Miniaturized Fully Passive Wireless Neural Recording With Heterogeneous Integration in Thin Packages

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Abstract—Wireless fully passive neural recording systems are demonstrated with heterogeneous integration of sensing, mixing, and communication components in flexible polymer dielectric films. The recording neurosensor has an antenna, antiparallel diode, and a bypass capacitor to modulate an incoming carrier signal with the neuopotentials and backscatter the mixed signal to an external reader circuit. Planar antennas are realized on liquid crystal polymer (LCP) flexible substrates with low dielectric constant. The designed antenna topology is realized with a single metal layer and eliminates the dependency on substrate thickness, and leads to thinner sensors. This is a major advance compared to prior passive neural recording with rigid and thicker substrates and components. Approximately, 80% thickness reduction and 20% volume reduction are achieved with a similar performance when compared to earlier studies. Our approach thus leads to low-cost and disposable wireless neurosensors on a flexible platform with heterogeneous integration. This article thus advances innovative wireless neural recording in three aspects: 1) advanced packaging for miniaturized neural recording with passive telemetry, through which the overall size of the sensor is reduced without degrading the performance; 2) new antenna topologies for neural recording applications; and 3) detailed power link analysis to estimate the minimum signal sensitivity.

Index Terms—Heterogenous integration, liquid crystal polymer (LCP), miniaturization, wireless fully passive neurosensory, wireless neural recording.

I. INTRODUCTION

EVER-INCREASING demand for medical devices has been fueling vital innovations in package integration of flexible bioelectronics for diagnostic and therapeutic systems. Medical device packaging can be classified into multiple categories depending on their use in either implantable or on-skin wearable environments. Rigid components on thick circuit boards in hermetic packages with feedthroughs and connectors are most common for implantable medical systems. Encapsulated rigid packaged devices on flexible carriers, embedding thin bare chips in flexible substrates, fabric-integrated electronics, and transferrable electronics as on-skin e-tattoos, form the key classes for wearable electronics. In all these classes of medical device packaging, there is a continual paradigm shift from rigid packages to thin, flexible packages while still meeting the demand for seamless power and data connectivity, and sensor integration for health monitoring. This shift is to enhance the user’s comfort in terms of skin conformity, minimal obtrusiveness, and invasiveness and also avoid the need for frequent replacement. Additionally, to facilitate easier surgical implantation and less inflammation in the human body, miniaturized flexible packaged devices are preferred for implantable devices. To achieve the system functionality in such thin and flexible packages, it is necessary to integrate power and data telemetry, signal conditioning, and wireless communication functions with thin, flexible, and embedded components. High-density circuitry with advanced component integration in flexible substrates is the gateway to realizing this technology evolution.

Neural recording systems sense and communicate action or field potentials for several applications such as identification of pathological conditions and motor intent, and use the biofeedback to drive therapeutic or functional neurostimulation systems. They are fabricated at different scales depending on the electrode or channel count for the intended function, data rates or bandwidth, and power budget. For systems that record from 32, 64, or 100 channels, the data rates exceed 50 Mb/s and the power requirements exceed 100 mW, leading to energy consumption of 0.1–2 nJ/bit. One common way to design such wireless neural recording systems is to have them head-mount on the top of the skull. In this scenario, the rigid package with communication front-end, analog front-end signal modulation, and the battery is assembled on PCB substrates and encased in hermetic packages with feedthroughs that interface with electrodes. The size of the system can be as high as 100 cc. For example, a system of $38 \times 38 \times 38$ mm$^3$ was designed with a power consumption of 63.2 mW, in which 28 mW is consumed by the RF transmitter [1]. A 1120 mA-h battery was designed and encapsulated in a system of $38 \times 38 \times 38$ mm$^3$. The energy consumption of the circuit is $63.2 \mathrm{mW}$, which in 1120 mA-h battery
of ultrasonic waves, the wavelengths are correspondingly smaller, thereby resulting in mm-scale transducers. A typical footprint of the sensor is $0.8 \times 3 \times 1 \text{ mm}$. However, a major drawback of the system is that ultrasound energy cannot penetrate efficiently through the skull [6]. RF backscattering technique is utilized to realize fully passive and implantable neural recording devices. In such systems, an antiparallel diode, a bypass capacitor, and a matching network are integrated together on a PCB substrate. The whole wireless neurosensor is designed to a size of $15 \times 16 \times 1.5 \text{ mm}$.

In this study, a miniaturized flexible single-layer passive neuropotential sensor ($9.5 \times 8.7 \times 0.5 \text{ mm}$) is demonstrated. Liquid crystal polymer (LCP) substrate is used as the flexible carrier for the circuit. A miniaturized spiral planar antenna is co-integrated with an antiparallel diode mixer and a bypass capacitor on the thin-flex LCP substrate with heterogeneous integration. The entire sensor is covered with biocompatible PDMS, leading to thin and flexible neural recording units for potential use in both on-skin and implantable applications.

II. NEUROSENSOR CIRCUIT SIMULATION AND FABRICATION

Neural recording system comprises two subsystems: 1) an implanted sensor and 2) an interrogator, as shown in Fig. 1. The external device or interrogator transmits a wireless 2.4 GHz carrier to the implanted transponder. After receiving the carrier signal, implanted sensor mixes the 2.4 GHz signal with low-frequency neural signals to produce the second harmonic at 4.8 GHz $\pm f_{\text{neuro}}$. The modulated signal is then sent back to the external device or interrogator. Eventually, 4.8 GHz $\pm f_{\text{neuro}}$ can be observed or viewed at the spectrum analyzer (frequency domain) or demodulated to the original time-domain signal representing $f_{\text{neuro}}$. The implanted sensor (or neural recorder) composes of two components: 1) passive mixer circuit and 2) wireless telemetry. An RF antenna operating at 2.4 and 4.8 GHz acts as the wireless telemetry interface, and the mixer modulates neural signal ($f_{\text{neuro}}$) and backscatters (4.8 GHz $\pm f_{\text{neuro}}$) to the interrogator or an external device via the antenna in the implanted sensor.

One of the challenges in implanted neurosensors is to decrease the antenna size to achieve minimally invasiveness. Several miniaturization techniques have been developed. These include: 1) Stacked multiple radiating patches on top of each other [9], [10]; 2) Planar-inverted F antenna configuration [11], [12], [13]; 3) Capacitive, inductive, or splitting loading to match impedance [14], [15], [16]; 4) Increased current path by meandering or spiraling [17], [18], [19], [20]; and 5) Antennas on substrates with high dielectric constant [21], [22]. Although these techniques document antennas...
Fig. 1. Neural recording system: sensor circuit diagram with APDP, antenna, and bypass capacitor. Interrogating signal 2.4 GHz is sent to the implanted sensor. $f_{\text{neuro}}$ comes from the brain and the mixer circuit yields $(4.8 \text{ GHz } \pm f_{\text{neuro}})$. The up-converted signal is sent back to the external circuitry.

with small footprint and good radiation performance, higher thickness has always been a major drawback. Since our application requires thinness, flexibility, and biocompatibility, these telemetries fall short in meeting the requirements. Recent work reported antennas on flexible substrates with low dielectric constant. However, their footprints are generally larger. To remove the substrate thickness, various coplanar antenna topologies are also investigated. In our work, to meet the flexibility, stability, and thinness requirements along with resistance to moisture, we chose LCP with a coplanar spiral structure to shrink the antenna size. Such a structure was first shown in [23], however, with a single band. Our approach advances this further to achieve a broad dual-band (2.4 and 4.8 GHz) performance.

This antenna-in-package approach also co-integrates the antenna with passive circuitry to realize a miniaturized neural sensor. The neural sensor’s schematic circuit is shown in Fig. 1. The planar antenna was designed on Rogers 3850HT substrates with permittivity of 2.9 and loss tangent of 0.002. A subharmonic mixer is operated with a local oscillator (LO) at half of the typical conventional mixer’s LO frequency. It yields the $2f_{\text{LO}} \pm f_{\text{neuro}}$ signal with $f_{\text{LO}}$, while conventional mixers generate the same output with $2f_{\text{LO}}$. The subharmonic mixer circuit’s components are an antiparallel diode pair (APDP), a matching circuit, a bypass capacitor, and a pair of electrodes. An APDP (Si MA4E2508L-1112) and matching network with transmission line and discrete bypass capacitors (7.5 pF) were integrated with the antenna to accomplish the complete design of the neural sensor.

Estimation of the overall system performance metrics, such as total end-to-end chain loss and backscattered power with the passive circuit requires detailed circuit simulations. The circuit comprises an external antenna and implanted antenna, diode, and capacitors. Cadence® AWR Design Environment was used to simulate the whole neurosensor and predicts the backscattered power. A matching network is required to match the antenna and the APDP impedance. The role of bypass capacitor is to shunt the 2.4 GHz wireless link. APDP performance metrics from vendor datasheets were input into the model. An ac signal source at an amplitude of 1 mV$_{pp}$ and frequency of 1 kHz ($f_{\text{neuro}}$) was used to mimic the neuopotential. For an input power of 15 dBm, the backscattered signal (4.8 GHz $\pm$ 1 kHz) power is approximately $-105$ dBm for the modulated neural signal, which is well above the sensitivity of the signal analyzer $-120$ dBm. The simulated circuit performance with sideband performance is shown in Fig. 2. To detect the neurosignal, the backscattered power has to be 5 dB above the signal analyzer sensitivity.

As described before, Rogers 3850 HT was used as the substrate because of its excellent properties at high frequencies, good dimensional stability, and low moisture absorption rate. The fabrication process is illustrated in Fig. 3. Subtractive copper patterning with dry-film photoresist sensors was performed to acquire the desired line definition and high conductivity for sensor traces in contrast to printed silver traces. Dry-film photoresist (Hitachi Chemicals, RY-5115) was utilized to eventually support large-area fabrication at low cost. The process starts with substrate cleaning and micro etching to enhance adhesion during photoresist patterning and copper etching. UV light was used for lithography at 360 nm narrowband wavelength with an energy dose of 100–120 mJ/cm$^2$. The photoresist pattern for the copper trace

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Fig. 2. Complete circuit model of neurosensor (top). Backscattered sideband power is shown in the frequency domain. The predicted power level is $-105$ dBm (bottom). Path 2 loss is 47 dB.

Fig. 3. (a) Device fabrication process and assembly. (b) Layout and geometry of the neurosensor. (c) Optical photograph of the fabricated neurosensor.

circuit was then developed in a 1% sodium borate solution at 60 $^\circ$C for 65 s, followed by rinsing in DI water. To improve the etching rate, the etchant was heated to 60 $^\circ$C. In the end, photoresist was removed with acetone to expose the copper traces. The substrate was subsequently laser-drilled with vias to interface with the electrodes on the backside. Silver loading with an ink content of 65% was used to fill the vias with a vacuum-assisted via-filling machine. The via electrical connectivity was tested with a multimeter. Compared to silver pastes with 85% particle loading, lower loading of 65% showed better mechanical strength and lower contact resistance. Conductive silver elastomer adhesive as chip-to-flex interconnection layers was utilized for surface assembly of devices to enhance electrical and reliability performance. For biocompatibility, the whole sensor was encapsulated with a PDMS coating.

III. RESULT AND DISCUSSION

A. Electrical Characterization Results

The measurement set-up for evaluating the prototype is shown in Fig. 4. A LO signal was generated by a signal generator. The transmitted power was set at 15 dBm, which is well below the APDP’s absolute maximum (20 dBm), and provided to the external interrogating antenna through a circulator. An arbitrary function generator was used for creating the emulated neurosignal $f_{\text{neuro}} = 1$ kHz with a magnitude of 1 mV$_{\text{pp}}$. The neurorecording device was positioned inside the brain phantom. The phantom consists of skin, bone, dura/gray, and white matter. Recipe of the phantom was taken from prior work [7]. A spectrum analyzer was used to detect the received signal ($4.8$ GHz $\pm f_{\text{neuro}}$) at the reader side, as shown in Fig. 4. Measured spectral response is also shown in the figure. For an input power of 15 dBm, the backscattered sideband RF power is measured as $-109$ dBm while the emulated neural signal is at 1 mV$_{\text{pp}}$ or $-58$ dBm. Overall system loss was 51 dB, which is reasonable to detect neural signal 1 mV$_{\text{pp}}$ at 1 kHz while the implantable and external antennas are 17 mm apart, as illustrated by Fig. 4(b). The discrepancy between the measured value of $-109$ dBm and the simulated value of the sideband power of $-105$ dBm from Fig. 2 is 4 dB. This difference between the simulation and measurement results is mainly because of fabrication and additional unaccounted parasitic values. When implantable neural sensors are placed under the skin and the interrogator is 2 mm away, the system loss is 31 dB, deduced as the difference between the input neural power of $-58$ dBm and the received sideband power of $-89$ dBm, as illustrated in Fig. 5, which is similar to the values reported in [8]. Fig. 6 (Bottom) indicates that the Path 2 loss for 5 dBm of incoming RF power is $\sim 27$ dB when the sensor and interrogator are 2 mm apart and the sensor is under the skin, again showing 4 dB discrepancy from simulations. This analysis is further discussed below.

B. System Performance Analysis and Projection

The key performance metrics for a neural recording unit are: 1) low interrogating power; 2) high backscattering power because of low conversion, propagation, and other system losses; and 3) high sensitivity to weak (20 $\mu$V) neurosignals. These three aspects are examined in detail here with a simple diagram Fig. 6 (Top).

The Path 1 loss is guided by two factors, viz., propagation loss at 2.4 GHz and matching loss. The incoming power should compensate for the propagation losses from the interrogator to the device through the air and tissue and losses from the mismatch

$$\text{Path 1 loss} = L_{\text{Propag@2.4 GHz}}[\text{dB}] + L_{\text{Match}}[\text{dB}]. \quad (1)$$
We will focus on Path 2 loss, which is also interpreted as system loss in this work because it results in lower recorded neuropotential signal power. The system loss in Path 2 is related to the conversion loss, matching, and propagation losses, as shown in Fig. 6 (top) and described in (2). System
loss can be represented as

\[
\text{System loss in Path 2 [dB]} = L_{\text{Conversion loss}}(8 \text{GHz} \pm f_{\text{neuro}}) + L_{\text{Propag}}(8 \text{GHz} \pm f_{\text{neuro}}) + L_{\text{Match}} \text{ [dB].} \tag{2}
\]

\(L_{\text{Conversion loss}}\) is the loss during the conversion from input carrier power to the backscattered carrier signal power, \(L_{\text{Propag}}(8 \text{GHz} \pm f_{\text{neuro}})\) is the propagation loss at 4.8 GHz \pm \(f_{\text{neuro}}\), and \(L_{\text{Match}}\) is the mismatch loss between implanted antenna and APDP. The conversion loss is lowest (9 dB) when the input power at the mixer without any path loss is \(-2\) dBm. As the power deviates from the optimal value, the losses increase. The plot of the conversion loss as a function of the incoming power was simulated and estimated, as shown in Fig. 7, for the completion of the discussion.

Tissue behaves as a conductor at high frequencies as the human body becomes a medium with high-dielectric loss. As the backscattered RF signal propagates from a high-dielectric-constant medium to a low-dielectric-constant medium, it bends away from the direction perpendicular to the interface. In other words, the signal effectively travels longer distances in an inhomogeneous tissue and also encounters multiple reflections. With higher propagation loss, the transmitter needs to transmit at higher power to keep the conversion loss under control so that a higher signal sensitivity is achieved. Therefore, placing the implantable neural sensor just under the skin [Fig. 4(b)] would decrease the propagation loss and lead to a lower power carried by the incoming carrier at 2.4 GHz. One other way to avoid this high propagation loss is to introduce multiferroic antennas [24] that operate at lower frequencies.

The key system metric is the sideband neural signal power that can be detected by the receiver circuit. The lowest neural signal that can be detected is dependent on minimum receiver sensitivity and system loss. With low system loss, we can record weaker neuropotential signals. Subtracting the lowest detectable neural signal power from minimum receiver sensitivity yields tolerable system loss, as described by the following:

\[
\text{Lowest Neuropotential [dBm]} = \text{Minimum of the receiver sensitivity} + \text{System loss in Path 2.} \tag{3}
\]

From the analysis, the backscattered power should be at least \(-125\) dBm so that the carrier power at the receiver is adequate and above \(-130\) dBm. For this receiver sensitivity of \(-130\) dBm, the lowest neuropotential signal is \(-90\) dBm (20 \(\mu\)V), when the target system loss is 35 dB. In other words, as long as the sum of conversion modulation loss, matching loss, and propagation loss at 4.8 GHz is less than 35 dB, we can detect neuropotential signals as low as 20 \(\mu\)V. Reducing the distance between the implant and sensor leads to a decrease in propagation loss. When fixing the distance between external antenna and implantable sensor to 2 mm, we predict that 20 \(\mu\)V can be acquired with the optimal incoming power at 2.4 GHz power because the measured Path 2 loss for 2 mm is only 31 dB. As seen from the equation, it is critical to minimize both the propagation and mixing losses to achieve superior signal sensitivity.

Additional mixing losses arise from the APDP and the impedance mismatch between the incoming neural signal and the APDP. For example, the electrodes are at high impedance, which invariably leads to substantial losses in the signal that the APDP receives at its input. Impedance-matched recording is hence preferred to lower these losses. One way to achieve this is by utilizing electrodes with higher surface area and closer to the neural signal source. Impedance transformation with a bipolar junction transistor (BJT) device is also an alternative approach to address this challenge [6]. We will briefly discuss this novel circuit topology shown in Fig. 8.

In this approach, the diode not only acts as a dc bias for BJT but also as a mixer, while the BJT converts the high impedance coming from the electrode-tissue interface into low impedance, resulting in recording low-potential brain signals [25]. However, small form factor with bare die PNP BJT and diode with embedded packaging is yet to be realized.

Specific absorption rate (SAR) is an estimate of the RF energy absorbed by the human body or tissue while a wireless transmission occurs. The SAR generated by our neurosensing system was assessed by ANSYS1HFSS. The implant was 1 mm below the skin layer, and the external antenna was 1 mm above the skin layer, as shown in Fig. 5(a). An RF input power of 5 dBm is assumed at 2.4 GHz. The calculated SAR1g is 0.97 W/kg while SAR10g is 0.13 W/kg, which are in compliance with Federal Communications Commission (FCC) for uncontrolled environment exposure (SAR1g < 1.6 W/kg).
TABLE I

| Size | Packaging |
|------|-----------|
| [1]  | 51 × 38 × 38 mm³ | Rigid substrate, thick prepacked dies, and rigid soldering for interconnections |
| [2]  | 56 × 42 × 9 mm³ | |
| [7,8]| 10 × 9 × 2.2 mm³ | |
| **Our work** | 9.5 × 8.7 × 0.5 mm³ | Flexible substrate, thin chip, and silver elastomer for interconnections |

[26] and International Commission on Non-Ionizing Radiation Protection (ICNIRP) (SAR_{1g} < 2 W/kg) [27], respectively. Therefore, our neurosensing system is projected to meet the patient safety guidelines.

C. Package Size Analysis

The package size is determined by the antenna footprint, mixer, matching components and interconnects. This size is usually much smaller than the electrode extensions between the neurosensor and the neural interfaces. Since the electrode design is customized to the target recording application and is not a key focus of this work, and we only compare the neurosensor size without including the electrode size. The size or form factor of previous rigid packages on circuit boards is compared with current packages in this work in Table I. This work implements planar antennas on flexible substrates. The proposed monopole spiral antenna is agnostic to the substrate thickness, making it much thinner than today’s neurosensor devices. As the telemetry is substrate thickness-independent, the neural recording system can be implemented on much thinner substrates (for example, 50 µm thick flexible substrate). Comparing to other published work, our work thus demonstrates thinner, smaller, and flexible packages due to planar antenna design, single-layer circuitry, and thin chips.

The next key factor that determines the module thickness is the component thickness and off-chip interconnection height. In traditional packaging, electronic components are assembled onto the substrate with thick solder joints at temperatures above 230 °C and encapsulated, which eventually lead to thicker packages. Moreover, prepackaged thicker chips in leadframe Quad Flat No-Lead (QFN) packages lead to bulky packages. To realize thinner substrates, it is necessary to embed electronic components into the flexible substrate. This is consistent with the emerging trend of sub-100 µm dies for embedding and fan-out packaging technologies. In fan-out packaging with chip-embedding in flexible carriers, substrates with cavities and a back-up layer are fabricated as the first step. Devices are then assembled into the cavities, and off-chip interconnects are directly formed from the chip pads to the package traces with printed conductive elastomers.

Additionally, a thinner die is connected with copper traces and silver-elastomer interconnects, which does not add any substantial height to the package. More importantly, advanced packaging with embedding dies and thin-film passive components in flex will further reduce the size of neurosensors. Such trends have been recently realized to further reduce the package form factor [28]. Passive electrode impedance matching in neural recording sensors, when realized with embedded packaging, will further reduce the system losses and increase the signal sensitivity.

IV. SUMMARY

Heterogeneous integration of thin active devices, power and data telemetry, and storage components in thin, flexible substrates will pave the way for miniaturized bioelectronics. Toward this goal, we have shown a passive telemetry communication system for transmitting neural signals. A simplified passive backscattering approach miniaturizes the topology and bill-of-materials for system components. Innovative topologies with monopole spiral miniaturized antennas further reduced the system size. Complete system loss analysis showed cumulative propagation losses of approximately 30 dB for the backscattered signals. The projected signal sensitivity for this system is 20 µV, assuming a receiver sensitivity of −130 dBm. With the decreasing distance between the interrogator antenna and the implanted sensor, the required input power is reduced to 5 dBm, which is adequate to acquire neuropotential as low as 20 µV. The lowest input power for the system function can be further reduced with advances in the diode or mixers. The article thus reports advances in package miniaturization, innovative telemetry topologies, and detailed power link analysis to predict the neural signal sensitivity.

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