They operate in absence of any reflective element such as gratings, filters, interferometers while maintaining the form factor of a reflective probe; the RI dependence is encoded in the Rayleigh backscattering signatures, and is interrogated using optical backscatter reflectometry. In this work, we present a new reflector-less fiber-optic sensor based on two-steps: (1) tapering an optical fiber, followed by (2) rapid etching of the fiber. The combination of the two steps allows obtaining significant advantages in terms of manufacturing, making the process simpler and increasing the distribution of the refractive index detection. The sensitivity obtained is up to 110.9 GHz/RIU (refractive index units), with the possibility of detection over a length of 2.1 mm, with 0.035 mm average spatial resolution over distributed RI sensing.

## 1. Introduction

The development of optical fiber refractive index (RI) sensors [1] and biosensors [2–3] based on single-mode fibers (SMFs) has been quite substantial in the recent years [4] with the direction of outperforming multimode fiber sensors in applicability and resolution.

Sensors operating with plasmonic fibers [5–6] based on a heavily multimode structure such as large-core glass fibers [7] or polymer fibers [8] have been commonly employed in RI detection, in applications such as water quality monitoring [9] or chemical sensing [7], taking advantage of low-cost hardware such as portable spectrometers [10] or smartphones [11]. RI sensors are also the building block of label-free biosensors, with the addition of a surface functionalization [2]. These systems however have three main problems that limit their potential for applications: 1) due to the large number of propagating modes, despite a high RI sensitivity, the biological or biochemical detections are hampered by large number of weakly guided modes; 2) spectrometers have resolution approaching the nanometer-level, while telecom-based interrogators can detect wavelengths below the picometer level [4]; 3) plasmonic structures operate in transmission, while *in situ* biomedical applications demand for reflective probes with compact form factors.

For this reason, SMF fibers biosensors have gained a significant

interest; the main architectures are long-period gratings [12], tilted fiber Bragg gratings [13–14], modal interferometers [15–16], micro-fiber interferometers [17] and on-tip fiber devices [18]. These systems allow achieving high performance ratings with narrow wavelength resolution hardware; however, as reviewed in [4], their fabrication is quite cumbersome, not industrially scalable due to the fact that high performance can be achieved only by the combination of inscription of reflective elements and fiber manipulations (such as splices of non-standard fibers or operation with photonic crystal fibers) that cannot be scaled, or by tapering structures that reduce the fiber thickness to few micrometers, making the fiber very fragile for any packaging [19].

In contrast, the concept of reflector-less RI sensors [20–22] and biosensors [23] was introduced as an alternative to these methods with a much simpler and scalable fabrication; the essence of reflector-less sensing is to operate without any reflective element, using a single-mode fiber with a simple and scalable fabrication, and encoding the entire detection in the interrogation of Rayleigh [24] or Brillouin [25] scattering signatures.

Reflector-less RI sensors are interrogated using optical frequency-domain reflectometry methods; in the most common implementation where the reflection is encoded in the Rayleigh scattering, the interrogator is an optical backscatter reflectometer (OBR) [26]. For this

\* Corresponding author at: Nazarbayev University, School of Engineering and Digital Sciences, Astana, Kazakhstan.

E-mail address: [daniele.tosi@nu.edu.kz](mailto:daniele.tosi@nu.edu.kz) (D. Tosi).

approach, the same hardware and software routines currently employed in high-resolution distributed sensing can be used for the RI detection, allowing a distributed detection of the RI over a short distance. The possibility of detecting the RI changes with the addition of a spatial distribution can open important avenues in the development of smart medical devices, that perform continuous monitoring of drug delivery [27]; exploiting the miniature form factor and the biocompatibility (ISO 10,993 standard) of optical fibers, a distributed detection of the RI can be achieved. While most recent distributed systems do not operate in reflection, as they employ macrobending configurations [28–29] that cannot be embodied in a miniature medical device, reflector-less biosensors operate in reflection with a terminated fiber, which is suitable for packaging and catheterization. Typical applications demand the detection of  $10^{-3}$  RIU (refractive index units), and since most OBR systems have accuracy around 0.07 GHz, sensitivity figures around 70 GHz/RIU are required [20].

In this work, we introduce a new reflector-less fiber optic RI sensor that combines fiber tapering and a subsequent wet-etching of the fiber surface; this approach provides the best trade-off between fabrication ease, sensitivity, and width of RI detection while maintaining the functionality as a reflective probe. The fabrication is based on a standard shallow taper on a telecom fiber, based on a CO<sub>2</sub> laser splicer; it is followed by a short wet-etching process which enhances the sensitivity. This two-step approach is first evaluated numerically, showing the changes in the fiber effective index (converted by the OBR into a frequency shift [30]); then, it is experimentally verified on a device etched up to 3 min in hydrofluoric acid. The proposed approach extends over previous reflector-less biosensors as it simplifies the fabrication process, using commercial splicers for taper fabrication and a shorter etching process for sensitivity enhancement.

## 2. Working principle

In order to understand the working principle of this reflector-less fiber sensor which presents the combination of two fabrications, namely tapering and etching, it is necessary to investigate the mode propagation in the fiber. Particularly, it is of interest to analyze the region of the fiber which presents the modification, since the tapered and etched region is where the sensing capability shall occur. The region presents a modified geometry that deviates from the ideal case of the pristine fiber where the modes are completely confined in the core and the cladding thickness is enough to prevent any interaction between the core modes and the external analyte. In the tapered and etched region, two relevant effects appear: the reduction of core size, given by the tapering, which can induce a migration of the core modes inside the cladding; the reduction of the cladding thickness, which can improve the interaction between the guided modes and the external analyte. These two combined effects have an impact on the sensing capability of the structure, and it is necessary to assess, from the theoretical point of view, when these effects start to appear and what is their impact on the propagation properties. It is of interest to estimate the effective refractive index ( $n_{\text{eff}}$ ) of the Fundamental Mode (FM), which gives an idea of how much the modes are confined in the core or expanded in the cladding. Furthermore, it is directly related to the backscattering spectrum taken by the OBR, whose shift is used to determine the value of the analyte refractive index. It is also of interest to estimate the Mode Field Diameter (MFD) because it gives an insight of the interaction of the propagating modes with the analyte.

In a standard telecommunication fiber, it is possible to estimate, under particular approximation, the guidance properties of the light by knowing the index contrast between core and cladding. The approximation relies on the fact that the cladding diameter is much larger than the core diameter, so that the reflection from the cladding-coating interface is negligible. In the structure presented here such an approximation is not valid anymore since the etching process significantly reduces the thickness of the cladding. The fiber becomes a three-layer

**Table 1**

Corning SMF-28 Ultra specifications.

| Fiber parameter                               | Value                         |
|-----------------------------------------------|-------------------------------|
| Core Diameter                                 | 8.2 $\mu\text{m}$             |
| Cladding Diameter                             | 125.0 $\pm$ 0.7 $\mu\text{m}$ |
| NA (measured on near field @ 1310 nm)         | 0.14                          |
| Effective Group Index of Refraction @ 1310 nm | 1.4676                        |
| Effective Group Index of Refraction @ 1550 nm | 1.4682                        |
| Mode Field Diameter @ 1310 nm                 | 9.2 $\pm$ 0.4 $\mu\text{m}$   |
| Mode Field Diameter @ 1550 nm                 | 10.4 $\pm$ 0.5 $\mu\text{m}$  |

structure consisting of core, cladding, and external analyte, which can be investigated by a numerical analysis based on Finite Element Method (FEM) [31]. FEM is particularly useful to assess the modes propagation properties of a cylindrical structure like an optical fiber, or in this case, like a tapered and etched region of an optical fiber, since the propagation properties are fully determined by the geometry of the fiber cross section. The problem to solve is a second order boundary value problem that derives from the application of time-harmonic Maxwell equations to the fiber structure where the proper boundary conditions are imposed. After the discretization of the 2D cross-section imposed by the FEM, the boundary value problem becomes an algebraic problem, namely a modified eigen value problem, where the eigenvectors represent the field distribution of the modes, while the eigenvalues are their related effective indices [32].

To simulate the three-layer structure, it has been considered a Corning SMF-28 Ultra, a typical Single Mode Fiber (SMF) for telecommunications that can also be effectively used as distributed sensor in OBR systems [33]. The fiber parameters, which are relevant for the simulations, are shown in Table 1.

The producer does not reveal the recipe of the fiber profile nor the value of the refractive index of the glass used in the fiber, therefore for the simulations it has been assumed a perfect step index profile and a value of the cladding refractive index equal to 1.4449 at the working wavelength of 1550 nm. With such assumptions the propagation characteristics of the Corning SMF-28 Ultra, in particular the MFD and the Effective Group Index, have been matched. For the numerical analysis it has been considered the working wavelength of 1550 nm, which is the central wavelength of the OBR spectrum. The parameters that have been considered for the simulations are the Tapering Ratio ( $TR$ ), which is the proportional resizing of core and cladding, and the Etching Factor ( $F_{\text{etch}}$ ), which represents the percentage of cladding that is removed through the etching process. The tapering ratio has been varied from 0.1 to 1, equivalent to the untapered fiber, and the etching factor has been varied for 0.2 to 1, equivalent to the unetched fiber. The final radii of the core and cladding are defined by the following formulas:

$$Rco_{\text{final}} = Rco_{\text{unetched}} \bullet TR \quad (1a)$$

$$Rcl_{\text{final}} = (\Delta cl_{\text{unetched}} \bullet F_{\text{etch}} + Rco_{\text{unetched}}) \bullet TR \quad (1b)$$

where  $Rco$  and  $Rcl$  are the radii of, respectively, core and cladding, while  $\Delta cl$  is the thickness of the cladding equal to  $Rcl - Rco$ . The schematic of the structure is shown in Fig. 1.

In order to understand the behavior of structure in response to the variation of the external analyte refractive index, three different RI have been considered: 1.318 which is the refractive index of pure water at 1550 nm, 1.36425 which corresponds to a 25% sucrose solution, and 1.4105 which corresponds to a 50% sucrose solution. From the simulation, it is possible to understand that both tapering and etching are necessary to increase the sensitivity of the reflector-less sensor to the external analyte. In particular, the tapered fiber without etching shows an extremely poor interaction with the external analyte, as depicted in Fig. 2.

The effective index of the fundamental mode of the unetched fiber, which can be considered the main factor for the backscattering spectral

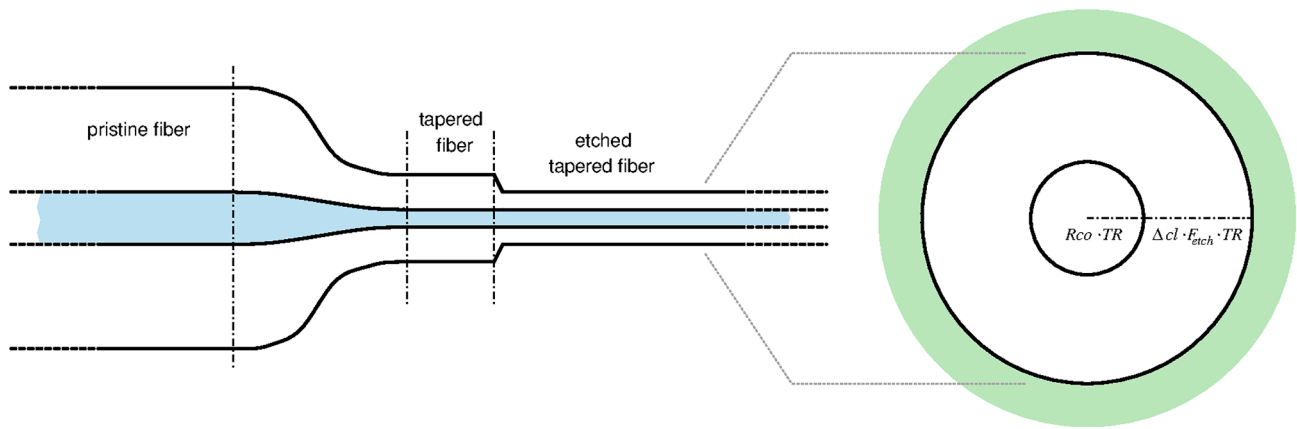


Fig. 1. The schematic of the structure which has been tapered and then etched. The modal analysis has been performed on the cross-section of the etched area.

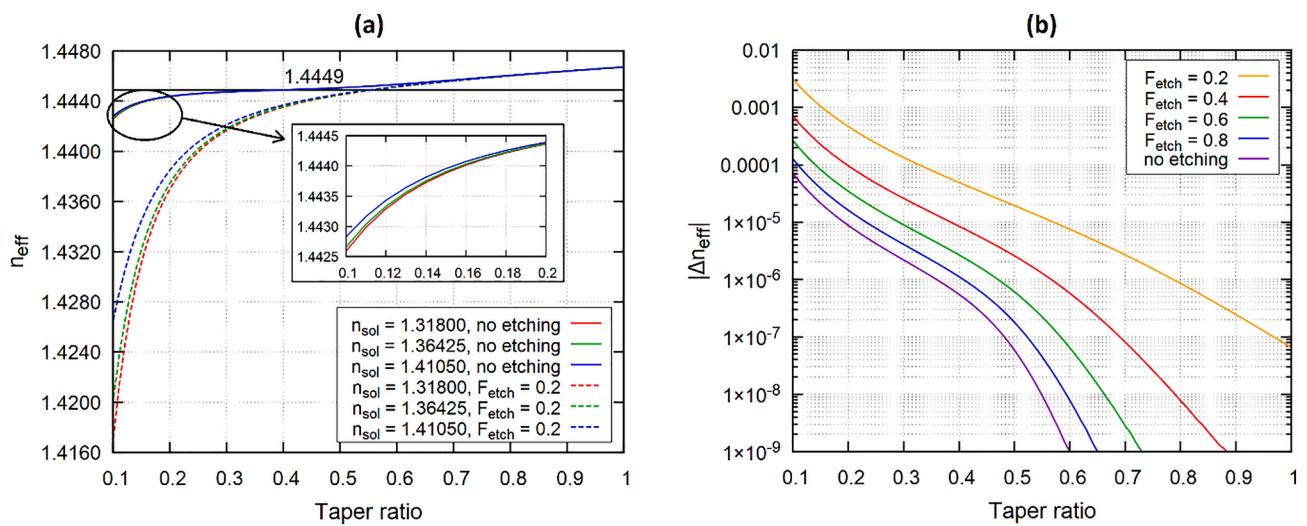


Fig. 2. Variation of the effective index of the fundamental mode with respect the variation of the tapering ratio, the etching factor, and the refractive index of the external analyte. (a) Comparison between the tapered fiber with no etching and etching of  $0.2 F_{etch}$ . (b) Shift of FM effective index calculated between a 25% sucrose solution and pure water.

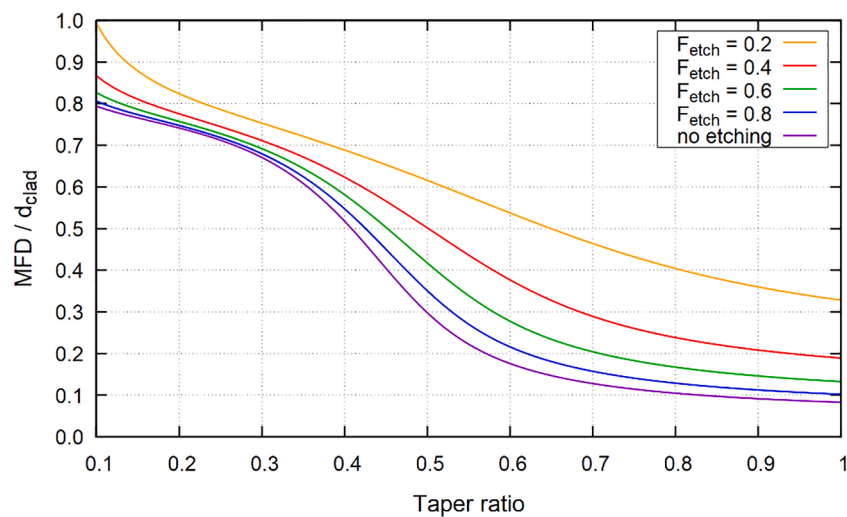


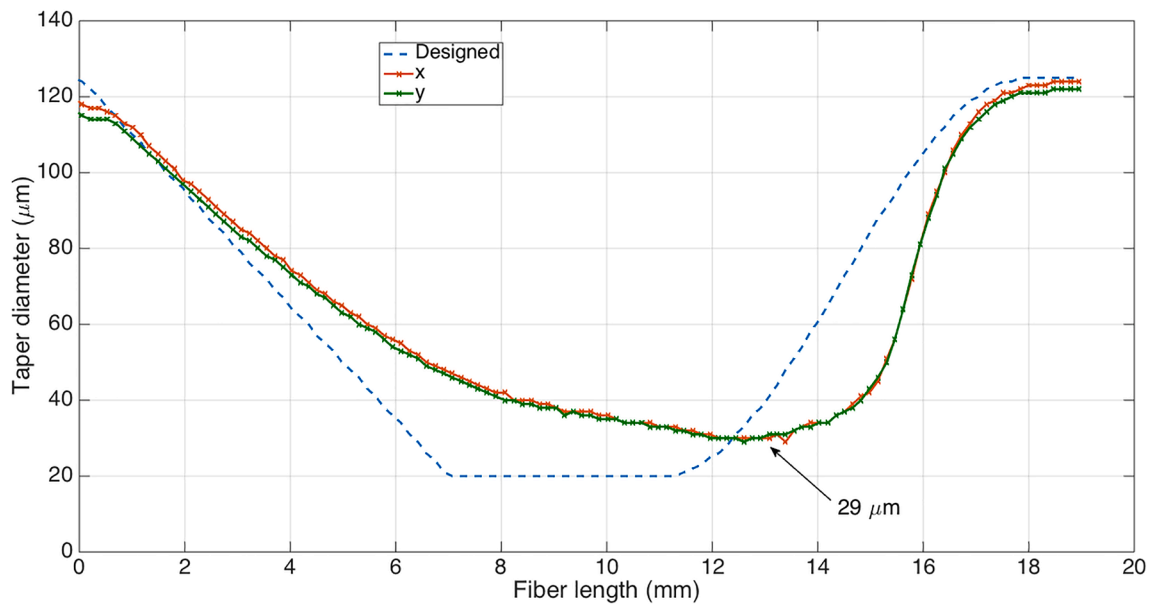
Fig. 3. Ratio between the MFD and the modified cladding diameter, simulated by varying the taper ratio and the etching factor, calculated using as analyte the 25% sucrose solution.

**Table 2**  
Taper fabrication parameters using CO<sub>2</sub> laser splicer.

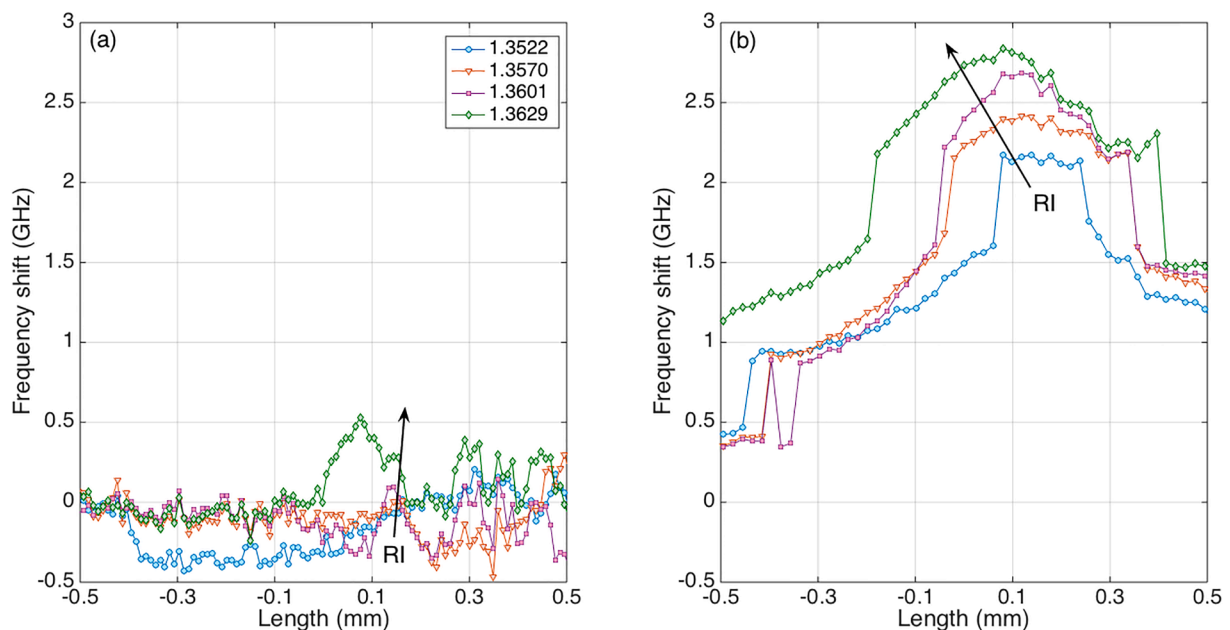
| Taper fabrication parameter | Value             |
|-----------------------------|-------------------|
| Initial fiber diameter      | 125 $\mu\text{m}$ |
| Waist diameter              | 30 $\mu\text{m}$  |
| Left taper length           | 2 mm              |
| Waist taper length          | 5 mm              |
| Right taper length          | 2 mm              |
| Pre-heat                    | 0 bits            |
| Absolute power              | 602 bits          |
| Relative power              | 8 bits            |
| Waist add                   | 20 bits           |
| Pulling speed               | 0.8 m/s           |

shift, presents a modest variation, in the best case of 0.1 tapering ratio, equivalent to  $2.4 \times 10^{-4}$  over a large refractive index variation of 0.093 RIU. With the same tapering ratio, a 0.2 etching factor fiber presents  $9.5 \times 10^{-3}$  effective index shift, roughly 40 times bigger. It is worth noting that the sensitivity of the FM effective index to the external analyte becomes more evident when the tapered fiber starts to excite the cladding modes. This happens with tapering ratios from 0.4 to 0.6, also depending on the etching factor. By considering analytes with refractive indices around 1.36, value more significant for the experimental work presented in the next sections, the etched fiber improves the FM effective index shift of a factor of roughly 100, as depicted in Fig. 2(b).

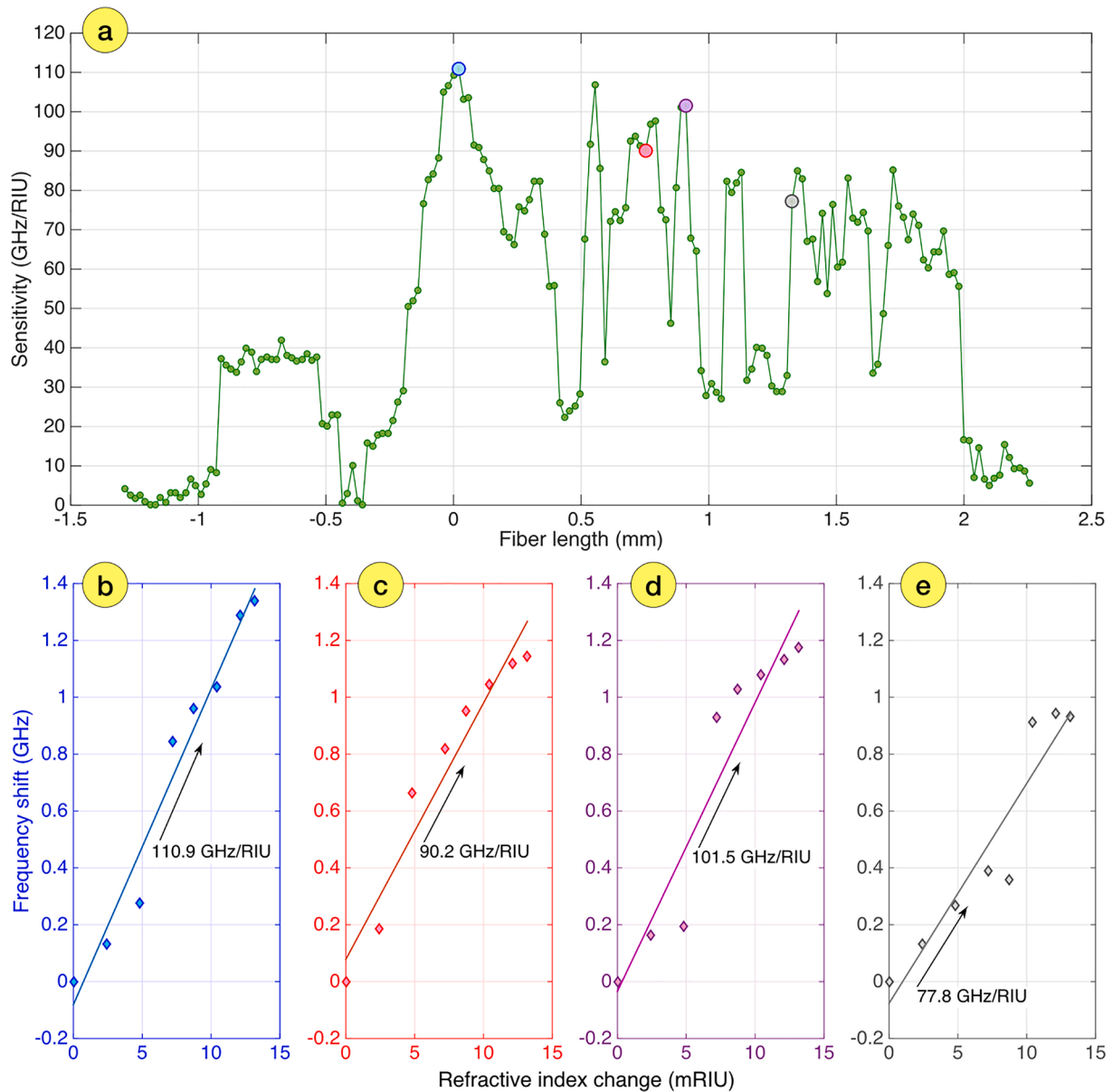
The improvement of the interaction between propagating light and analyte can be detected also by the analysis of the MFD. Fig. 3 shows ratio between the MFD and the modified cladding diameter, after the



**Fig. 4.** Profilometry of the fiber taper, along the x/y directions compared to the designed taper profile.



**Fig. 5.** Frequency shift along the fiber taper, sampled with 19.8  $\mu\text{m}$  resolution for RI values increasing from 1.3522 to 1.3629. The Rayleigh frequency shift is recorded with reference value as 10% sucrose concentration. (a) Bare taper, with no etching. (b) Taper etched for 3 min in HF.



**Fig. 6.** Performance analysis for the etched-tapered reflector-less sensor after 3 min of wet-etching. (a) Sensitivity estimated for each point along the fiber length, sampled with 19.8  $\mu\text{m}$  resolution; for each section of the taper, the RI sensitivity has been estimated through a linear regression for the frequency shift for 8 different RI values ranging from 1.3497 to 1.3629. (b–e) Highlight for the RI sensitivity visualized for 4 different taper sections corresponding to the markers in the sensitivity trace: (b) Length 0.02 mm, corresponding to 110.9 GHz/RIU sensitivity; (c) Length 0.75 mm, corresponding to 90.2 GHz/RIU sensitivity; (d) Length 0.91 mm, corresponding to 101.5 GHz/RIU sensitivity; (e) Length 1.32 mm, corresponding to 77.8 GHz/RIU sensitivity.

tapering and the etching, calculated with the 25% sucrose solution analyte. It is possible to see that with stronger etching the penetration of the propagating light into the analyte is consistent also for moderate tapering ratios.

### 3. Fabrication and interrogation

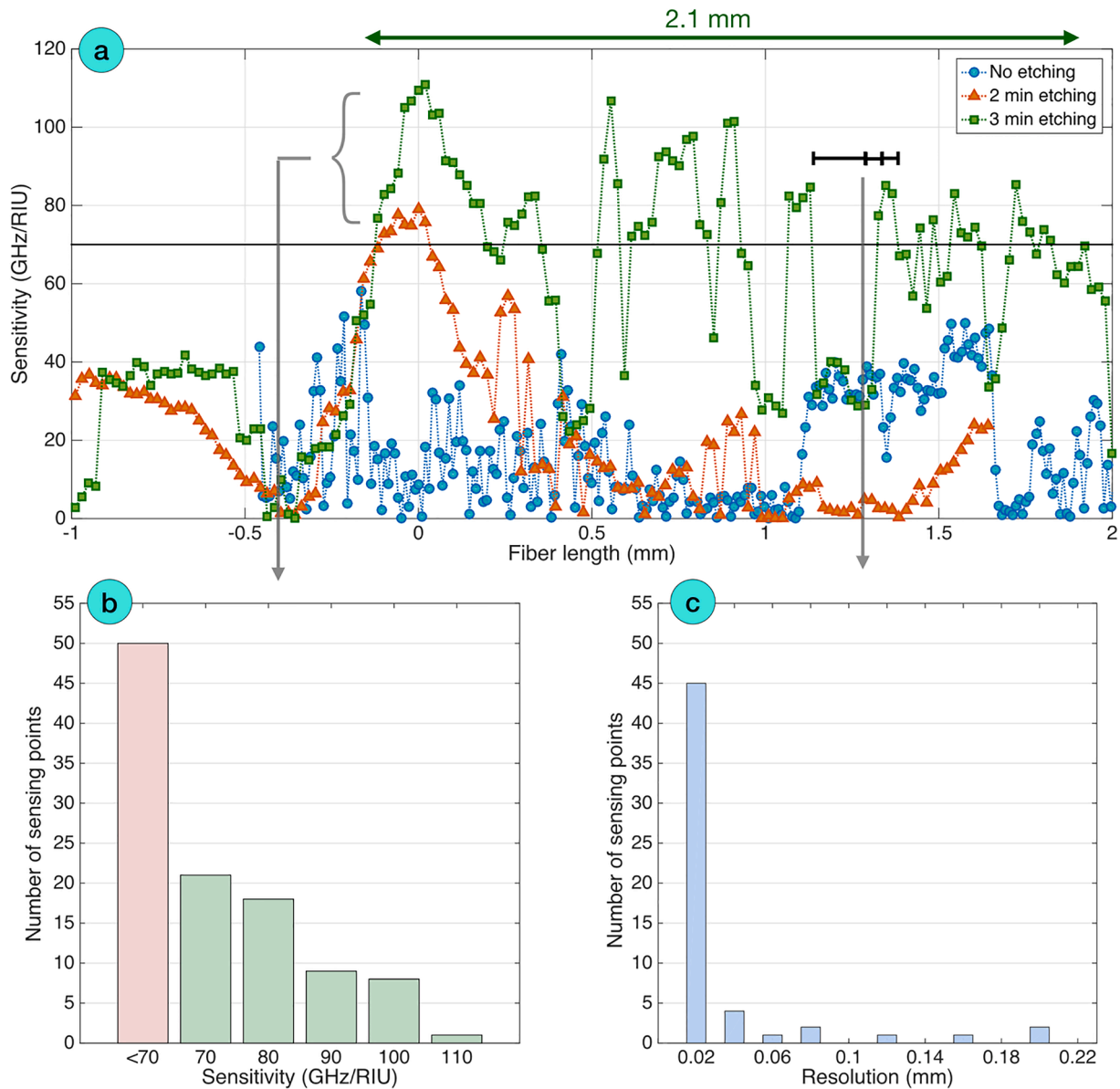
As for the previous works related reflector-less sensors [20–23], a high-scattering fiber was employed in the etched taper structure in order to compensate for the propagative losses. The fiber with enhanced backscattering was obtained by drawing a preform of 10 mm diameter. This preform was fabricated using a conventional MCVD (Modified Chemical Vapor Deposition) process [34]. A GeO<sub>2</sub>-doped silica porous core layer was immersed in the doping solution for 3 h. 30'. This solution was composed of a mixture of chloride salts, LaCl<sub>3</sub> (0.7 mol/L), dissolved in ethanol. La ions were introduced to trigger amorphous phase-

separated nanoparticles formation. After the doping step, the porous layer was sintered applying a temperature ramp up to 1800 °C. Then the tube collapsed into the final preform at a temperature above 2200 °C. The drawing of the fiber took place at a temperature around 2050 °C.

Subsequently, the fiber was tapered down to about 30  $\mu\text{m}$  with a CO<sub>2</sub> laser splicer (Fujikura LZM-100). The taper fabrication parameters, as set on the splicer, are listed in Table 2; the power is expressed in “bits” units to maintain the uniformity to the splicer settings. The profile of a taper with 29  $\mu\text{m}$  thickness is shown in Fig. 4.

After tapering, the fiber was wet-etched with Hydrofluoric acid (HF) (48%) for 3.5 min in fume hood at room temperature. The tapered region was located inside plastic lid of Eppendorf tube as it remains intact under the effect of HF acid. In 3.5 min, the taper was rinsed with DI water 3 times.

Etching was followed by calibration of the tapered fiber with various concentrations of sucrose solution diluted in water in order to measure



**Fig. 7.** (a) Sensitivity along each point of the fiber length, for the fabricated taper, and for the etched taper with 2 and 3 min of etching in HF; the upper bandwidth of 2.1 mm identifies the portion of lengths in which the sensitivity overcomes the target 70 GHz/RIU for the 3-min etched taper. (b) Within the RI sensing range (106 sensing points), the histogram shows the frequency of each sensitivity value. (c) For the 56 values having sensitivity > 70 GHz/RIU, the histogram shows the frequency of the spatial resolution between two neighboring points with sensitivity > 70 GHz/RIU.

refractive index change. RI values of sucrose solution used in our study range from 1.33304 up to 1.36292.

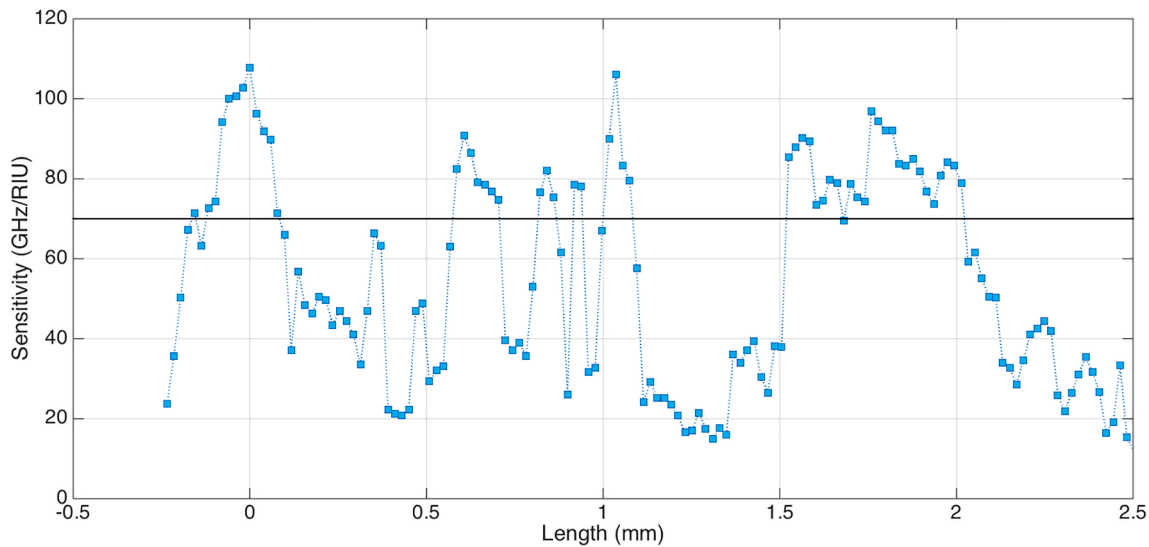
#### 4. Experimental results

Fig. 5 shows the working principle of the etched tapered reflectorless RI sensor. The chart displays the frequency shift observed along the fiber axis, whereas the zero location is in correspondence of the estimated location of the taper waist; the shift is recorded through the OBR with spatial resolution 19.8  $\mu\text{m}$ , for RI values increasing from 1.3522 to 1.3629 ( $\Delta\text{RI} = 10.7 \text{ mRIU}$ ).

The chart shows the effect of the fiber etching as a tool to increment the sensitivity. The first chart shows the bare taper, in absence of etching. The frequency shift is a noisy signal, limited by the OBR accuracy, in which the shift is perceived only at the highest value of RI. The second chart shows the frequency shift observed on the same device, after 3 min etching in HF. We can observe a visible increment of

sensitivity, as the shift is 2.1 GHz for the smaller RI value and rising over 2.8 GHz for the largest RI change. We also observe that the frequency shift envelope appears to have a clearer pattern, achieving its peak around 0.1–0.2 mm after the estimated position of the taper waist. From this chart, we can perceive the distributed pattern of the RI detection, as the sensitivity to RI is encoded in a set of locations in proximity of the fiber taper waist.

The effect of a 3-minute wet-etching in HF is visualized in Fig. 6. The first chart shows the sensitivity of the Rayleigh frequency shift to the changes of surrounding RI, measured over the whole taper region (3.9 mm). We observe that in correspondence of the taper waist we observe the maximum sensitivity (110.9 GHz/RIU); in this region the sensitivity peaks at the taper waist, and then drops below 60 GHz/RIU when moving  $-0.2 \text{ mm}$  to  $+0.3 \text{ mm}$  away from the peak of the sensitivity. Towards the negative lengths, in correspondence to the abrupt transition in the taper, the sensitivity appears to drop; on the other side, in the rightmost part of the taper the sensitivity holds high values in several



**Fig. 8.** Refractive index sensitivity for a different etched fiber taper, made from a taper with 30  $\mu\text{m}$  diameter after 2 min exposure in HF; the horizontal line marks the 70 GHz/RIU threshold.

data points, as in the 0.5–2.0 mm region we can observe that in several location the sensitivity is high.

Fig. 6(b-d) breaks down the RI sensitivity for some of the fiber locations, showing how the RI sensitivity is computed in several points along the taper length. The charts report the Rayleigh frequency shift for 8 different RI values spaced over 13.2 mRIU, estimated in the peak sensitivity point (0.02 mm length) and for different data points from 0.6 to 1.4 mm whereas sensitivity ratings  $> 70$  GHz/RIU are observed. The presence of a plurality of sensing points with relevant sensitivity enables short-range distributed RI sensing over a distance of about 2 mm.

The effect of wet-etching is shown in Fig. 7(a), where the sensitivity for each sensing point is compared, displaying the original taper and the etched taper (2–3 min in HF). For the taper without etching, the maximum figure rating that can be observed is  $< 60$  GHz/RIU; the most notable feature is bandwidth around 1.1–1.6 mm but the sensitivity is low. By etching for 2 min in HF, the sensitivity curve observes a rounder pattern, with a peak around the taper waist. The maximum value is 79.9 GHz/RIU, and we observed a 0.2 mm range of values in which the sensitivity overcomes the 70 GHz/RIU threshold. By further etching (3 min in total), we observe that the sensitivity rises across the whole tapered region: from  $-0.12$  mm to 1.98 mm (2.1 mm range in total), we observe a bandwidth of sensitivity values that overcomes the threshold.

Within the operative bandwidth of the taper (106 sensing points) we can perform a statistical analysis of the RI distributed sensitivity. Fig. 7(b) shows the frequency of each sensitivity value: 50 sensing points (47%) have sensitivity lower than the threshold, while 37% of the sensing points achieve 70–90 GHz/RIU sensitivity. We can define as the spatial resolution the distance between two adjacent sensing points with sensitivity over the threshold; the statistics of the spatial resolution are listed in Fig. 7(c), where the majority of the sensing points have resolution 0.02 mm equal to the nominal resolution of the OBR, and a few other points have resolution up to 0.20 mm due to some low-sensitivity points. Overall, the average spatial resolution is 0.035 mm which confirms the distributed refractive index detection of the etched-taper reflector-less sensor.

In order to prove the repeatability of the etching-tapering approach, Fig. 8 shows the RI sensitivity made for a different taper, having similar characteristics of the previous one (30  $\mu\text{m}$  diameter, etching for about 2 min). We again observe a similar peak sensitivity (107.7 GHz/RIU) and sensing range ( $-0.15$  mm to 2.01 mm, for a total of 2.16 mm).

## 5. Conclusions

This work reports a reflector-less refractive index fiber-optic sensor, based on a two-step etched-tapered fiber approach with interrogation based on OBR distributed sensing. The structure of the sensor is based on a shallow taper ( $\sim 1/5$  of the original fiber diameter), subsequently wet-etched in a HF bath. The obtained sensitivity achieves a maximum value of 110.9 GHz/RIU, evaluated as the Rayleigh scattering frequency shift as a function of the surrounding RI change. The sensor operates as a distributed RI sensor, with 2.1 mm operative range and 56 sensing points with sensitivity  $> 70$  GHz/RIU, which allows demodulating RI changes  $> 10$  mRIU with average spatial resolution 0.035 mRIU.

The achieved results are in line with previous reflector-less biosensors; in terms of RI sensitivity, the obtained results have sensitivity similar to the stand-alone etched fiber [23] however while the etching process for a fiber requires several minutes ( $\sim 45$  min as previously reported), which causes the fiber to become fragile and less suitable to sustain a biofunctionalization process and a biological measurement, in this work wet-etching is limited to 3 min and maintains the tensile strength and mechanical robustness of the original fiber. In addition, the etching process broadens the sensitive area, reaching a length of 2.1 mm, which is 40–20 times wider than the taper alone [20]. Overall, the two-step approach provides a scalable manufacturing process, since the fabrication of a shallow taper with a  $\text{CO}_2$  laser splicer is  $< 1$  min, and the etching process can be automated as in industrial electronics [35]. The fabrication process proposed in this work enables reflector-less sensors with adequate sensitivity ratings, while obtaining a high fabrication throughput. With respect to tapered fibers, the proposed approach allows loosening the specifications for taper waist diameter, which results in a simpler tapering process with most commercial  $\text{CO}_2$  laser or arc-induced splitters; conversely, the etching process is much faster and scalable than etching-only reflector-less biosensors, which has been reported for over 45 min wet-etching.

The proposed method extends reflector-less RI sensors to over 2 mm sensing range, maintaining a distributed sensing factor and enabling the detection of small RI changes. The probe is compatible to label-free biosensors, as the surface of the fiber can be functionalized with bio-receptors; the system can also open important perspectives in the design of smart drug delivery systems [27–36].

## Funding

The research was funded by Nazarbayev University, under grants SMARTER (code: 091019CRP2117), EPICGuide (code: 240919FD3908), “Distributed Measurement of Bite Force by Fiber-Optic Sensorized Dental Guard” (code: 021220FD1851), and M2O-DISK (code: 20122022FD4134).

## CRediT authorship contribution statement

**Madina Shaimerdenova:** Conceptualization, Methodology, Writing – review & editing. **Carlo Molardi:** Methodology, Software, Validation, Writing – original draft, Writing – review & editing. **Wilfried Blanc:** Resources, Writing – review & editing. **Daniele Tosi:** Conceptualization, Methodology, Formal analysis, Data curation, Supervision, Writing – original draft, Writing – review & editing, Funding acquisition.

## Declaration of Competing Interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

## Data availability

Data will be made available on request.

## References

- [1] A. Iadicicco, A. Cusano, S. Campopiano, A. Cutolo, M. Giordano, Thinned fiber Bragg gratings as refractive index sensors, *IEEE Sens. J.* 5 (6) (2005) 1288–1295.
- [2] A.B. Socorro-Leránz, D. Santano, I. Del Villar, I.R. Matias, Trends in the design of wavelength-based optical fiber biosensors (2008–2018), *Biosens. Bioelectron.* X 1 (2019) 100015.
- [3] N. De Acha, A.B. Socorro-Leránz, C. Elosua, I.R. Matias, Trends in the design of intensity-based optical fiber biosensors (2010–2020), *Biosensors* 11 (6) (2021) 197.
- [4] D. Tosi, M. Shaimerdenova, M. Sypabekova, T. Ayupova, Minimalistic design and rapid-fabrication single-mode fiber biosensors: Review and perspectives, *Opt. Fiber Technol.* 72 (2022) 102968.
- [5] A.K. Sharma, R. Jha, B.D. Gupta, Fiber-optic sensors based on surface plasmon resonance: a comprehensive review, *IEEE Sens. J.* 7 (8) (2007) 1118–1129.
- [6] A.K. Sharma, C. Marques, Design and performance perspectives on fiber optic sensors with plasmonic nanostructures and gratings: A review, *IEEE Sens. J.* 19 (17) (2019) 7168–7178.
- [7] C. Fallauto, Y. Liu, G. Perrone, A. Vallan, Compensated surface plasmon resonance sensor for long-term monitoring applications, *IEEE Trans. Instr. Meas.* 63 (5) (2013) 1287–1292.
- [8] N. Cennamo, M. Pesavento, L. Zeni, A review on simple and highly sensitive plastic optical fiber probes for bio-chemical sensing, *Sens. Actuators B: Chem.* 331 (2021) 129393.
- [9] P. Zhang, Y.P. Chen, W. Wang, Y. Shen, J.S. Guo, Surface plasmon resonance for water pollutant detection and water process analysis, *TrAC Trends Anal. Chem.* 85 (2016) 153–165.
- [10] C. Leitao, S.O. Pereira, C. Marques, N. Cennamo, L. Zeni, M. Shaimerdenova, T. Ayupova, D. Tosi, Cost-Effective Fiber Optic Solutions for Biosensing, *Biosensors* 12 (8) (2022) 575.
- [11] A. Sultangazin, J. Kusmangaliyev, A. Aitkulov, D. Akilbekova, M. Olivero, D. Tosi, Design of a smartphone plastic optical fiber chemical sensor for hydrogen sulfide detection, *IEEE Sens. J.* 17 (21) (2017) 6935–6940.
- [12] F. Esposito, A. Srivastava, L. Sansone, M. Giordano, S. Campopiano, A. Iadicicco, Label-free biosensors based on long period fiber gratings: a review, *IEEE Sens. J.* 21 (11) (2020) 12692–12705.
- [13] S. Maguis, G. Laffont, P. Ferdinand, B. Carbonnier, K. Kham, T. Mekhalif, M. C. Millot, Biofunctionalized tilted Fiber Bragg Gratings for label-free immunosensing, *Opt. Expr.* 16 (23) (2008) 19049–19062.
- [14] M. Loyez, J. Albert, C. Caucheteur, R. Wattiez, Cytokeratins biosensing using tilted fiber gratings, *Biosensors* 8 (3) (2018) 74.
- [15] W. Yu, T. Lang, J. Bian, W. Kong, Label-free fiber optic biosensor based on thin-core modal interferometer, *Sens. Actuators B: Chem.* 228 (2016) 322–339.
- [16] X. Li, L.V. Nguyen, Y. Zhao, H. Ebendorff-Heidepriem, S.C. Warren-Smith, High-sensitivity Sagnac-interferometer biosensor based on exposed core microstructured optical fiber, *Sens. Actuators B: Chem.* 269 (2018) 103–109.
- [17] S. Gao, L.P. Sun, J. Li, L. Jin, Y. Ran, Y. Huang, B.O. Guan, High-sensitivity DNA biosensor based on microfiber Sagnac interferometer, *Opt. Expr.* 25 (12) (2017) 13305–13313.
- [18] M. Shaimerdenova, T. Ayupova, M. Sypabekova, D. Tosi, Fiber optic refractive index sensors based on a ball resonator and optical backscatter interrogation, *Sensors* 20 (21) (2020) 6199.
- [19] P. Xu, X. Yu, Z. Chen, L. Sheng, J. Liu, S. Zhou, K. Wen, O. Xu, X. Dong, J. Yang, Y. Qin, Distributed refractive index sensing based on bending-induced multimodal interference and Rayleigh backscattering spectrum, *Opt. Expr.* 29(14), 21530–8, 2021.
- [20] M. Shaimerdenova, T. Ayupova, Z. Ashikbayeva, A. Bekmurzayeva, W. Blanc, D. Tosi, Reflector-Less Shallow-Tapered Optical Fiber Biosensors for Rapid Detection of Cancer Biomarkers, *J. Lightwave Technol.* 41 (13) (2023) 4114–4122.
- [21] M. Sypabekova, S. Korganbayev, W. Blanc, T. Ayupova, A. Bekmurzayeva, M. Shaimerdenova, K. Dukenbayev, C. Molardi, D. Tosi, Fiber optic refractive index sensors through spectral detection of Rayleigh backscattering in a chemically etched MgO-based nanoparticle-doped fiber, *Opt. Lett.* 43 (24) (2018) 5945–5948.
- [22] T. Ayupova, M. Shaimerdenova, S. Korganbayev, M. Sypabekova, A. Bekmurzayeva, W. Blanc, S. Sales, T. Guo, C. Molardi, D. Tosi, Fiber optic refractive index distributed multi-sensors by scattering-level multiplexing with MgO nanoparticle-doped fibers, *IEEE Sens. J.* 20 (5) (2019) 2504–2510.
- [23] M. Sypabekova, A. Aitkulov, W. Blanc, D. Tosi, Reflector-less nanoparticles doped optical fiber biosensor for the detection of proteins: Case thrombin, *Biosens. Bioelectron.* 165 (2020) 112365.
- [24] W. Blanc, Z. Lu, T. Robine, F. Pignonneau, C. Molardi, D. Tosi, Nanoparticles in optical fiber, issue and opportunity of light scattering, *Opt. Mat. Expr.* 12 (7) (2022) 2635–2652.
- [25] A. Minardo, L. Zeni, R. Bernini, E. Catalano, R. Vallifuoco, Quasi-distributed refractive index sensing by stimulated Brillouin scattering in tapered optical fibers, *J. Lightw. Technol.* 40 (8) (2022) 2619–2624.
- [26] M. Froggatt, J. Moore, High-spatial-resolution distributed strain measurement in optical fiber with Rayleigh scatter, *Appl. Opt.* 37 (10) (1998) 1735–1740.
- [27] F. Stradolini, A. Tuoheti, T. Kilic, N.L. Ntella, N. Tamburrano, Z. Huang, G. De Micheli, D. Demarchi, S. Carrara, An IoT solution for online monitoring of anesthetics in human serum based on an integrated fluidic bioelectronic system, *IEEE Trans. Biomed. Circ. Syst.* 12 (5) (2018) 1056–1064.
- [28] Y. Du, S. Jothibasu, Y. Zhuang, C. Zhu, J. Huang, Rayleigh backscattering based macrobending single mode fiber for distributed refractive index sensing, *Sens. Actuators B: Chem.* 248 (2017) 346–350.
- [29] P. Hua, Z. Ding, K. Liu, H. Guo, M. Pan, T. Zhang, S. Li, J. Jiang, T. Liu, Distributed optical fiber biosensor based on optical frequency domain reflectometry, *Biosens. Bioelectron.* 115184 (2023).
- [30] P.J. Thomas, J.O. Hellevang, A fully distributed fibre optic sensor for relative humidity measurements, *Sens. Actuators B: Chem.* 247 (2017) 284–289.
- [31] T. Ayupova, M. Sypabekova, C. Molardi, A. Bekmurzayeva, M. Shaimerdenova, K. Dukenbayev, D. Tosi, Wavelet-Based Demodulation of Multimode Etched Fiber Bragg Grating Refractive Index Sensor, *Sensors* 19 (2019) 39.
- [32] S. Selleri, L. Vincetti, A. Cucinotta, M. Zoboli, Complex FEM modal solver of optical waveguides with PML boundary conditions, *Optical and Quantum Electronics* 33 (2001) 359–371.
- [33] Corning® SMF-28® Ultra: <https://www.corning.com/optical-communications/worldwide/en/home/products/fiber/optical-fiber-products/smf-28-ultra.html>.
- [34] W. Blanc, V. Mauroy, L. Nguyen, B.N. Shivakiran Bhaktha, P. Sebbah, B.P. Pal, B. Dussardier, J. Heo, Fabrication of rare earth-doped transparent glass ceramic optical fibers by modified chemical vapor deposition, *J. Am. Ceram. Soc.* 94 (8) (2011) 2315–2318.
- [35] A. Jayarama, G.K. Kannarpady, S. Kale, S. Prabhu, R. Pinto, Chemical etching of glasses in hydrofluoric Acid: A brief review, *Mat. Today: Proc.* 55 (2022) 46–51.
- [36] M.J. Webber, D.G. Anderson, Smart approaches to glucose-responsive drug delivery, *J. Drug Targeting* 23 (7–8) (2015) 651–655.