Investigation of a highly sensitive vortex beam-shined optical sensor for the detection of glucose concentration in urine

Arijit Datta*, Mukta Chaturvedi

Department of Electrical & Electronics Engineering, CMR Institute of Technology, Bengaluru, Karnataka-560037, India.

*arijit.datta048@gmail.com

Abstract. The present work investigates the notion of a highly sensitive optical sensor for the detection of urinary glucose concentration by shining an optical Vortex Beam. The proposed sensing scheme relies on the theory of multimode interference where the transmitted output power varies due to the change in the refractive index of cladding as formed by various concentrations of glucose in urine. With a significant amount of energy at the beam’s periphery, Vortex beam can energize a massive number of higher order modes inside the sensing system. In support of theoretical analysis, an Eigenmode expansion solver is employed in Lumerical Mode solution software to visualize the full transmission spectrum of the sensor. In contrast to the conventional sensor, the proposed sensor proffers a 3.65 times superior sensitivity of around 1.116 dB/gm/dL over the glucose concentration range from 0-10 gm/dL. The presence of minimum 0.008 gm/dL glucose level in urine could be detected using the proposed technique; whereas the traditional Gaussian-beam based technology could detect the presence of around 0.032 gm/dL of the same. Due to the high sensing performance, the present paradigm can be a useful tool for monitoring the elevated glucose levels which is responsible for triggering Diabetes Mellitus.

1. Introduction

The physical characteristics of the urine including the refractive index, viscosity and specific gravity are affected by the glucose concentration contained in it. In general, glucose is not found in the urine under normal human body condition. Though, glucose can often be found in a limited range from 0-15 mg/dL and this level is named as 'Renal Threshold of Glucose (RTG)' [1]. However, due to kidney and liver dysfunction; blood glucose level increases and the presence of glucose is seen in the urine. This disease is called as 'Glycosuria' [1]. According to the World Health Organization (WHO), there had been roughly 422 million people living with diabetes globally in 2014 [2]. WHO also predicts that diabetes will be the world's seventh leading disease. The accumulation of glucose in the urine is also responsible for gestational diabetes which mostly occurs because of changes in the filtration mechanism of the kidney.

Till date, diverse approaches have been attempted towards the detection of glucose levels in human body. Among them, the traditional invasive method examines of the glucose content in blood as collected by the process of pinpricking [3, 4]. Another technique uses "Test strips" where the strip colour will be altered by the amount of urinary glucose, but the results may be erroneous due to the
contamination from air/finger and light source interference [4]. The method of fluorescent signal analysis is not label-free and also requires costly equipment for the photon detector [5, 6]. Besides this, the performance of surface plasmon resonance-based sensor is strongly dependent on the deposited metal's stability [7].

In recent times, beam shaping has been an alluring theme which renders non-Gaussian beam profiles like Bessel-Gauss, Mathieu-Gauss, Airy, Laguerre–Gaussian, and Vortex. With this enticing arrival of non-diffracting beams, the Vortex beam has captivated an immense interdisciplinary attention and has made prodigious advances in optical manipulation, atom optics, and non-linear optics [8, 9]. In our prior works, the benefits of a \( \theta \text{th} \) order Bessel-Gauss beam and Vortex beam have already been explored in the usage of fiber-optic sensing [10-19].

In this report, the authors introduce a new fiber-optic probe for the urinary glucose detection by injecting a Vortex beam inside a No-core fiber. The analysis indicates that such glucose level in urine contributes to changing the sample's refractive index and therefore leads to a change in the optical power loss. By following the classical Wave optics, proper excitation of different linearly polarized modes in the sensing system was evaluated. Moreover, a Beam propagation method is employed in Mode Solutions simulation tool (Lumerical Inc) to envisage the transmission property of the sensing device. The corresponding simulation outcomes have proved the bettered sensitivity of the proposed sensing device in comparison to the conventional one utilizing the Gaussian beam.

2. Proposed Sensing Scheme

In this work, the realization of an ultra-sensitive fiber-optic sensor for detecting the urinary glucose concentration has been achieved by exploiting a special class of Laguerre-Gaussian beam called ‘Vortex beam’. As illustrated in Figure 1, the suggested method of sensing includes a No-core fiber (NCF) and two specially engineered air-core fibers (called as ‘Vortex fiber’). The unique structural characteristic of this NCF is used to mold the evanescent field at the core-cladding boundary. Here, the Vortex fiber is used for guiding the Vortex beam up-to the sensing region so that effective higher order modal excitation can be achieved inside the NCF [11, 20, 21]. Initially, a Gaussian beam from a single-mode pigtailed laser diode (1550 nm) impinges on the spiral phase plate. By controlling the phase pattern of the spiral phase plate, the input Gaussian beam is allowed to modulate with some phase delays which transforms to a Vortex beam having a dark central spot with an adjacent luminous area. Afterward, this Vortex beam is focused by using a convex lens and the focused beam is coupled inside the waveguide structure with the help of an input fiber coupler. When the NCF is submerged into the sensing liquid, the transmitted optical power will vary because of variations in the cladding refractive index as produced by different urine samples. Finally, the transmitted output power is collected by the use of an output fiber coupler and is read-out from an optical power meter.

In this study, we have conducted the theoretical investigation with the following finite-power input Vortex beam [11]:

\[
E_x(r, \theta, 0) = \left(\frac{r\sqrt{2}}{w}\right)^L L_0^L\left(\frac{2r^2}{w^2}\right)e^{-\frac{r^2}{w^2}}e^{-iL\theta} 
\]

(1)

Where \( r \) is the radial and \( \theta \) is the azimuthal coordinate. The parameter \( L_0^L(x) \) and \( w \) is the Laguerre function and beam waist, respectively. The factor \( L \) signifies the order of Vortex beam that determines the helicity of the beam throughout the fiber transverse plane.
Figure 1. Design of the proposed glucose sensor by shining a Vortex beam.

The power coupling coefficient signifies the coupling of energy from the input field to different order modes as supported by the multimode fiber. To estimate the modal excitation inside the NCF, the power coupling co-efficient for various linearly polarized \((LP_{\mu,v})\) mode can be estimated by doing the overlap integration between the applied input field \((E_i(r, \theta))\) and the electric field of \(LP_{\mu,v}\) mode \((E_{\mu,v}(r, \theta))\) as \([22]\):

\[
\eta_{\nu,\mu} = \frac{\int_0^{2\pi} \int_0^\infty |E_i(r, \theta)|^2 E_{\nu,\mu}(r, \theta) rdrd\theta}{\int_0^{2\pi} \int_0^\infty |E_i(r, \theta)|^2 rdrd\theta \int_0^{2\pi} \int_0^\infty |E_{\nu,\mu}(r, \theta)|^2 rdrd\theta} \tag{2}
\]

The suffixes \(\nu\) and \(\mu\) correspond to the indexes for the radial and azimuthal components, respectively. It is noteworthy that as the injected Vortex beam possesses topological charge \((L)\) of 6; thus the modal fields of NCF are also influenced by the azimuthal variation of \(\mu = L = 6\). This implies that only those modes which are having the azimuthal component of \(\mu = \pm 6\) will be energized inside the NCF. Also, there will be excitation of corresponding \(LP_{-\mu,v}\) mode in combination with \(LP_{\mu,v}\) mode which is analogous but having the reverse polarity. Under such condition, the modal fields of NCF can be described as \([23]\):

\[
E_{\nu,6}(r, \theta) = \frac{J_6(\frac{u_{\nu,6}r}{a})}{J_6(u_{\nu,6})} \cos(6\theta); \quad r \leq a
\]

\[
= \frac{K_6(\frac{w_{\nu,6}r}{a})}{K_6(w_{\nu,6})} \cos(6\theta); \quad r > a \tag{3}
\]

Where \(u_{\nu,6}\) and \(w_{\nu,6}\) is the normalized transverse wave number inside the core and cladding, respectively. Further, Eq. 2 can be explicitly solved as:
Due to the multimodal property of NCF, the electrical field distribution can be formulated as the summation of all propagating modes as:

\[
E(r, \theta, z) = \sum_{v=1}^{N} n_{v,6} E_{v,6}(r, \theta) \exp(i\beta_{v,6}z)
\]

(5)

Where \( n_{v,6} \) can be directly calculated through \( n_{v,6} = \sqrt{\eta_{v,6}} \) and \( \beta_{v,6} \) implies the propagation constant.

The total transmission loss of the sensor can be represented as [24]:

\[
L_z(z) = 10\log_{10} \left( \sum_{\mu=L}^{ \lambda } \sum_{v=1}^{N} c_{v,\mu}^2 \exp(-i\beta_{v,\mu}z) \right)^2
\]

(6)

Based on the concept of multimode interference, the self imaging length \( (L_z) \) can be formulated by combining the effect of center wavelength \( (\lambda) \), refractive index of NCF \( (n_{v,6}) \), diameter of NCF \( (D) \), and penetration depth of evanescent field \( (P) \) as [24]:

\[
L_z = \frac{4n_{v,6}(D+2P)^2}{\lambda}
\]

(7)

Here, the value of penetration depth is decided by the refractive index of cladding. In our sensing technique, various urine samples with different glucose concentration behave like a cladding layer of the NCF. As mentioned in Eq. 7, the self image length tends to shift horizontally with varying cladding material as caused by the various urinary glucose levels. Owing to the higher energy density at the edge of Vortex beam, energy coupling in the higher order modes takes place. The quantity of energy flows in the sensing medium also increases for these higher order modes as they are having extended decaying energy tail. As a result, an efficient evanescent coupling and major modulation are detected in response to the urinary glucose concentration. Therefore, the designed Vortex beam-based sensor has an exceptionally high sensitivity as evaluated against the standard Gaussian beam-shined sensor.

3. Results & Discussions

By exploiting the computer-aided simulation, we have plotted the power coupling coefficients for different linearly polarized modes as supported by a 125 micron NCF having a refractive index of 1.4446. The wavelength of single-mode diode laser source is presumed as 1550 nm and the value of axicon apex angle is taken as 130°. The beam waist for both Gaussian and Vortex beam is considered as 7 μm. In this research, five different urine samples with glucose concentration of 0 gm/dL, 0.625 gm/dL, 1.25 gm/dL, 2.5 gm/dL, 5 gm/dL, and 10 mg/dL have been considered which corresponds to a refractive index of 1.335, 1.336, 1.337, 1.338, 1.341, and 1.347, respectively [1]. As revealed in Ref. [11], a Vortex beam can be guided by appropriately selecting the refractive index profile of the air-core Vortex fiber. It is inferred from Fig. 2(a) that the four most dominating modes for Gaussian beam excitation are the \( LP_{0,3}, LP_{0,4}, LP_{0,5}, LP_{0,6} \) with relative power coupling coefficient of
0.131, 0.147, 0.144, and 0.127, respectively. Notably, the four most dominating modes for Vortex beam excitation are the $LP_{6,6}, LP_{6,8}, LP_{6,6}$ (as shown in Fig. 2(b)) having estimated power coupling coefficient of 0.077, 0.099, 0.095, and 0.078, respectively. Moreover, for each $LP_{\mu,v}$ mode, there will be excitation for corresponding $LP_{-\mu,v}$ mode with same power coupling coefficient value but with opposite polarity.

![Figure 2](image1.png)

**Figure 2.** Variation of power coupling coefficient with mode order when the input is (a) Gaussian beam and (b) Vortex beam of order 6. Inset shows the four most dominating modes.

By following Eq. 5, the lateral field profiles are simulated subjected to various propagating distances along the NCF and are plotted in Figure 3. It is apparent that the lateral field distribution at self-imaging length ($z=59.139$ mm) is identical to the input Vortex mode, when the glucose content in urine sample is 10 gm/dL.

![Figure 3](image2.png)

**Figure 3.** Simulated Electric field profiles at (a) $z=5$ mm, (b) $z=40$ mm, and (c) $z=59.139$ mm (self-imaging length)
To envisage the transmitting property of the proposed device, we have simulated the proposed framework in Mode solution software (Lumerical Inc) by using an Eigenmode expansion solver. A perfectly matched layer (PML) boundary condition is incorporated in our study with a simulation mesh accuracy of 0.1 μm. The resultant field evolution and the interference carpet inside the NCF have been rendered in Figure 4 when the glucose content in urine sample is 10 gm/dL.

Figure 4. Transmission and propagation of Vortex beam through the waveguide structure when the glucose content in urine sample is 10 gm/dL.

Figure 5 illustrates the simulated transmission loss with respect to propagation distance along the NCF when the glucose concentration is 10 gm/dl. It is discerned that the suggested fiber-optic glucose sensor manifests a greater transmission loss of about -127.8 dB since there are several higher order modes within the NCF. But, it is worth mentioning that commercially available Femtowatt level optical power meter can measure low power accurately up to -100 dB. To comply with practical feasibility, the length of NCF will be chosen in such a way so that the transmission loss or sensor response lies within the range of minimum detectability of optical power meter.

In vision of Figure 6, it is anticipated that the transmission loss for the proposed glucose sensor fluctuates from -80.13 dB to -68.77 dB while the glucose level in urine sample varies from 0-10 gm/dL. On the contrary, a narrower discrimination range from -38.72 dB to -35.59 dB is observed for the Gaussian beam-based sensor. Through the study of the curve-fitting, a sensitivity of ~1.116 dB/gm/dL is obtained for the proposed glucose sensor. Whereas, it is monitored that the conventional glucose sensor (using the Gaussian beam) has a much lower sensitivity of ~0.305 dB/gm/dL. Thus, our sensing system envisages a 3.65-fold higher sensitivity. The sensing resolution can also be calculated by using a high-resolution optical power meter with 0.01 dB accuracy and is
presented in Table 1. Because of this excellent sensitivity, the proposed sensor provides a new degree of freedom that finds intense implication in the field of biomedical engineering.

![Image of graph showing sensor response]

**Figure 6.** Sensor response when the NCF length is 29.30 mm.

| Length of NCF (mm) | Sensitivity (dB/gm/dL) | Sensing Resolution (gm/dL) |
|-------------------|------------------------|---------------------------|
|                   | Conventional Sensor    | Proposed sensor           | Conventional Sensor    | Proposed sensor |
| 29.30             | 0.305                  | 1.116                     | 0.032                  | 0.008          |

**Table 1.** Comparative analysis of sensor’s performance

4. **Conclusion**

The research mentioned in this paper offers the possibility of using Vortex beam in fiber-optic sensing to detect the glucose levels in urine. Such type of non-Gaussian beam will maximize the sensor's performance by exciting many higher order modes within the framework of fiber. The enticing aspect of such system is that it challenges the traditional wisdom and significantly improves the sensitivity without the need of any intricate fabrication techniques like in tapering, bending etc. With the intervention of Mode Solution software, the modal interference and the transmission of the overall sensor structure was visualized by using a Beam propagation method. While the sensing length is 29.30 mm, the proposed sensor proffers a maximum sensitivity of 1.116 dB/gm/dL which is 3.65 times greater than the standard Gaussian beam-based sensor. The sensor being proposed is extremely sensitive as it can detect even minute variation in the glucose concentration present in urine. Such novel urinary glucose sensor is very much effective in the diagnosis of diabetes or diseases associated with liver and kidney failure.

**Acknowledgment**

The authors would like to thank the Department of Electrical & Electronics Engineering, CMR Institute of Technology, Bengaluru, India, for the valuable motivation and support.

**References**

[1] Sharma P and Sharan P 2015 *IEEE Sensors J.* 15 1035.
[2] Robinson S, Dhanlaksmi N 2017 *Photonic Sensors* 7 11–19
[3] Bratlie K M, York R L, Invernale M A, Langer R, and Anderson D G 2012 Advanced Healthcare Materials 1 267.

[4] Bruen D, Delaney C, Florea L, and Diamond D 2017 Sensors(Basel) 17 1866.

[5] Ganesh A B and Radhakrishnan T K 2005 Sensors & Transducers 60 439.

[6] Rosenzweig Z and Kopelman R 1996 Sensors and Actuators B: Chemical 36 475.

[7] Chiu M H, Wand S F, and Chang R S 2005 Opt. Lett. 30 233.

[8] Yao A M and Padgett M J 2011 Advances in Optics and Photonics 3 161.

[9] Banerji A, Singh R P, Banerjee D, and Bandhopadhyay A 2016 Optics Communications 380 492.

[10] Datta A and Saha A 2017 J. Opt. Soc. Am. B 34 1327.

[11] Datta A, Saha A, and Shukla A 2017 J. Opt. Soc. Am. A 34 2034.

[12] Datta A, Babu A M, and Saha A 2019 Optical Engineering 58 056112.

[13] Saha A, Datta A, and Kaman S 2018 Optical Engineering 57 036118.

[14] Datta A and Saha A 2020 Optik 218 165006.

[15] Saha A, Datta A, and Kaman S 2016 Proc.of International Conference on Fiber Optics and Photonics (Optical Society of America) paperTh4F.

[16] Saha A and Datta A 2017 Proc of Conference on Lasers and Electro-Optics /Pacific Rim (Optical Society of America) paper s1796.

[17] Datta A and Saha A 2020 Proc of 2nd International Conference on Innovative Mechanisms for Industry Applications (IEEE Xplore, 2020) pp. 515-519.

[18] Datta A and Saha A 2020 AIP Conference Proceedings 2281 020023.

[19] Datta A, Karmakar S, and Saha A 2020 Proc. SPIE 11525 115250F.

[20] Ramachandran S and Kristensen P 2013 Nanophotonics 2 455.

[21] Gregg P, Kristensen P, and Ramachandran S 2015 Optica 2, 267.

[22] Mohammed W S, Mehta A, and Johnson E G 2004 J. Light. Technol. 22 469.

[23] Ghatak A and Thyagarajan K 1998 An Introduction to Fiber Optics ( Cambridge University Press).

[24] Wang Q, Farrell G, and Yan W 2008 J. Light. Technol. 26 512.