Abstract—Molecular Communication (MC) is a bio-inspired communication method based on the exchange of molecules for information transfer among nanoscale devices. MC has been extensively studied from various aspects in the literature; however, the physical design of MC transceiving units is largely neglected with the assumption that network nodes are entirely biological devices, e.g., engineered bacteria, which are intrinsically capable of receiving and transmitting molecular messages. However, the low information processing capacity of biological devices and the challenge to interface them with macroscale networks hinder the true application potential of nanonetworks. To overcome this limitation, recently, we proposed a nanobioelectronic MC receiver architecture exploiting the nanoscale field effect transistor-based biosensor (bioFET) technology, which provides noninvasive and sensitive molecular detection while producing electrical signals as the output. In this paper, we introduce a comprehensive model for silicon nanowire (SiNW) FET-based MC receivers by integrating the underlying processes in MC and bioFET to provide a unified analysis framework. We derive closed-form expressions for the noise statistics, the signal-to-noise ratio (SNR) at the receiver output, and the symbol error probability (SEP). Performance evaluation in terms of SNR and SEP reveals the effects of individual system parameters on the detection performance of the proposed MC receiver.

Index Terms—Molecular communication, receiver, SNR, SEP.

I. INTRODUCTION

Molecular communication (MC) defines the technology where molecules are used to encode, transmit, and receive information. MC is a biocompatible communication method providing efficient and reliable information transfer between living entities at nanoscale. Hence, it has been regarded as the most promising paradigm to realize nanonetworks by enabling the communication among nanoscale devices, i.e., nanomachines [1]-[3].

MC paradigm has been extensively studied from various aspects. A large body of work has been devoted to modeling the MC channel from information theoretical perspective [4]-[5], designing modulation schemes [6] and developing communication protocols [7] and optimal detection algorithms compatible with MC [8]. While adapting the tools of conventional communication techniques to MC, these studies have mostly ignored the physical design of system components such as transmitter and receiver. Besides these contributions, there are only a few studies focusing on the physical design of communicating units. For example, in [9], the authors propose a biotransceiver architecture which can realize transmitting, receiving and basic processing operations based on the functionalities of genetically engineered bacteria. Similarly, a layered architecture for MC based on functionalities that can be acquired from bio-engineered bacteria, such as molecular motors, is presented in [7]. Furthermore, numerous studies have investigated MC-based networks of bacteria colonies [10], [11]. Common to these studies is the assumption that the MC nanonetwork consists of nanomachines, which are entirely made up of biological components.

Although designing nanomachines with only biological components provides the advantage of biocompatibility, which is crucial for biomedical applications, it has also numerous disadvantages that restrict the application domain of nanonetworks. First of all, very low computational capacities of biological devices, which are evident from [9], limit the speed of information processing, and thus, the extent of the tasks that these nanomachines can undertake in a nanonetwork application. This limitation points out a major discrepancy between the envisaged applications of nanonetworks [1], most of which require the implementation of complex communication protocols and algorithms, and the very limited processing capabilities of the biological devices. Another critical drawback of the entirely biological device architectures is that they are operational only in in vivo applications, i.e., applications within living organisms like human body. Moreover, they do not allow the incorporation of a noninvasive and seamless interface between molecular nanonetworks and macroscale cyber networks such as the Internet. This is one of the key challenges to realization of Internet of nanotings (IoNT), which is a visionary concept that promises for groundbreaking medical and environmental applications [12], [13]. Furthermore, the current state-of-the-art of synthetic biology research is not advanced enough to take the full control over the functionalities of living cells to design engineered cells that can operate in one of the envisioned nanonetwork applications [14].

The discrepancies and the challenges pertaining to the entirely biological architectures have led us to consider different design solutions. In our recent review of design options [15], interfacing the biochemical environment, where molecular messages propagate, with a nanobioelectronic architecture, that can provide fast information processing and wireless interface with macroscale networks has been revealed to be the most promising and feasible solution. This nanobioelectronic design approach implies transmitters that can release molecular messages upon being triggered by electrical signals, and...
receivers that can detect molecular messages and transduce them into electrical signals for further processing.

In [15], for implementing a nanobioelectronic MC receiver, we have proposed the use of FET-based biosensors, i.e., bioFETs, optimized from MC theoretical perspective. BioFETs have emerged as promising analytical tools, which enable the label-free electrical sensing of target molecules [16]. We have shown that they satisfy the basic requirements of an MC receiver such as the capability of precise, continuous and noninvasive detection of molecular concentrations. Use of novel nanomaterials, such as nanowires, carbon nanotube (CNT) and graphene, has enabled them to be designed with nanoscale dimensions [17], [18]. Transduction of biochemical concentrations into electrical signals in bioFETs could provide a fast in-device information processing for the MC receiver. It could also enable the design of a seamless interface between the nanobioelectronic receiver and macroscale networks by means of electromagnetic signals.

Towards the goal of optimizing bioFETs as MC receivers, one of the major challenges pointed out in [15] is the lack of a comprehensive analytical model for bioFETs. Although there are a vast number of experimental works reported for bioFETs and a few theoretical studies focusing on the noise processes effective on their operation [19], [20], none of them is able to entirely capture the physical processes in stochastic sensing of molecular concentrations.

In [21], we introduced deterministic and noise models for SiNW FET-based MC receiver antenna, and derived the SNR at the antenna output by neglecting the effects of the ligand transport dynamics. In this study, we revisit and extend this model by accounting for the effects of the MC channel on the receiver operation. The resultant unified model capturing all of the stochastic processes regarding the MC and biosensing enables the derivation of closed-form expressions for the decision statistics at the electrical end of the receiver, and thus, provides a complete analytical framework for the performance analysis and design optimization of bioFETs as MC receivers. The major contributions of this study can be explained as follows:

- We present a comprehensive model of SiNW bioFETs by integrating the contributions of all the noise processes affecting the electrical output of the devices. As such, the model also contributes to the nanobiosensor literature.
- Based on the recent findings in biophysics literature, we incorporate the spatial and temporal correlation effects resulting from finite-rate diffusion into the stochastic ligand-receptor binding process, and obtain the statistics of the diffusion-influenced binding noise at the receiver side. This development leads to a combined channel and receiver noise model, which can be applicable for MC systems with various receiver architectures, including the entirely biological ones, and channel geometries. The current studies in MC investigates the reception problem only in diffusion-limited or reaction-limited regime. Up to now, a vast majority of MC studies focus on the diffusion-limited regime by assuming that the receiver with infinite reaction rates could perfectly count the number of molecules in an arbitrarily defined reception space [4].
- There are also a few studies, such as [22], [23], that focus on the reaction-limited case and derive the statistics of ligand-receptor binding process, which is Markovian in this regime. However, both approaches neglect the extended correlations in the reception process resulting from the finite transport and reaction rates.
- In another modeling approach based on reaction-diffusion master equation (RDME), which divides the channel into discrete voxels and provides a combined channel-receiver model by taking the propagation and ligand-receptor binding process as a continuous-time Markov process (CTMP) [26], [27]. Although the derived model is more comprehensive in its approach to small-scale systems (limited in size and number of information molecules), it does not allow the derivation of closed-form expressions for the statistics of transport-influenced reception noise. The unified noise model developed in this paper overcomes this limitation with a steady-state assumption in the received concentration signal.
- We provide the first unified channel and receiver model by incorporating the SiNW bioFET into an MC system as a receiver. The use of electronic chemical sensors, similar to bioFETs, for MC receiver has been previously addressed in [28]. In that study, the authors have designed an experimental testbed for MC with macroscale dimensions and use a metal oxide semiconductor sensor for detecting chemical messages encoded into isopropyl alcohol. Based on the experimental data obtained using the testbed, they developed a combined channel and receiver model [29]. Although similar trends are observed in receiver detection performance, the analytical framework developed in this study is based on a more comprehensive and unified approach, thus, better suited for the purpose of optimizing biosensors from MC theoretical perspective.
- We obtain closed-form expressions for the deterministic response of the receiver, the statistics of the noise processes and the SNR at the receiver output. We also investigate an MC testbed that utilizes M-ary Concentration Shift Keying (M-CSK) [6] for modulation at the transmitter side and Maximum Likelihood (ML) method for detection at the receiver side, and derive an analytical expression for the corresponding SEP.
- Evaluating the receiver in terms of SNR and SEP, we reveal the effects of individual system parameters on the detection performance. The obtained results underline the feasible optimization pathways that can be targeted to improve the receiver.

The remainder of the paper is organized as follows. In Section II, we describe bioFETs and explain their operation principles. We develop the model of SiNW FET-based MC receivers in Section III. In Section IV, we derive the SEP for an MC testbed that employs the nanobioelectronic receiver and utilizes M-CSK scheme for modulation. The performance evaluation results are presented in Section V. Finally, the concluding remarks are given in Section VI.
II. PRINCIPLES OF BIOFETS

Operation principles of bioFET, which is the basis of the nanobioelectronic MC receiver architecture, are similar to the ones of the conventional FETs. In conventional FET type transistors, current flows from the source electrode to the drain electrode through a semiconductor channel, conductance of which is controlled by the electric field created by the potential applied on the gate electrode. Conductivity is proportional to the density of the carriers accumulated in the channel, and the variations of the electric field resulting from the additional surface potential is reflected to the changes in the voltage-current characteristics between the drain and the source electrodes.

BioFETs slightly differ from the conventional FETs by including an additional biorecognition layer that is capable of selectively binding the target molecules \[16\]. This layer is composed of a high number of receptor molecules tethered on the surface of the FET channel, and replaces the gate electrode of conventional FETs, as shown in Fig. 1. Binding of ligands to the surface receptors results in accumulation or depletion of the carriers in the semiconductor channel due to the field effect generated by the intrinsic charges of the bound ligands. Hence, the ligand binding modulates the channel conductance and current, and thus, the output current becomes a function of the ligand density and the amount of ligand charges. Label-free, continuous and in situ sensing of the molecules by not requiring any complicated processes such as the use of macroscale equipments for readout and processing operations makes bioFETs a natural candidate for the architecture of MC receivers.

Several ligand-receptor pairs, e.g., antibody-antigen, aptamer-natural ligand, natural receptor/ligand, have proven suitable for bioFETs \[30\]. Various types of semiconductors, such as SiNW, Carbon NanoTube (CNT) and graphene, can be used as the FET channel, i.e., transducer channel \[16\]. The basics of biorecognition and transducing operations and the noise processes do not fundamentally differ based on the type of the ligand-receptor pair and the semiconductor channel. However, the literature is currently dominated by the studies focusing on SiNW bioFETs. One inherent drawback of SiNW bioFETs is the thin oxide layer built up around the SiNW surface, which deteriorates its detection performance \[31\]. Therefore, researchers are in search for other nanomaterials suitable for application in biosensing. For example, metal-oxide nanowires, such as SnO$_2$, ZnO, In$_2$O$_3$, are rapidly proving to be useful in nanoscale biosensing applications as the active channels of thin film biotransistors \[32\]. Though, in several cases, they could outperform the SiNW counterparts in terms of sensitivity and limit of detection, their fabrication is currently constrained by bottom-up methods, which limits the controllability of their size and doping characteristics \[31\]. In contrast, SiNW provides a wider range of fabrication options including well-developed lithographic top-down methods, which allows their seamless integration into the current fabrication technologies \[33\]. Together with a more established literature, easier and controllable fabrication of SiNWs lead us to focus on SiNW bioFETs in this study to develop a model for bioFET-based MC receiver.

III. SiNW FET-BASED RECEIVER MODEL

A. Model Description

We consider a time-slotted molecular communication system between a single transmitter receiver pair, which are assumed to be perfectly synchronized with each other in terms of time. The system utilizes M-ary concentration shift keying (M-CSK) modulation such that information is encoded into the concentration, i.e., the number, of molecules. Given that the input alphabet is $\mathcal{M} = \{0, 1, \ldots, M-1\}$, to send the symbol $m \in \mathcal{M}$ for the $k$th time slot, the transmitter releases $N_m$ molecules at the beginning of the signaling interval, i.e., at time $t_k = kT_s$, where $T_s$ is the slot duration, i.e., the symbol period.

For the propagation medium, we consider a simple straight microfluidic channel with a rectangular cross-section, as shown in Fig. 2. The transmitter is assumed to be located at the entrance of the channel. A SiNW FET-based MC receiver is considered to be located at the bottom of the microfluidic channel at position $x = x_R$, with its SiNW transducing channel covered by the oxide layer and surface receptors, which are directly exposed to the information molecules, i.e., ligands, of varying concentration. The receiver samples the concentration of ligands flowing over its surface based on ligand-receptor binding kinetics \[24\]. A fluid flows unidirectionally from the transmitter to the receiver location in the microfluidic channel such that the transmitted ligands are propagated towards the receiver through advection and diffusion. The trajectories of each molecule along the propagation channel are assumed to be independent of each other.
Employing microfluidic channels for the propagation medium is prevalent in both MC [34], [35], and biosensing literature [36], [37]. We will frequently make use of the results obtained in these studies to derive analytical expressions for the performance metrics of the SiNW receiver.

In the considered scenario, ligands are not absorbed by the receiver, instead they temporarily bind to the surface receptors and unbind after a random amount of time. This characteristic of the ligands together with their long propagation times and spread over the x-axis through diffusion make the propagation channel have a memory, that may result in intersymbol interference (ISI) [38]. ISI can be overcome by selecting the symbol period \( T_s \) sufficiently long, or employing auxiliary enzymes in the channel that degrade the information molecules in the environment after the detection, as proposed in [39]. To simplify the derivation of the receiver model, without loss of generality, we assume that the propagation channel is memoryless, thus, we neglect ISI.

The block diagram of the communication system including the SiNW FET-based MC receiver is shown in Fig. 3. The receiver’s operation can be described by the operations of three consecutive functional units. The Biorecognition Unit (BU) constitutes the interface of the receiver with the communication channel and is responsible for selectively sensing the concentration of ligands. In the Transducer Unit (TU), the ligands, which stochastically bind the surface receptors, modulate the gate potential of the FET through the field effect resultant from their intrinsic charges. In the Output Unit (OU), the modulated gate potential is immediately reflected into the current flowing through the SiNW channel between the drain and the source electrodes of the FET.

In the following, we determine the characteristics of the receiver input signal by modeling the propagation inside the microfluidic channel, and then model each functional unit of the receiver individually to obtain the input-output relations and derive the statistics of the additive noise processes effective on the output current.

### B. Molecular Transport in Microfluidic Channel

Transport dynamics of ligands inside the microfluidic channel can be described by the advection-diffusion equation. The fluid flow, which may be created by a pressure difference between the two ends of the channel, is taken as laminar, steady, and unidirectional along the channel’s longitudinal axis, i.e., x-axis [35]. Assuming that a relatively low number of ligands reversibly react with the surface receptors and do not substantially change the concentration in the channel, the propagation of the ligands throughout the channel is handled as a one-dimensional advection-diffusion problem such that the ligand concentration \( \rho \) and the fluid velocity \( u \) are represented by their average over the channel’s cross section [37]. Therefore, the concentration and fluid flow are invariant along the y- and z-axis. One dimensional advection-diffusion equation along the direction of the fluid flow, \( \vec{x} \), can be written as

\[
\frac{\partial \rho(x,t)}{\partial t} = D \frac{\partial^2 \rho(x,t)}{\partial x^2} - u \frac{\partial \rho(x,t)}{\partial x},
\]

where \( \rho(\vec{x},t) \) is the average ligand concentration at position \( \vec{x} \) and time \( t \), and \( u \) is the x-axis fluid flow velocity averaged over the channel’s cross section. \( D \) is the effective diffusion coefficient that accounts for the effect of Taylor-Aris type dispersion of ligands [40]. For a channel with rectangular cross-section, it is given by

\[
D = \left( 1 + \frac{8.5u^2h_{ch}^2l_{ch}^2}{210D_0^2(h_{ch}^2 + 2Ah_{ch}l_{ch} + l_{ch}^2)} \right) D_0,
\]

where the intrinsic diffusion coefficient is denoted by \( D_0 \) [35]. \( h_{ch} \) and \( l_{ch} \) are the cross-sectional height and length of the channel, respectively. Transmitter is assumed to be a planar ligand source located at \( x = x_T \) and release a preset number of ligands \( N_m \) at time \( t_k = kT_s \) into the channel for representing symbol \( m \). Assuming that the ligands are uniformly distributed over the channel’s cross-section at the instant of release, the transmitter signal can be represented by an impulse scaled by the surface concentration:

\[
\rho_m(x,t) = \frac{N_m}{A_{ch}} \delta(x-x_T,t-t_k),
\]
where $A_{ch} = h_{ch} \times l_{ch}$ is the cross-sectional area of the channel. Since we neglect ISI, to model the propagation, it is sufficient to consider only one signaling interval, say $k = 0$. Taking the transmitter location as $x_T = 0$, with the given initial condition [5], the solution of one dimensional advection-diffusion equation (1) is given by [40] as follows

$$\rho_m(x,t) = \frac{N_m/A_{ch}}{\sqrt{4 \pi D t}} \exp \left( - \frac{(x-ut)^2}{4Dt} \right). \quad (4)$$

Note that although the initial condition is represented by a surface concentration, the resulting solution is given in terms of volumetric concentration of ligands.

C. Received Signal

The SiNW FET-based receiver is considered to be placed at the bottom of the microfluidic channel and located along the transverse axis perpendicular to fluid flow. We assume that the length of the SiNW, $l_R$, is equal to the cross-sectional length of the channel, i.e., $l_R = l_{ch}$, and the drain, source and gate electrodes of the receiver are buried inside the channel walls and do not affect the fluid flow and ligand propagation. The radius of the SiNW is denoted by $r_R$, and the position of the radial axis of the SiNW is taken as the center position of the receiver and denoted by $x_R$.

As can be inferred from [4], the concentration profile of the ligand plug is Gaussian at any observation time. The plug disperses and its peak concentration attenuates as the plug is transported by the fluid flow along the channel. We define the delay between the transmitter and the receiver as the time it takes for the plug’s peak concentration to arrive at the center position of the receiver $x = x_R$:

$$t_D = \frac{x_R}{u}. \quad (5)$$

Following [37], we define an effective plug width $w_p$ in the spatial domain, as the width that encloses 95% of the Gaussian area of $\rho_m(x,t_D)$. This corresponds approximately to the four times the standard deviation of the distribution, and can be given by

$$w_p = 4 \sqrt{2Dt_D}, \quad (6)$$

which is independent of the ligand concentration released from the transmitter.

We neglect the attenuation and dispersion of the ligand plug during its passage over the receiver surface, as in [37]. We also assume that all of the points on the receiver surface are exposed to the same concentration of ligands. This assumption is reasonable since the effective width of the receiver, $w_R \approx \pi l_R$, along the flow direction is very small, i.e., typically on the order of 30 nm, compared to the effective plug width $w_p$, typical values of which are on the order of 100 $\mu$m (for example, in the case of $x_R = 1$ mm and $u = 10 \mu$m/s, $w_p \approx 565 \mu$m). Thus, the ligand concentration seen by the receiver at time $t$ can be given by

$$\rho_m(x,t) = \rho_{R,m}^{\max} \exp \left( -8 \frac{(x-R - ut)^2}{w_p^2} \right), \quad (7)$$

where $\rho_{R,m}^{\max}$ is the peak concentration at time $t_D$ at the receiver location

$$\rho_{R,m}^{\max} = \rho_m(x_R,t_D) = \sqrt{\frac{8}{\pi}} \frac{N_m}{A_{ch} w_p}. \quad (8)$$

Given the effective plug width $w_p$, the ligand plug is assumed to complete its passage in a time duration of $w_p/u$. During the passage of the plug, we assume, as in [37], that the receiver is exposed to a stationary concentration $\rho_{R,m}$.

$$\rho_{R,m} = \langle \rho_m(x,t) \rangle = \int_{t_D-w_p/2u}^{t_D+w_p/2u} \rho_m(x,t)dt$$

$$= \rho_{R,m}^{\max} \sqrt{\frac{\pi}{8}} \text{erf} \left( \frac{\sqrt{2}}{2} \right), \quad (9)$$

for $t \in [t_D - w_p/2u, t_D + w_p/2u]$, which is the time average. In the considered scenario, the receiver samples the ligand concentration at a single time instant in this interval, e.g., at $t = t_D$. Therefore, the received signal is taken as equal to $\rho_{R,m}$, when the transmitter sends symbol $m$.

D. Biorecognition Block and Binding Noise

We start modeling the biorecognition block by first investigating the ligand flux to the receiver surface. In some MC studies, the transport of ligands is assumed to be fast enough to assure that the ligand-receptor binding kinetics is reaction-limited, i.e., governed simply by the intrinsic binding and unbinding rates of the reactants, thus, the successive samples of the concentration taken by the receptors becomes uncorrelated [24]. The opposite case, i.e., transport-limited kinetics, is more prominently accepted in MC studies which assume that the receiver is perfect observer that counts every single molecule that enters into a defined reception volume. This corresponds to an infinite binding and unbinding rates, thus, a continuous sampling of the concentration signals brings
along an additive counting noise (rather than the ligand-receptor binding noise), correlation of which is governed only by the transport rate, e.g., diffusion coefficient, of the ligands. However, this is not always the case, because we usually observe binding/unbinding and transport rates comparable to each other, i.e., the receptor-ligand binding kinetics is neither reaction-limited nor transport-limited. This problem has been recently addressed by several studies, which arrive at the same correlation time depending on both transport and reaction rates. To make use of the analytical results obtained in these studies, we first investigate the transport of ligands to the receiver surface.

Since the SiNW resides on the surface of bulk SiO$_2$, which occludes the ligand flux from the bottom, the ligands flowing over the receptor can be assumed to interact only with the receptors tethered to the top surface of the SiNW, which is covered by a SiO$_2$ layer, as shown in Fig. 4. Thus, we can consider that the receptor has a hemicylindrical interface to the fluid. The transport rate, i.e., flux, of ligands, flowing with a stationary concentration, to a hemicylindrical surface, placed at the bottom of a microfluidic channel with rectangular cross-section is investigated in §6. The authors show that it can be approximated (with an error of $\sim 5\%$) by the flux to a sensor rectangular surface with the same length and the width $w_R = \pi r_R$, providing a closed-form expression in §6:

$$k_T = DL_s \times \begin{cases} 
\left(0.8075 P_s^{1/3} + 0.7058 P_s^{-1/6} - 0.1984 P_s^{-1/3}\right), & \text{if } P_s > 1 \\
2.836 \ln(P_s) \left(1 - \frac{0.092266 P_s}{4.885 - \ln(P_s)}\right), & \text{if } P_s < 1 
\end{cases}$$

(10)

where $P_s$ is defined as

$$P_s = \frac{6Qw_R^2}{DL_{ch}h_{ch}^3},$$

(11)

where $Q$ is the volumetric flow rate, which is given by $Q = u \times A_{ch}$.

The ligands flowing over the receiver surface interact with the surface receptors through ligand-receptor binding mechanism. Assuming that the surface receptors are kinetically non-interacting, i.e., there is no cooperativity among them, the number of bound receptors can be described by the following kinetic scheme,

$$0 \xrightarrow{\alpha_n} 1 \xrightarrow{\beta_1} 2 \xrightarrow{\alpha_2} \ldots \xrightarrow{\alpha_{N_R-1}} N_R \xrightarrow{\beta_{N_R}} 0,$$

(12)

where $N_R$ is the number of surface receptors. In the case that the ligand-receptor binding process is reaction-limited (no influence of ligand transport with infinite transport rate), the state-dependent rates $\alpha_n$ and $\beta_{n+1}$ can be given as follows

$$\alpha_n = (N_R - n)k_1\rho_{R,m},$$

(13)

$$\beta_{n+1} = nk_{-1},$$

(14)

where $k_1$ and $k_{-1}$ are the intrinsic binding and unbinding rates of ligand-receptor pair, respectively, and $n$ denotes the number of bonded receptors. In this reaction-limited case, $\rho_{R,m}$, which is the concentration of ligands flowing over the receptor location, is equal to the concentration of ligands at contact with the surface receptors. When the transport rate cannot be assumed to be infinite, the binding statistics are affected by the transport dynamics. This effect is captured by Eq. (10), which obtains the transport-influenced state-dependent reaction and transport-limited, nor transport-limited. This problem has been recently addressed by several studies, which arrive at the same correlation time depending on both transport and reaction rates. To make use of the analytical results obtained in these studies, we first investigate the transport of ligands to the receiver surface.

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The effective ligand charges on the surface are transduced into a surface potential as follows [19],

\[ \Psi_m = \frac{Q_m}{C_{eq}}, \]  

(27)

where \( C_{eq} \) is the equivalent capacitance of the transducer. As demonstrated in Fig. 5, three capacitances are effective on the transduction of the surface charges: (i) Diffusion layer capacitance, \( C_{DL} \), resulting from the double layer created by the medium counterions accumulated at the interface between oxide layer, i.e., SiO\(_2\) layer, and the electrolyte medium; (ii) the capacitance of the oxide layer; (iii) the SiNW capacitance, \( C_{NW} \), which is again a double layer capacitance caused by the accumulation of carriers to the SiO\(_2\)/SiNW interface [44], [45]. Therefore, the equivalent capacitance can be given by

\[ C_{eq} = \left( \frac{1}{C_{OX}} + \frac{1}{C_{NW}} \right)^{-1} + C_{DL}. \]  

(28)

As we assume a nanowire-on-insulator (NWoI) configuration for the receiver design, individual capacitances can be obtained by considering the SiNW as a hemicylinder with an oxide layer

\[ C_{DL} = (\epsilon_M/\lambda_D)w_Rl_R. \]  

(29)

Similarly, the oxide layer capacitance can be given by

\[ C_{OX} = (\epsilon_OX/t_OX)w_Rl_R, \]  

(30)

where \( \epsilon_{OX} \) and \( t_{OX} \) are the permittivity of the oxide layer. For high values of hole density, e.g., \( p \sim 10^{18} \text{ cm}^{-3} \), corresponding to the linear operation regime of the FET, the double layer capacitance emerged in the NW channel can be obtained as follows [44]

\[ C_{NW} = (\epsilon_{Si}/\lambda_{Si})w_Rl_R, \]  

(31)

where \( \epsilon_{Si} \) is the dielectric permittivity of SiNW, and \( \lambda_{NW} \) is the thickness of the double layer created in the inner surface of the NW, which is given by [44]

\[ \lambda_{NW} = \sqrt{\frac{\epsilon_{Si}k_BT}{pq^2}}. \]  

(32)

Fig. 5. Equivalent circuit model for the transducer of the SiNW FET-based MC receiver [44], [45]. REF denotes the reference electrode.
F. Output Block and 1/f Noise

In the output block, the potential induced at the SiNW/oxide layer interface is reflected into a variation in the current flowing through the SiNW transducer channel. We assume that the p-type FET is operated in the linear (Ohmic) region, with the electrode voltage values satisfying the following conditions [46]:

\[ V_{SG} > |V_T|; \quad V_{SD} \leq V_{SG} - |V_T|, \]  

where \( V_{SG} = -V_{GS} \) is source to gate voltage, \( V_{SD} = -V_{DS} \) is the source to drain voltage, and \( V_T \) is the threshold voltage of the SiNW FET. The voltage values are assumed to be held constant during the receiver operation. In the linear region, the current flowing between the source and drain electrodes is given by

\[ I_{SD} = \mu_p C_{OX} \frac{w_R}{l_R} \left[ (V_{SG} - |V_T|) V_{SD} - \frac{V_{SD}^2}{2} \right], \]  

(34)

where \( \mu_p \) is the carrier (hole) mobility, which depends on the impurity density of the transducer channel [46], and \( l_R = l_{ch} \) as stated before. The partial derivative of the source-drain current with respect to source-gate voltage gives the transconductance of the FET:

\[ g_{FET} = \frac{\partial I_{SD}}{\partial V_{SG}} = \mu_p C_{OX} \frac{w_R}{l_R} V_{SD}, \]  

(35)

A potential created at the electrolyte/oxide layer interface due to the bound ligands acts upon the FET channel in the same way as the gate voltage does in a conventional FET. Therefore, the part of the channel current, \( I_m \), generated by the surface potential, \( \Psi_m \), can be written in terms of transconductance as follows

\[ I_m = g_{FET} \times \Psi_m, \]  

(36)

Combining the equations (34), (18), (19), (36), the mean of the generated output current can be given by

\[ \mu I_m = g_{FET} \Psi_L N_R \left( 1 + \frac{K_{D}A_{ch}w_p}{N_m \text{erf}(\sqrt{2})} \right)^{-1}, \]  

(37)

where \( \Psi_L \) is defined as the surface potential created by the binding of a single ligand:

\[ \Psi_L = \left( q \text{eff} \times N_c^{-} \right) / C_{eq}. \]  

(38)

As in all transistor devices, low-frequency operation of bioFET-based MC receiver is suffered from 1/f noise. Although the origin of flicker noise and its full analytical model are still open issues, there are several models, including the well-known Hooge’s model, that approximate the noise power in frequency domain [47].

In this paper, we use the correlated carrier number and mobility fluctuation model, which provides a more accurate description of the 1/f noise for FET type devices, compared to the Hooge’s model and the carrier number fluctuation model, attributing the source of 1/f noise to the random generation and recombination of charge carriers due to the defects and traps in the SiNW channel resulting from imperfect fabrication. It also accounts for the correlation of the carrier mobility with the carrier number fluctuations [49]. The model expresses the resulting output current-referred noise PSD as

\[ S_{I_m}(f) = S_{V,FB}(f) g_{FET}^2 [1 + \alpha_s \mu_p C_{OX} (V_{SG} - |V_{TB}|)]^2, \]  

(39)

where \( \alpha_s \) is the Coulomb scattering coefficient which depends on the temperature, and \( \mu_p \) is the mobility of the hole carriers, which depends on the impurity concentration of the SiNW channel [48]. \( S_{V,FB} \) is the PSD of the flatband-voltage noise, given as

\[ S_{V,FB}(f) = \frac{\lambda k_BT^2}{w_R l_R C_{OX}^2 |f|}, \]  

(40)

where \( \lambda \) is the characteristic tunneling distance, \( N_{ot} \) is the oxide trap density, i.e., impurity concentration, of the SiNW channel [48]. 1/f noise is independent of the received signals, and shows an additive behavior on the overall output current fluctuations [49]. Theoretically, 1/f noise does not have a low frequency cutoff, and has infinite power at zero frequency. However, in experimental studies with a finite measurement time, a finite variance for 1/f noise is observed. The reason is related to the low frequency cutoff set by the observation time \( T_{obs} \) [50], [51]. Considering that the received molecular signals are at the baseband, to be able to calculate the total noise power, we assume one-year operation time, i.e., \( \sim \pi \times 10^7 \) s, for the antenna such that the low cutoff frequency is \( f_L = 1/T_{obs} \approx 1/\pi \times 10^{-7} \) Hz. At frequencies lower than \( f_L \), the noise is assumed to show the white noise behavior, i.e., \( S_{I_m}(f) = S_{\Delta I_m}(fL) \) for \( |f| < f_L \).

G. Overall Noise PSD and Output SNR

The PSD for the output current fluctuations due to the additive binding noise \( S_{I_{m0}} \) can be written as

\[ S_{I_{m0}}(f) = S_{N_{B,m0}}(f) \times \Psi_L^2 \times g_{FET}^2. \]  

(41)

Including the additive 1/f noise, the overall PSD of the output current referred noise is given by

\[ S_{I_m}(f) = S_{I_{m0}}(f) + S_{I_m}(f). \]  

(42)

Given the noise PSD, the output SNR of the receiver can be computed by

\[ SNR_{out,m} = \frac{\mu I_{m}}{\sigma^2 I_{m}}, \]  

(43)

where \( \sigma^2 I_{m} \) is the output current variance, obtained as follows

\[ \sigma^2 I_{m} = \int_{-\infty}^{\infty} S_{I_m}(f) df. \]  

(44)

IV. GAUSSIAN APPROXIMATION AND SYMBOL ERROR PROBABILITY

A. Gaussian Approximation for Noise Processes

To analytically derive the symbol error probability (SEP), we make some reasonable approximations for the noise statistics. We can expect that a significant number of receptors, e.g., \( > 1000 \), are tethered to the top surface of a SiNW channel. For large number of surface receptors, the binomial distribution given in (17) can be approximated as Gaussian, i.e., \( N_{B,m} \sim \)
\[ \mathcal{N}\left(\mu_{N_{B,m}}, \sigma^2_{N_{B,m}}\right) \text{ with } \mu_{N_{B,m}} \text{ and } \sigma^2_{N_{B,m}} \text{ given in (19) and (20), respectively. Thus, the zero-mean additive binding noise follows the normal distribution } \mathcal{N}\left(0, \sigma^2_{N_{B,m}}\right). \] 

The binding noise keeps the zero-mean Gaussian characteristics when it is passed through the linear filter in the transducing block.

1/f noise is resulting from the bias current flowing through SiNW channel, therefore, it is independent of the binding noise. Although, there has been a long-standing discussion about the 1/f noise statistics \cite{52}, in many well-accepted experimental studies in the literature, it has been reported that 1/f noise can be approximated to follow a Gaussian distribution \cite{53-55}. In this paper, we rely on these reports to provide an analytical expression for the SEP.

Therefore, the overall noise process effective on the output current is the sum of two additive stationary noise processes that independently follow Gaussian statistics; thus, it is a stationary Gaussian process with a colored PSD.

### B. Derivation of SEP for M-CSK Modulation

Let \( H_m \) be the hypothesis that the symbol \( m \in \mathcal{M} \) is transmitted at the beginning of \( k^{th} \) time slot, and \( Z_k \) be the output current sampled by the receiver for the \( k^{th} \) slot. Then, with the Gaussian approximation of the additive noises, the conditional probability of \( Z_k \) given that the hypothesis \( H_m \) is true can be written as

\[ P(Z_k|H_m) = \frac{1}{\sqrt{2\pi\sigma^2_{I_m}}} e^{-\frac{\left(z_k - \mu_{I_m}\right)^2}{2\sigma^2_{I_m}}}, \quad (45) \]

where \( \mu_{I_m} \) is the mean of the output current given by \cite{47}, and \( \sigma^2_{I_m} \) is the output current variance given by \cite{44}. Assuming that maximum likelihood (ML) detection is applied by the receiver, the decision rule can be expressed by

\[ \hat{m}_k = \arg \max_m P(Z_k|H_m), \quad (46) \]

where \( \hat{m}_k \) is the symbol decided at the receiver for the \( k^{th} \) transmission. The ML decision rule divides the entire range of the output current into \( M \) decision regions corresponding to the \( M \) symbols in the source alphabet. Decision region \( D_m \) for the transmitted symbol \( m \) can be defined as

\[ D_m = \{ Z_k : P(Z_k|H_m) > P(Z_k|H_j), \forall j \neq m \}, \quad (47) \]

for \( m = 0, \ldots, M - 1 \).

Assuming \( N_0 < N_1 < \ldots < N_{M-1} \), from \cite{19} and \cite{37}, we know that the symbols satisfy the following condition

\[ \mu_{I_0} < \mu_{I_1} < \ldots < \mu_{I_{M-1}}. \quad (48) \]

Given this condition, the decision thresholds \( \lambda_k \) separating the decision regions \( D_{m-1} \) and \( D_m \) can be obtained by comparing the conditional probabilities of adjacent symbols \cite{56}

\[ \frac{1}{\sqrt{2\pi\sigma^2_{I_m}}} e^{-\frac{(\lambda - \mu_{I_m})^2}{2\sigma^2_{I_m}}} \quad \text{for} \quad m = 1, \ldots, M - 1. \quad (49) \]

Solving we obtain the decision thresholds as

\[ \lambda_m = \frac{1}{\sqrt{\sigma^2_{I_m} - \sigma^2_{I_{m-1}}}} \left(\sigma^2_{I_m}\mu_{I_{m-1}} - \sigma^2_{I_{m-1}}\mu_{I_m} + \sigma_{I_m}\sigma_{I_{m-1}}\sqrt{(\mu_{I_m} - \mu_{I_{m-1}})^2 + 2(\sigma^2_{I_m} - \sigma^2_{I_{m-1}}) \ln \frac{\sigma_{I_m}}{\sigma_{I_{m-1}}}}\right), \quad (50) \]

for \( m = 1, \ldots, M - 1 \).

The error probability of detection based on the decision thresholds given in \cite{50} can be computed as

\[ P(e|H_m) = \int_{z \notin D_m} P(z|H_m) dz. \quad (51) \]

Assuming that a priori probabilities for all symbols are equal, SEP can be given as follows

\[ P_e = \frac{1}{M} \sum_{m=0}^{M-1} P(e|H_m) \]

\[ = \frac{1}{2M} \left[ \text{erfc}\left(\frac{\lambda_1 - \mu_{I_0}}{\sigma_{I_0}\sqrt{2}}\right) + \text{erfc}\left(\frac{\mu_{I_{M-1}} - \lambda_{M-1}}{\sigma_{I_{M-1}}\sqrt{2}}\right) \right. \]

\[ \left. + \sum_{m=1}^{M-2} \left(\text{erfc}\left(\frac{\mu_{I_m} - \lambda_m}{\sigma_{I_m}\sqrt{2}}\right) + \text{erfc}\left(\frac{\lambda_{m+1} - \mu_{I_m}}{\sigma_{I_m}\sqrt{2}}\right)\right)\right], \quad (52) \]

where \( \text{erfc}(z) = \frac{2}{\sqrt{\pi}} \int_z^\infty e^{-y^2} dy \) is the complementary error function.

### V. Performance Analysis

In this section, we present the numerical results obtained based on the developed model under different settings to reveal the performance of the SiNW FET-based MC receiver. The default values for the controllable parameters used in the analyses are listed in Table \ref{table1}.

We assume that the microfluidic channel is filled up with an electrolyte, which is moderate in its ionic concentration (with \( c_{\text{ion}} = 30 \text{ mol/m}^3 \)) compared to the diluted solutions (with \( c_{\text{ion}} < 1 \text{ mol/m}^3 \) used in \textit{in vitro} biosensing experiments \cite{57} and physiological solutions (with \( c_{\text{ion}} > 70 \text{ mol/m}^3 \) \cite{58, 59}). The employed receptors on the FET surface are considered to be aptamers, the production process of which provides full control over the selection of length, binding and unbinding rates, as well as the type of corresponding ligand molecules \cite{13}. The default length of receptors is set to 2 nm, which corresponds to 6 base pair-aptamers. Binding and unbinding rates, \( k_+ \) and \( k_- \), are set, considering the accepted values in the MC literature \cite{24} and the range of rates that aptamers can provide \cite{60}. Aptamers can bind to a large set of ligands, such as, aptamers, small proteins, RNA and DNA, and even non-organic molecules, which can attain a broad range of elementary charges, as reviewed in \cite{13}. The relative permittivity of SiO$_2$ layer is reported as \( \epsilon_{\text{ox}}/\epsilon_0 = 3.9 \) \cite{19}. The thickness of the SiO$_2$ layer, \( t_{\text{ox}} \), is a design parameter, for which we select a default value of 2 nm. Depending on
the fabrication, the tunneling distance for SiO$_2$ is on the order of 0.01-0.1 nm [61]. We set $\lambda = 0.05$ nm as reported in [19]. We assume the use of a p-type SiNW, which is moderately clean with the impurity density $N_{it} = 10^{16}$ eV$^{-1}$cm$^{-3}$. This corresponds to a low-field mobility of $\mu_p = 500$ cm$^2$/Vs for hole carriers in SiNW (see Fig. 15 in [46]). The Coulomb scattering coefficient is taken as $\alpha_s = 1.9 \times 10^{14}$ Vs/C, which is the value at $T = 300$ K [48]. We assume that a shallow microfluidic channel with cross-sectional height $h_{ch} = 3 \mu$m and length $l_{ch} = 15 \mu$m, resulting in a laminar and steady flow [44]. The fluid flow velocity can be adjusted through the pressure difference between the two ends of the microfluidic channel. We assume an average flow velocity of $u = 10 \mu$m/s, a moderate value widely observed in microfluidic literature [54], [57].

A. Receiver Response and Noise Power

1) Receiver Response: We first investigate the expected response of the receiver to varying number $N_m$ of ligands released by the transmitter. The mean output current generated in the SiNW channel for several transmitter-receiver distances $d = x_R - x_T$ is plotted in Fig. 6. For each distance setting, we observe that the output current increases as the transmitter releases more ligands. However, at some value of $N_m$, the current begins to saturate. This is because as the receptors on the biorecognition layer are occupied by higher number of ligands; the receptors lose their sensitivity to the varying ligand concentration.

From the same figure, we can also infer that for the investigated range of number of released ligands, the receiver is most sensitive to the concentration variations when $d = 1$ mm. Since the attenuation of the concentration is proportional to $\sqrt{d}$ (see Equations (5)-(9)); in the minimum distance case, i.e., when $d = 0.1$ mm, the ligand concentration observed at the receiver location is expected to be much higher compared to the other settings. On the other hand, higher concentration of ligands in the receiver location leads to a more rapid saturation of the biorecognition unit, as evident from Fig. 6. When the distance is increased to 10 mm, the ligand concentration over the receiver significantly decreases for the whole range of $N_m$, so that the variations of $N_m$ do not result in significant differences in the output current. Therefore, the dynamic range of the receiver and the attenuation of the concentration signals in the channel should be carefully considered while designing the overall MC system.

2) Noise Power: To analyze the effect of noise on the receiver operation, we plot the individual PSDs of binding and output noises as well as the overall noise PSD effective on the output current. As seen in Fig. 7, the frequency domain is virtually divided into three regions, in each of which one of the two noise sources is prevailing. At very low frequencies, e.g., $f \ll 0.1$ Hz for the default setting here, $1/f$ noise is dominating over the binding noise, since the binding noise has a flat power density for frequencies

![Fig. 6. Expected output current $\mu_{als}$ as a function of number of ligands $N_m$ released by transmitter and transmitter-receiver distance $d$.](image)

![Fig. 7. PSD of noise effective on the output current of SiNW FET-based MC receiver. The plot reveals the individual contributions of binding and 1/f noise.](image)

| Table I: Default Values of Simulation Parameters |
|-----------------------------------------------|
| Microfluidic channel height ($h_{ch}$)  | 3 $\mu$m |
| Microfluidic channel width ($l_{ch}$)    | 15 $\mu$m |
| Number of transmitted ligands for symbol $m$ ($N_m$) | $10^5$ |
| Max number of ligands TN transmits ($K$)   | $2 \times 10^4$ |
| Transmitter-receiver distance ($d$)        | 1 mm |
| Average flow velocity ($u$)                | 10 $\mu$m/s |
| Intrinsic diffusion coefficient of ligands ($D_0$) | $10^{-10}$ m$^2$/s |
| Binding rate ($k_1$)                       | $2 \times 10^{-14}$ m$^3$/s |
| Unbinding rate ($k_1$)                     | $10^{-3}$ |
| Average number of electrons in a ligand ($N_e$) | 3 |
| SiNW radius ($r_B$)                        | 10 nm |
| Concentration of receptors on the surface ($\rho_{GR}$) | $4 \times 10^{14}$ m$^{-2}$ |
| Length of a surface receptor ($l_{GR}$)    | 2 nm |
| Temperature ($T$)                          | 300K |
| Relative permittivity of oxide layer ($\varepsilon_{ox}$) | 3.9 |
| Relative permittivity of SiNW ($\varepsilon_{SiNW}$) | 11.68 |
| Relative permittivity of medium ($\varepsilon_{M}$) | 78 |
| Ionic strength of electrolyte medium ($\varepsilon_{\mu b}$) | 30 mol/lm$^3$ |
| Source-drain voltage ($V_{GS}$)            | 0.1 V |
| Source-gate voltage ($V_{GS}$)             | 0.4 V |
| Threshold voltage ($V_{TH}$)               | 0 V |
| Hole density in SiNW ($p$)                 | $10^{18}$ cm$^{-3}$ |
| Tunneling distance ($\lambda_i$)           | 0.05 nm |
| Oxide trap density ($N_{ox}$)              | $10^{16}$ eV$^{-1}$ cm$^{-3}$ |
| Effective mobility of hole carriers ($\mu_p$) | 500 cm$^2$/Vs |
| Coulomb scattering coefficient ($\alpha_s$) | $1.9 \times 10^{-14}$ Vs/C |
below a critical value determined by the correlation time, i.e., $f_B = 1/\tau_B$, whereas the power of $1/f$ noise is increasing as proportional to $1/f$. Around $f_B$, the binding noise may become dominant, depending on the total variance of the binding process. At frequencies higher than $f_B$, the power of binding noise is attenuated more rapidly than the $1/f$ noise. Although $1/f$ noise is dominating again at high frequencies, its power decreases under $-220$ dBm, thus, the overall noise power is negligible in this frequency range.

As the bandwidth of the received signal is expected to be at most on the order of Hz [24], contributions of both binding noise and $1/f$ noise at low frequencies should be accounted for while designing the MC receiver.

The overall noise PSD is analyzed also for varying number of ligands released by the transmitter. As can be seen from Fig. 8, the contribution of $1/f$ noise dominating at very low frequencies does not vary as $N_m$ is changed; however, the contribution of the binding noise at low frequencies becomes more prevailing for lower ligand concentrations. This is originating from the fact that the correlation time of the binding fluctuations increases with decreasing ligand concentration, which decreases the critical frequency; see [22].

B. SNR Analysis

In this section, we investigate the effect of main system parameters on the receiver’s output SNR, which is formulated in [43]. We group the system parameters under three main categories: (i) communication system parameters related to the transmitter and communication channel, (ii) molecular parameters related to the characteristics of information carriers and corresponding receptors, (iii) receiver parameters related to the design of the SiNW FET-based MC receiver.

1) Effect of Communication System Parameters: SNR of the output current for varying number of ligands released by the transmitter is plotted in Fig. 9(a) which clearly shows that SNR is significantly improved with increasing number of ligands. However, it begins to saturate at around 45 dB for the default setting due to the saturation of the surface receptors for very high concentrations of ligands. Given a number of ligands released by the transmitter, the output SNR decreases with increasing transmitter-receiver distance $d$, as demonstrated in Fig. 9(b). This is because the ligand concentration is attenuated (proportional to $\sqrt{d}$) as the distance increases.

The effect of ionic strength of the fluidic medium on the receiver SNR is demonstrated in Fig. 9(c). When the ionic concentration increases above 100 mol/m$^3$, the Debye length decreases below 1 nm resulting in substantial screening of ligand charges. Therefore, SNR significantly decreases with increasing ionic strength. Physiological conditions generally imply ionic concentrations higher than 100 mol/m$^3$. To compensate the attenuation of SNR, receptors with lengths comparable to Debye length should be selected.

The velocity of the fluid flow $u$ also has a remarkable effect on the SNR, as is seen in Fig. 9(d). As the velocity increases, the plug arrives more rapidly at the receiver location, resulting in less attenuation. Therefore, higher velocity means higher ligand concentration at the receiver side, and this implies continuously improved SNR until it leads to the saturation of the surface receptors.

2) Effect of Molecular Parameters: Diffusion coefficient $D_0$ is an important characteristic of the information ligands although it also depends on the temperature and the viscosity of the fluid. Its effect on the SNR is shown in Fig. 10(a). As is seen, the SNR decreases with increasing $D_0$. This is mainly caused by the increased dispersion of the concentration plug, which decreases the average ligand concentration that the receiver observes.

The effect of binding constant, $k_1$, is plotted in Fig. 10(b). Increasing $k_1$ means that more ligands can bind to the surface receptors as the plug flows over the receiver, this obviously results in an improved SNR.

We also investigate the effect of receptor length, $l_{SR}$, when the ionic strength is set to 30 mol/m$^3$ which makes the Debye length equal to 1.75 nm. As seen in Fig. 10(c) SNR in dB decreases linearly with the increasing length. Considering also the results plotted in Fig. 9(c), $l_{SR}$ have a substantial effect on the receiver performance.

The number of free charges per ligand also critically affects the receiver response, since the operation the SiNW transducer is mainly based on the field effect generated by the ligand charges. As demonstrated in 10(d), employing highly charged ligands would improve the receiver SNR. As reviewed by [15], oligonucleotides are negatively charged in physiological conditions, i.e., at pH 7.4, due to their highly charged phosphate backbone. For example, a DNA sequence with 4 base-pairs at pH 7.4 can attain a net charge of $-8e$. Likewise, small proteins and antigens, depending on the pH of the environment, can attain a net charge of up to $\pm 4e$.

3) Effect of Receiver Parameters: An important parameter of the receiver design is the size of the SiNW, which has a direct impact on the the ligand flux to the receiver, the number of surface receptors, and the capacitance values of the oxide layer, the SiNW double layer and the diffusion layer. Since we fixed the length of the SiNW as equal to the cross-sectional length of the microfluidic channel, we only change its radius...
and thus, its effective width $w_{R}$ (recall that $w_{R} \approx \pi \times r_{R}$). The effect of SiNW radius is demonstrated in Fig. 11(a).

Increasing the radius also increases the SNR. This is mainly because the transport rate of the ligands $k_T$ and the number of surface receptors $N_R$ increase with the radius.

Given that the SiNW radius is fixed to its default value $r_{R} = 10$ nm, increasing the concentration of receptors on the receiver surface $\rho_{SR}$, which means increasing the number of surface receptors (note that $N_R = w_{R} \times \rho_{SR}$), is another way of improving SNR, as can be inferred from Fig. 11(b). However, size restrictions imposed by the receptor size and possible interactions among densely deployed receptors, such as negative cooperativity, which are not captured by this model, should be accounted for in a real world implementation.

We also analyze the effect of oxide layer thickness $t_{OX}$, which determines the oxide layer capacitance $C_{OX} = \varepsilon_{OX}/t_{OX}$. Fig. 11(c) demonstrates the results for conventional values of $t_{OX}$. As can be inferred, lower $t_{OX}$ implies an improved SNR for the default system settings used in our analysis. This is mainly because increasing $C_{OX}$ results in higher values of transconductance $g_{FET}$, which means more effective transduction of the surface potential to the output current. On the other hand, the effect of $C_{OX}$ on the equivalent capacitance of the transducer is usually negligible compared to the substantial effect of the diffusion layer capacitance $C_{DL}$.

Lastly, we analyze the SNR for varying oxide trap density $N_{ot}$, which is proportional to the impurity of the SiNW. Trap density affects the carrier mobility, and increases the $1/f$ noise, which is very effective in the frequency range of the receiver’s operation. The negative effect of increasing trap density on the SNR is evident from Fig. 11(d).
C. SEP Analysis

In the last analysis, we evaluate the performance of the receiver when M-CSK is utilized for the modulation. We find the SEP for binary, 4-ary, 8-ary, 16-ary cases for different system settings. For better visualization, we present the results for each setting in two different plots separating the binary case from the 4-, 8-, 16-ary cases.

For the constellation design of M-CSK, we assume

\[ N_m = [(m + 1)^s \times (K/M^s)] \]

(53)

where \( K \) is the maximum number of molecules that TN can release in a single transmission, \( s \) is the exponent defined to obtain a non-uniform constellation, and \( m \in M = \{0, 1, ..., M - 1\} \) with \( M \in \{2, 4, 8, 16\} \). Except for the analysis where we investigate the effect of distance, we set \( K = 2 \times 10^4 \) and \( s = 1 \), so that we obtain a uniform constellation where the adjacent symbols are separated by \( K/M \) number of ligands. For the distance analysis, to be able to reveal the correlated effect of the distance and maximum number of ligands, we set \( K = 2 \times 10^5 \).

Figs. [12(a)] and [12(d)] show the SEP as a function of transmitter-receiver distance \( d \). As is seen, for all modulation schemes, the SEP is minimum for intermediate distances, e.g., 1–2 mm, and begins to increase when the distance is below or above this range. The reason can be explained as follows. As the distance gets smaller, the receiver operates near saturation because the concentration of ligands at the receiver location significantly increases when the transmitter and receiver are close to each other. This is reflected to the output current, and results in a decrease in the sensitivity of the receiver so that it cannot discriminate different levels of ligand concentration corresponding to different symbols. In a similar way, when the distance is increased, the ligand concentration is substantially attenuated until the plug reaches to the receiver location, which also leads to a degradation in the receiver sensitivity. Hence, we can conclude that there is an optimal range of distance for a given maximum number of ligands \( K \).

Next, we analyze the effect of maximum number of ligands that the transmitter can release. As can be inferred from the results presented in Figs. [12(b)] and [12(e)], increasing \( K \) up to \( 2 \times 10^5 \) decreases the SEP for binary CSK. However, when we further increase \( K \) above this range, the SEP begins to get higher. Similar trends can be observed for 4-, 8-, and 16-ary cases. The reason is closely related to the reasons that lead to the results obtained with varying distance in the previous analysis. As the transmitter releases higher number of ligands for each symbol, the receiver begins to operate near saturation, which degrades its ability to discriminate different symbols.

The nontrivial results obtained for varying distance and maximum number of transmitted ligands are the consequences of the dynamic range imposed by the receiver, which has a limited reception capacity set by the number of surface receptors. Similar trends also have been noted in the MC experiments conducted with metal oxide semiconductor alcohol sensors [28], [29]. Although these sensors are not operating based on ligand-receptor binding mechanism, they result in saturation when they are exposed to a high concentration of alcohol molecules, since the devices have an active channel limited in size.

Since the response of the receiver is nonlinear as obvious
from Fig. [6] utilizing a uniform constellation for M-CSK modulation obviously is not optimal. Particularly, there is a need to place more symbols on the lower half of the modulation range, for which the receiver is more sensitive to concentration variations, regardless of the maximum number of ligands that TN can transmit. We analyze the performance of the receiver employing a simple non-uniform M-ary modulation scheme by varying the exponent $s$ in (53). For binary CSK case, rising the exponent significantly decreases the SEP, as demonstrated in Fig. [12(c)]. However, when more than two symbols are transmitted as in cases of 4-, 8-, and 16-ary CSK, different trends are observed in Fig. [12(f)]. After some threshold, the SEP begins to increase again. This is because the low level symbols are approaching to each other when we increase $s$, which complicates the detection in the receiver. Hence, there is a need for a more complex constellation design that can properly exploit the nonlinear response of the receiver. The results of these analyses reveal that there is plenty of room for optimization of the MC settings to obtain lower values of SEP.

Second set of analyses is performed for controllable parameters related mostly to the receiver and propagation medium, i.e., ionic concentration of the medium $c_{\text{ion}}$, receptor length $l_{SR}$, and oxide trap density in SiNW $N_{\text{ot}}$. The same trends observed in the SNR analyses are also seen for the SEP. For example, in Figs. [13(a) and 13(d)] we can see that increasing the ionic concentration substantially degrades the receiver performance for all M-CSK modulation schemes. As previously discussed in the SNR analysis, screening of the ligand charges by the medium counterions near the biorecognition layer leads to a lower signal power at the receiver output. In the presence of signal-independent $1/f$ noise, this is reflected to a degraded detection performance of the receiver. The same reason leads to the results obtained for varying receptor lengths, which are presented in Figs. [13(b) and 13(e)]. As can be inferred, it is possible to obtain a SEP lower than $10^{-10}$ for binary CSK case by utilizing receptors with lengths smaller than 2 nm. The trap density of the SiNW channel is also a critical parameter for the receiver performance, as it influences the extent of the $1/f$ noise. Accordingly, lower values of $N_{\text{ot}}$ indicate a clean semiconductor channel, and thus, lead to an improved receiver performance, as evident from Figs. [13(c) and 13(f)].

VI. CONCLUSION

In this paper, as the first step towards implementing a human-made MC system with nanobioelectronic devices, we have developed a communication theoretical model for SiNW FET-based MC receivers integrating all the underlying processes in MC and bioFET operation. Focusing on a microfluidic MC system, we have derived closed-form expressions for fundamental performance metrics, such as SNR and SEP, to provide an analysis and optimization framework for MC with nanobioelectronic receivers. The results of performance evaluation have pointed out several optimization pathways that need to be taken to improve the detection performance of the receiver. The developed model can be extended to incorporate the transient dynamics of MC and bioFETs for enabling analysis also in the frequency domain. Open issues include the design of optimal constellations and optimal receiver detection schemes for MC systems equipped with bioFETs.
receivers. Further research on devising nanobioelectronic MC receivers could enable the implementation of all the theoretical protocols and algorithms designed for reliable and efficient MC and the development of seamless interfaces between MC nanonetworks and macroscale networks towards realizing IoNT.

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