A 21.3%-Efficiency Clipped-Sinusoid UWB Impulse Radio Transmitter With Simultaneous Inductive Powering and Data Receiving

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Abstract—An ultra-wide-band impulse-radio (UWB-IR) transmitter (TX) for low-energy biomedical microsystems is presented. High power efficiency is achieved by modulating an LC tank that always resonates in the steady state during transmission. A new clipped-sinusoid scheme is proposed for on-off keying (OOK)-modulation, which is implemented by a voltage clipper circuit with on-chip biasing generation. The TX is designed to provide a high data-rate wireless link within the 3-5 GHz band. The chip was fabricated in 130 nm CMOS technology and fully characterized. State-of-the-art power efficiency of 21.3% was achieved at a data-rate of 230 Mbps and energy consumption of 21pJ/b. A bit-error-rate (BER) of less than $10^{-6}$ was measured at a distance of 1 m without pulse averaging. In addition, simultaneous wireless powering and VCO-based data transmission are supported. A potential extension to a VCO-free all-wireless mode to further reduce the power consumption is also discussed.

Index Terms—Ultra-wideband (UWB) transmitter, impulse radio, high efficiency, low-power wireless, inductive powering, simultaneous power and data transfer.

I. INTRODUCTION

Recent advances in biomedical devices have created demands for short-range high data-rate wireless transmission. As biomedical devices are integrating versatile sensors with higher spatial-temporal resolutions, the volume of data that needs to be streamed outside the device increases accordingly. Examples of such high data-rate biomedical devices include electronic neural interfaces [1], [2], the targeted application of this work, as well as retinal prosthetic implants [3], wireless biomedical sensors [4], and optical imaging devices [5]. Fig. 1(a) illustrates an example of an electronic neural interface for chronic neuroscience studies in rodents, where a cellular inductive powering floor delivers energy to a freely-moving rodent for remote electrophysiological brain monitoring [6]. A simplified block diagram of such a wireless system is depicted in Fig. 1(b).

A major challenge in designing high data-rate biomedical devices in general, and brain implants in particular, is the limited power budget. The power source of such devices is either a lightweight low-capacity battery or a wireless power link by means of inductive coupling or energy harvesting [7]. The heat density requirement of miniaturized implantable devices adds additional limitations. As a result, the available power of these devices is often limited to a few milliwatts per centimeter square, which is not sufficient for most conventional transmitters (TXs) to support the required data-rate [8].

Ultra-wideband impulse radio (UWB-IR) is one of the most suitable architectures for short-range (<10 m) high data-rate (>10 Mb/s) transmission [9], [10], [11], [12]. A UWB-IR TX directly radiates a train of short pulses (<1 ns). The direct
transmission of short pulses results in a high data-rate as the symbol period can be as small as the duration of an individual pulse.

Compared with the state-of-the-art low-power narrow-band transmitters (e.g., Bluetooth or Zigbee [13], [14]), UWB-IR TX designs typically offer ×10 bandwidth at a per-bit energy dissipation that is orders of magnitude smaller [15]. However, the output power efficiency of existing UWB-IR TX architectures is often low [16]. This is mainly due to the poor power efficiency of the output stage that drives the antenna [10], [11]. As a result, UWB-IR TXs are often designed with limited operating distance, which makes them unsuitable for many practical applications.

Several innovative low-power UWB-IR architectures have been proposed [18], [19], [20], [21], [22], [23], [24], [25], [26], [27]. The designs in [18], [19] use a combination of CMOS inverters as delay lines in order to generate the UWB waveform and drive the antenna. Theoretically, the TX only radiates power during the logic-state transitions and thus the power is limited by the rising and falling times of the inverters. In practice, due to the additional power consumption overhead from the digital delay lines and pulse-shaping circuits, the resulting overall TX power efficiency is often poor [9].

On the other hand, the designs in [20], [21] generate UWB pulses by turning on and off a digitally-controlled cross-coupled LC oscillator, as illustrated in Fig. 2(a). In these cases, the TX efficiency is limited by the startup time of the oscillator. Because of the lower voltage swing, the oscillator’s power efficiency during startup is much lower than its steady state. In addition, the oscillation frequency in these designs often needs to be several times larger than the pulse bandwidth, which makes the oscillator further more power-hungry.

If the UWB pulses can be generated from an LC oscillator operated in the steady state, the LC tank can always resonate with high power efficiency [28]. Moreover, by eliminating the need for periodically turning on and off the LC tank, the bitrate is no longer limited by the startup time of the LC tank, thus a more power-efficient LC tank with a lower resonant frequency can be employed.

This paper introduces a new UWB-IR TX architecture that achieves a state-of-the-art TX power efficiency. The proposed design has three key advantageous features:

- First, a novel clipped-sinusoid pulse generation scheme is proposed, which integrates a voltage clipper circuit at the output of an LC tank that is always in the steady-state, thus avoiding the energy-costly on-off transitions.
- Secondly, the UWB-IR transmitter can be inductively powered (and can receive control commands) while simultaneously transmitting data.
- Thirdly, by coupling the TX inductor with the power receiving coil, the frequency synthesizer circuit can be eliminated to enable a VCO-free all-wireless mode. This mode can further reduce the power consumption and is especially suitable for ultra-low-power biomedical sensors.

A part of this design and preliminary experimental results have been briefly reported in [28]. In this paper, we expand on that report and present the following additional aspects of the work that have not been previously covered:

- A detailed analysis of the design methodology for improving the TX efficiency and of its circuit implementation and operation.
- Introduction and validation of the simultaneous inductive powering and data transmission scheme, which is an essential newly reported feature of the system.
- Discussion of the newly reported all-wireless VCO-free configuration with both power and clock received wirelessly, which can further simplify the circuit implementation and reduce the power consumption.
- A detailed description of the experimental setup and additional measurement results and a comparison with state-of-the-art designs.

The rest of the paper is organized as follows. Section II analyzes the power efficiency in UWB-IR TX designs. Section III presents the new UWB-IR TX system and circuit implementation, as well as an all-wireless configuration supporting simultaneous powering and data transmission. Section IV presents the detailed measurement results of the prototype system in various test modes. Section V discusses an extension of this work to support a VCO-free all-wireless mode to further reduce the power consumption. Section VI compares the performance of the presented work with state-of-the-art UWB TX designs. Finally, Section VII concludes the paper.

II. IMPROVING TX POWER EFFICIENCY

Fig. 2(a) shows the operational principles of a conventional UWB-IR TX based on switching on and off a cross-coupled LC tank VCO. A UWB pulse is generated when the tank is turned on by the baseband signal V2. Considering only the power loss in the LC tank and neglecting the loss in the active components of the VCO, the power efficiency can be approximated by:

\[
\eta_{TX} = \frac{R_A I(t)^2}{V_1 I(t)}
= \frac{R_A \left[u(t_p - t)(e^{\alpha t} - 1) - u(t - t_p)e^{-\alpha(t-t_p)}\right]}{R_A + R_s}
\times 100%,
\]
where $\alpha$ is the time constant equal to $-L/[2(R_S + R_A)]$, and $t_p$ is the pulse width (during which $V_2 = 1$ in Fig. 2(a)). Due to the highly underdamped response of the LC tank in this case ($\alpha \ll \omega_0$), the numerator is a small fraction of $R_A$, therefore $\eta_{TX}$ is limited by the transient behavior of the LC tank.

It should be noted that the above expression of power efficiency of the LC tank VCO-based UWB-IR TX is based on the assumption that the duration of signal $V_2$ is much smaller than the time constant of the tank, i.e. $e^{-\alpha t_p} \ll 1$. This condition is critical because as the oscillation builds up and the swing of the signal $V_2$ increases, the cross-coupled transistors in the LC tank VCO become increasingly non-linear, while these non-linearities are not reflected in the expression above.

In the actual implementation of the UWB-IR designs as described in [20], [21], the condition $e^{-\alpha t_p} \ll 1$ is valid since the pulse duration $t_p$ lasts only for a few oscillation cycles. This is while the number of oscillation cycles that would make the condition valid must be comparable to the ratio of the tank’s resonant frequency $\omega_0$ to its nuer frequency $\alpha$. Since this ratio is essentially equal to $2Q$, which is an order of magnitude larger than the number of oscillation cycles, the above approximation is considered valid.

![Fig. 3. Block diagram of the presented UWB-IR TX system.](image)

In the operational principle of the proposed UWB-IR TX, Two pulses are generated in every oscillation period of the LC tank. These high-bandwidth pulses are generated by voltage clipping at the output of the LC tank by the two diodes connected in series between $V_{MAX}/2$ and $-V_{MAX}/2$. The clipped signal, $V_2$, contains higher-order harmonics due to the abrupt limiting action of the diodes. The spectral power of the higher-order harmonics depends on the threshold voltages $V_{MAX}/2$ and $-V_{MAX}/2$, which can be digitally set by digital-to-analog converters (DACs). As shown in Fig. 2(b), the raw UWB pulse train, $V_3$, is created at the antenna by high-passing the clipped signal $V_2$. It should be noted that Fig. 2(b) is a conceptual representation of the key idea, which excludes the OOK coding scheme. Each pulse in $V_3$ may actually represent multiple pulses in a ripple, depending on the quality factor of the LC tank and the clipping voltages.

In the proposed UWB-IR TX architecture, since all the pulse power is sourced from the LC tank that is resonating in the steady state, the overall efficiency of the system is given by:

$$\eta_{TX} = \frac{R_A}{R_A + R_S} \times 100\%.$$  

Therefore, the power efficiency of the TX remains high at all times, as long as the LC tank is implemented with a high-Q inductor, and the diode junctions are abrupt enough to extend the pulse bandwidth over the frequency band of interest.

III. UWB-IR TX DESIGN

A. TX System Architecture

Fig. 3 shows a detailed block diagram of the proposed UWB-IR TX system. It consists of a 915 MHz frequency synthesizer, a pre-amplifier, a power amplifier (PA), a resonant LC tank, a two-diode clipper circuit, and a delay-locked loop (DLL) for the OOK pulse modulation. A 915 MHz pure tone is generated by the on-chip frequency synthesizer from an off-chip 14.3 MHz crystal. The frequency synthesizer uses a cross-coupled LC tank voltage-controlled oscillator (VCO) and a true single-phase clock (TSPC) flip-flop phase-frequency detector (PFD) [18]. The synthesizer is connected to a pre-amplifier which drives the inductive-load PA, which in turn drives a high-Q series LC tank. The LC tank is implemented off-chip. The threshold voltage $V_{MAX}$ and the AC ground $V_{MID}$ (ideally equal to $V_{MAX}/2$) are set by 8-bit DAC1 and DAC2, respectively. These voltages can be digitally adjusted for optimizing the radiated power. The DACs can also be used for compensating PVT variations.

The UWB pulse train is modulated by switching the output of the LC tank based on the data stream. When shorting $V_2$ to the AC ground $V_{MID}$, both diodes remain off and no pulse is generated. Since the swing of $V_2$ remains many times larger than the swing of $V_1$ regardless of the switch state, the switching action does not impact the quality factor of the LC tank. The DLL is situated between the data stream and the switching node to ensure proper timing between the switching signal and the transitions of the diodes.

B. TX Circuit Implementation

The proposed design is implemented in a standard 130 nm CMOS technology. Fig. 4 shows the simplified schematic of the pulse generation circuit including the pre-amplifier, PA, LC tank, diode clipppers, and the OOK switch. The pre-amplifier is a differential pair with a diode-connected current-mirror load. It is used to convert the differential VCO outputs to a single-ended output, and provide an additional gain for driving the PA. The choice of the PA architecture is important and requires careful consideration. While most of the advantage in the proposed design comes from the fact that the LC tank is always on and does not require energy-consuming on-off transitions, as was described in Fig. 2, the PA is another aspect of the design that can have a significant effect on energy efficiency. In this work, we adopted a Class-C PA, as depicted in Fig. 4 (top, middle). Although Class-C PAs are not as power efficient as their Class-D counterparts, the amplifier efficiency in this work is maintained high as the PA is on only for a small fraction of the input signal.
period. Class-C amplifiers are inherently nonlinear. Distortion is reduced by the tuned LC components at its output, eliminating the need for active filters at the cost of area. The filter is then coupled to an off-chip LC tank connected to a chip antenna. Impedance variation in the external components such as in the bondwires and antenna have only a minor effect on the PA efficiency, as bondwire geometry is design-controlled and the antenna is packaged.

The clipping diodes are implemented by two diode-connected triple-well NMOS devices $M4$ and $M5$, which have a width of 50 $\mu$m and the minimum length. The OOK switch is implemented by an NMOS device $M3$, which has a width of 200 $\mu$m and the minimum length. When transmitting “1,” the OOK signal is low and the clipped signal $V2$ is sent to the antenna; when transmitting “0,” the OOK signal is high and $V2$ is shorted to $V_{MID}$. When shorting $V2$ to $V_{MID}$, both diodes remain off, therefore no pulse is generated. $C1$ and $C2$ are used as decoupling capacitors at the outputs of DAC1 and DAC2, respectively. Each of them has a total capacitance of 200 pF. DAC1 and DAC2 can be individually programmed. Despite the fact that $V_{MID}$ is set to be half $V_{MAX}$ by default, using two DACs allows for more flexibility in choosing the parameters and potentially compensating for mismatches (e.g., the threshold variation of the two diodes). The matching of the two DACs is not a major concern since the output voltages can be calibrated and the 8-bit resolution is sufficient for the purpose of this design.

Fig. 5(a) shows the schematic of the DLL circuit. The DLL is comprised of a phase detector (PD), a low-pass filter (LPF), and a variable delay line. The loop regulates the delay of the inverter-chain-based delay line until its output precedes the PA’s output ($V1$) by exactly $T/4$, where $T$ is the oscillation period of the LC tank. The DLL quantifies the misalignment between the two rising edges by comparing the duration of every OOK “1” bit with the duration of the concurring “1” bit at $V1$. To implement this, $V1$ is first AC-coupled and digitally buffered to generate a rectangle-pulse signal $V1'$. The PA’s output is designed to not exceed the supply voltage at all output power levels, and a diode protection circuit was added to its output pad to avoid damaging the input gate of the digital buffer. $V1'$ is then inverted and NORed with the output of the delay line. Two pulse-averaging RC filters quantify the pulse widths of the data “1” bits as seen at the output of the delay line and the NORed output. The RC filter for the delay line output has a DC gain of 1/2, such that these two filters’ outputs are at equal levels when the PA and the data bits are exactly $T/4$ apart in phase. The difference between the outputs of the two RC filters is quantified by a differential amplifier which is implemented as a self-biased differential pair. A compensation capacitor $C1$ is added to the differential amplifier’s output, $V_{ctrl}$, to stabilize the feedback loop.

It should be noted that an auxiliary quarter-period “1” bit is inserted after every data bit, as shown in Fig. 5(b). This is to ensure that the loop settles only when the bits precede the PA zero crossings by a $T/4$. Without the auxiliary “1” bit, the DLL may settle falsely when the bits follow the PA’s output by a $T/4$.

C. VCO-Based All-Wireless Mode

The proposed TX system features a VCO-based all-wireless mode that supports simultaneous powering and data transmission. Fig. 6 shows the block diagram of the inductive power receiver designed in this work to power the UWB-IR TX. The level of the received power is regulated by adjusting the tuning of the LC tank, such that the tank becomes detuned slowly when there is excess in the received power. The detuning may result in degradation in the overall power transfer efficiency, but the highest priority for power receiver design in applications such as implantable medical devices is to ensure sufficient power receiving for normal operation, while making sure the heat dissipation will not damage the tissue environment. The detuning helps reduce the excess heat from the LC tank. The efficiency of the power transfer link is a secondary concern in this case, since the power is generated by external devices with less energy.
IV. EXPERIMENTAL RESULTS

A. IC, TX PCB, & RX PCB Prototypes

The design was fabricated in a 130 nm standard CMOS technology, occupying a silicon area of 1.7 mm \times 0.7 mm. The micrograph of the fabricated chip is shown in Fig. 7.

To validate the wireless operation of the designed UWB-IR chip in experimental neuroscience applications, a mini-board was developed as shown in Fig. 8(a). The mini-board has a dimension of 2 cm by 2 cm, which integrates the fabricated chip, a chip antenna, a low-power FPGA, and power management units. The size of the board permits many wearable and implantable applications [2], while maintaining flexibility and generality as needed to accommodate multiple applications. The generated UWB pulses were radiated from the chip antenna assembled in the upper-left corner of the board. The chip antenna return loss is plotted in Fig. 8(b), which verifies that the antenna radiates best within the 3.3 GHz-8 GHz band. In addition, an on-board power management block was integrated to rectify, down-convert, and regulate the high-power signal from the inductive coil connected to the board. For extreme volume-constrained applications, the device form factor can be further reduced by integrating the power management circuits and the FPGA digital logic circuits on the chip. This would support an even wider range of applications, such as implantation in the deeper brain [29], but at the expense of reduced general-purpose utility.

The transmitted UWB-IR pulses are picked up by an RX antenna, which is shown in Fig. 9(a). The RX antenna was used only with the external receiver and was not designed to be integrated into the miniature module. Since there is no form factor constraint for the external RX antenna, it was designed to radiate best within 2.4-8 GHz and its size was set by the lower limit of the radiation frequency (i.e., 2.4 GHz).

B. Wired-Power Wired-Data Test Mode

We first characterized our design in a wired-power wired-data test mode. This test mode allows us to fully evaluate the functionalities and performance of the modulation and demodulation scheme and characterize the TX output power for ensuring its compliance with the standards approved by the Federal Communications Commission (FCC). In this test mode, an external wired power supply was used to power the on-board regulators. The output of the TX was first measured directly using an Agilent DSO-X 92004 A oscilloscope at a sampling rate of 80 GSa/s. The experimental setup is illustrated in Fig. 10(a). A low-power Actel FPGA was used for chip configuration and data handling. The baseband data used for testing the wireless link was fed constraints. The RF power of the LC tank is transformed to DC using a dual-halfwave rectifier, which offers better power efficiency than a full-wave bridge rectifier. A limiter is used at the output of the rectifier to avoid a sudden surge in the rectified voltage. Slow-varying feedback is fed from the limiter to the regulator for keeping the output voltage level steady over a long time without dissipating excess power and generating unnecessary heat.
by a pseudorandom binary sequence (PRBS) generator. The PRBS sequence was simultaneously fed to another channel of the oscilloscope for post-processing and bit error rate (BER) computation.

The modulated output of the chip was measured at the maximum output power allowed by the FCC mask. Fig. 10(b) shows the transient output of the TX modulated by a PRBS. Fig. 10(c) shows the spectrum of the OOK modulated pulse train, which spreads over the 3GHz-5 GHz frequency range. A spur at 915 MHz is visible in the measured spectrum. We hypothesize that the spur is due to the coupling from PLL from the layout. A more careful layout review should be conducted in the future to avoid potential coupling effect. Nevertheless, the measured output spectrum of the design was under the UWB spectral mask approved by FCC and the spur didn’t affect the transmission performance. At higher UWB frequencies, antennas either have a small aperture or are extremely sensitive to misalignment. Therefore extending radiated spectral power beyond this frequency range is of less interest, especially for wearable and implantable biomedical devices where misalignment of antennas cannot be totally avoided.

C. Wired-Power Wireless-Data Test Mode

Next, we tested the design in the wired-power wireless-data mode. This test mode allows us to characterize the performance of wireless transmission over the air without uncertainties in the power supply (since the device is powered through a wire). The data transmission was tested at different distances between the TX and RX modules. The experimental setup of this test mode is illustrated in Fig. 11(a). The UWB receiver was the oscilloscope with an RX antenna. Similar to the previous setup (as shown in Fig. 10(a)), the board was powered by an external supply and the biasing levels were generated by the on-chip DACs. The PRBS test data was fed to the TX board by the signal generator and was sent to the oscilloscope simultaneously for post-processing and BER computation. In the experiments with long distances (e.g., 1 m and above), a synchronization signal was routed between the external power transmitter and the oscilloscope for triggering the segmented signal storage function of the oscilloscope. This is because the received pulses were too small to be detected by the notch trigger of the oscilloscope. Figs. 11(b) and (c) show the experimentally measured UWB signal and the corresponding power spectrum, both measured at the RX antenna at a distance of 0.5 m.

The receiver recovered the data by interpreting the measurements based on the scheme illustrated in Fig. 12. The oscilloscope was set to be triggered by any notch in the received signal that was narrower than 500 ps. Once a notch was detected, a 10 ns segment of the recorded RX signal containing the notch was stored in the memory. In an actual receiver circuit, the notch
trigger can be replaced by an ultra-fast logic gate circuit that implements the 2-step edge detection scheme.

The stored RX segments were processed offline to determine the BER. A correlated double sampling scheme was performed. By taking three consecutive samples from the RX signal in 100 ps intervals, the algorithm detected whether a transmitted UWB pulse existed within each stored segment of the scope. The “EDGE1” and “EDGE2” signals were the outputs of two slope detection blocks which evaluated the rise in amplitude from \( t_0 \) to \( t_1 \), and from \( t_1 \) to \( t_2 \), respectively. A UWB pulse was flagged to be present within the segment when the output of both slope detectors “EDGE1” and “EDGE2” were high. A bit “1” was assigned to each recorded RX segment when the algorithm detected a UWB pulse during that segment. Each segment also had a time stamp recorded using a separate channel. A bit “0” was assumed where no pulse was detected by the algorithm. By comparing the bit “1” segments with the original transmitted PRBS sequence, the BER was calculated.

D. VCO-Based All-Wireless Test Mode

Lastly, we characterized the system in the VCO-based all-wireless mode. This test mode allows us to test the wireless data transmission while receiving power simultaneously. We benchmarked the final BER and power efficiency in this mode. Fig. 13(a) illustrates the experimental setup for this all-wireless mode. In this mode, the power management circuits receive power directly from the power receiver (as shown in Fig. 6) and power the whole TX circuit. The buck converter regulates its input (between 10V-40 V) to a constant output of 3.6 V, which powers two on-board regulators, one for the chip and the FPGA IO banks, the other for the FPGA core. A high-frequency choke isolates the chip’s supply from the FPGA’s supply to minimize the noise coupling. Figs. 13(b) and (c) show the measurement results of the chip in the all-wireless mode at a distance of 0.5 m.

The TX used a 2 cm × 2 cm planar rectangular coil with the same footprint as the receiver. The coil was developed on a 2-layer flexible PCB substrate with turns on the top and bottom layers in series to increase the quality factor. The inductive coil (positive) terminal was routed to the on-board FPGA, which generated the reference clock from the coil. To avoid shunting the coil current through the ESD protection diodes of the FPGA IO banks, a 30KΩ resistor was placed between the coil and the FPGA (not shown in Fig. 13(a) for simplicity).

The power efficiency was measured in the VCO-based all-wireless test mode. When the output power is programmed to -1 dBm and the chip is transmitting at 230 Mbps, the TX power efficiency was measured to be 21.35%. Fig. 14 shows the measured BER in the all-wireless test mode at a distance of 1 m. During measurement, we swept the TX output power to collect the data points for different Rx input power levels. The RX input power was measured using the spectrum analysis function of the oscilloscope.

We first tested device with a high data rate of 230 Mbps. At a distance of 1 m and a TX output power of -1 dBm, the RX input power is about -59 dBm, which includes the pass loss over
the air and the antenna gains. The integrated noise of the RX over a bandwidth of 1.5 GHz is about -70 dBm, measured using a resolution bandwidth (RBW) of 100 kHz. This yields a SNR of about 11 dB, which is needed to achieve a $10^{-6}$ BER for OOK modulation [30]. To improve the BER performance, we also tested a pulse averaging scheme. The averaging scheme uses multiple transmitted pulses to represent one bit, and the averaged power of the received consecutive pulses is used to determine the bit value [9]. We applied a 5-pulse averaging scheme to the 230 Mbps transmission, resulting in a reduced data rate of 46 Mbps. The experimental results in Fig. 14 show that the applied 5-pulse averaging scheme achieved a 3 dB improvement for the BER. However, it should be noted that the energy per symbol for using 5 pulses averaging increases by 5x ($\sim$14 dB). In conclusion, the pulse averaging scheme is not as efficient as directly increasing the power per pulse, which can be achieved by adjusting the clipping threshold voltages (i.e., $V_{\text{MID}}$ and $V_{\text{MAX}}$) in this work. However, increasing TX power is essentially limited by the total power budget of the device. For energy-constrained applications such as small medical devices, increasing TX power may not be feasible. In these cases, pulse averaging may be used as an alternative to improve the BER at the cost of additional power dissipation.

The inductive powering and data receiving were also characterized. The operating frequency of the inductive link is 1.5 MHz. The frequency is chosen as it provides much stronger magnetic field compared to that at higher frequencies. Also, since the frequency is very low, the power transfer link causes negligible inference in the TX data transmission. The max power transfer efficiency was 28% with a 4 mA load current, and 40% with a 10 mA load current.

V. DISCUSSION: POTENTIAL FOR EXTENSION TO VCO-FREE ALL-WIRELESS MODE

The proposed UWB-IR TX architecture has an additional advantage of being easily integratable with resonant inductive power harvesting front-ends. The main design feature that eases the integration is the presence of a resonant LC tank in both the air and the antenna. The LC tank is driven by a synthesizer and a PA. The synthesizer generates the high-frequency signal, which is amplified by the PA and transmitted to the LC tank. The LC tank receives power directly from the power receiver (Fig. 15). As shown in Fig. 15, the frequency synthesizer, pre-amplifier, and the PA all can be removed if the LC tank receives power directly from the power receiver via magnetic coupling. This eliminates the dominant power consumption required to generate the high-precision tone to drive the LC tank, which would significantly further improve the power efficiency for low-power applications, such as energy-efficient sensors.

One limitation of using direct-coupled LC tanks is that the power coil must resonate at the same frequency as the UWB-IR TX. The TX pulse generation frequency, however, may not always be the optimal frequency for the power transfer system. The operating frequency of inductive power transfer systems is often selected at the lower MHz frequency (usually 13.67 MHz ISM band) [7]. Higher MHz operating frequencies are often avoided because the coils become increasingly radiative at these high frequencies, the quality factor is more limited, and the allowable magnetic field intensity is also more limited [32]. Therefore, when directly coupled to an inductive power receiving inductor, the maximum practical data-rate of the UWB-IR TX is about 20Mbps-30 Mbps. However, even this reduced data-rate is still sufficient for many sensory microsystems.

Another challenge in sourcing the driving signal of the UWB-IR TX LC tank from the inductive power receiving coil is the lower rising and falling times of the clipped signal $V_2$ for the same center node voltage $V_c$ swing of the LC tank. This is again caused by the lower oscillation frequency of the tank, which is set by the inductive power transfer frequency. Fig. 16 shows the relationship between the rising and falling times of the clipped signal $V_2$, the operating frequency, and the center node voltage swing $V_c$. When the operating frequency is lowered, the quality factor of the LC tank must be increased proportionally. The increased quality factor and signal swing ensure that the rising time of the clipped signal $V_2$ remains the same, and so does the power and bandwidth of the generated UWB pulse, despite...
TABLE I
COMPARATIVE ANALYSIS OF ENERGY-EFFICIENT TX DESIGNS

| Architecture | JSSC ’16 [10] | JSSC ’17 [22] | JSSC ’19 [24] | JSSC ’21 [9] | JSSC ’17 [18] | JSSC ’19 [23] | ISSCC ’22 [1] | This Work |
|--------------|---------------|---------------|---------------|---------------|---------------|---------------|--------------|-----------|
| Modulation   | BPSK          | BPSK          | FM            | BPSK          | PPM           | MPPM          | Hybrid       | OOK       |
| CMOS Process | 130nm         | 28nm          | 65nm          | 28nm          | 130nm         | 65nm          | 28nm         | 130nm     |
| Area (mm²)   | 4.6           | 0.93          | 1.1           | 0.154         | 0.5           | 2.88          | 0.16         | 1.19      |
| Supply       | 1V            | +/-1.8V       | 1V            | 0.9V          | 0.5V          | 1.1V          | -            | 1.2/3.3V  |
| Bandwidth (GHz) | 7             | -             | 0.5           | 1             | 1.25          | 2             | -            | 2         |
| Power (mW)   | 22.6          | 0.65          | 0.575         | 4.9           | 0.47          | 7             | 9.69         | 3.7       |
| Data-rate (Mbps) | 1000         | 27.24         | 0.1           | 6.81          | 20            | 500           | 1660         | 230/46    |
| Energy/bit (pJ) | 102.2         | 14            | -             | 1.12          | 2.76          | 4.7           | 5.8          | 21        |
| Distance (mm) | 1000          | 50-300        | -             | -             | 100           | 500           | 150^*        | 2000      |
| BER          | 10^-3         | -             | 10^-3         | -             | -             | 10^-3         | 10^-4        | 10^-6‡    |
| TX Efficiency | 0.59%         | 2.6%          | 12.2%         | 4.3%          | 11.7%         | 3.29%         | 10.32%       | 21.35%    |

*This distance includes a 15 mm thick tissue and an implant antenna.
†This BER was measured with 230Mb/s at 1m. It can also be achieved at 2m with a 46Mb/s data rate.

Fig. 16. Illustration of the increased requirement in quality factor when the LC tank is coupled to a low-frequency inductive resonator.

VI. COMPARATIVE ANALYSIS

Table I compares key specifications of the presented work with the state-of-the-art TX designs, including both pulse-radio and carrier-based architectures. Fig. 17 plots the power efficiency as a function of the data rate for this and other existing designs. As discussed in Section I, the output power efficiency of existing UWB-IR TX architectures is low, leading to a limited operating range. Fig. 17 compares the TX power efficiency and data rate of the state-of-the-art and the proposed design. It is challenging to achieve both high power efficiency and high data rate simultaneously, as high data rate often comes at the cost of circuit complexity and power overhead, such as a more complicated modulation scheme [1]. The proposed work represents a good trade-off and design improvement in both dimensions. Owing to the proposed clipped-sinusoid pulse generation scheme, the presented work shows a TX power efficiency of 21.3%, at a data-rate of 230 Mb/s, which is the highest reported power efficiency, even when including both carrier-based narrow-band TXs and pulse radios, as shown in Table I, on the left and right panels, respectively. Unlike several designs operating at short transfer distances [1], [18], [22], this design’s performance was achieved at an over-the-air distance of up to 2 m. This longer distance is essential for many biomedical sensors and body-area sensor networks applications [16], including electronic neural interfaces in freely behaving animals [6]. Finally, the energy consumption is calculated to be 21pJ/b, which is partially limited by the conservative 130 nm CMOS technology used for prototyping. Although the design in [18] was able to achieve a lower energy per bit performance using the same CMOS node, it was designed to operate at a distance of 100 mm, which is 20x shorter than this work. In a technology with smaller feature size, a lower supply voltage should yield further reduced energy.
consumption. In addition, techniques such as phase scrambling can be further used to reduce the spurious tones in the TX output spectrum. The corresponding small power consumption overhead due to such an implementation may result in a SNR improvement that directly benefits the BER.

VII. CONCLUSION

In this paper, a power-efficient clipped-sinusoid UWB-IR TX design was presented. Simultaneous powering and data transmission is supported. A 130 nm prototype was fabricated and tested. A state-of-the-art TX power efficiency of 21.3% at a data-rate of 230 Mb/s has been achieved. A BER of less than 10^{-6} was measured at 46 Mbps over a distance of 2 m, and 230 Mbps over a distance of 1 m. The energy consumption is 21 pJ/λ. Additionally, the design can be configured in a VCO-free all-wireless mode for even lower power low-data-rate applications. The proposed clipped-sinusoid UWB-IR TX design can be deployed in a broad range of applications, especially in power-constrained biomedical wearable and implantable sensory microsystems with high data rate requirements or when wireless powering and data receiving are desirable.

REFERENCES

[1] M. Song et al., “A 1.66 Gb/s and 5.8 pJ/b transcutaneous IR-UWB telemetry system with hybrid impulse modulation for intracranial brain-computer interfaces,” in Proc. IEEE Int. Solid-State Circuit Conf., 2022, pp. 394–396.

[2] M. Zhang et al., “Electronic neural interfaces,” Nature Electron., vol. 3, no. 1, pp. 61–65, 2020.

[3] A. Akinin et al., “An optically addressed nanowire-based retinal prosthesis with wireless stimulation waveform control and charge telemeasuring,” IEEE J. Solid-State Circuits, vol. 56, no. 11, pp. 3263–3273, Nov. 2021.

[4] R. Liu et al., “A 264-μW 802.15.4a-compliant IR-UWB transmitter in 22 nm FinFET for wireless sensor network application,” in Proc. IEEE Radio Freq. Integ. Circuits Symp., 2018, pp. 164–167.

[5] G. Gagnon-Turcotte et al., “A 0.13um CMOS SoC for simultaneous multi-channel optogenetics and neural recording,” IEEE J. Solid-State Circuits, vol. 53, no. 11, pp. 3087–3100, Nov. 2018.

[6] G. Soltani, M. S. Aliroteh, M. T. Salam, J. L. Perez Velazquez, and R. Genov, “Low-radiation cellular inductive powering of rodent wireless brain interfaces: Methodology and design guide,” IEEE Trans. Biomed. Circuits Syst., vol. 10, no. 4, pp. 920–932, Aug. 2016.

[7] X. Liu et al., “A fully integrated wireless compressed sensing neural signal acquisition system for chronic recording and brain machine interface,” IEEE Trans. Biomed. Circuits Syst., vol. 10, no. 4, pp. 874–883, Aug. 2016.

[8] S. Geng, D. Liu, Y. Li, H. Zhuo, W. Rhee, and Z. Wang, “A 13.3 mW 500 Mb/s IR-UWB transceiver with link margin enhancement technique for meter-range communications,” IEEE Int. Solid-State Circuit Conf. Dig. Tech. Papers, 2014, pp. 160–161.

[9] G. de Streel et al., “SleepTalker: A ULV 802.15.4a IR-UWB transmitter SoC in 28-nm FDSOI achieving 14 pJ/λt with channel selection based on adaptive FBB and digitally programmable pulse shaping,” IEEE J. Solid-State Circuits, vol. 52, no. 4, pp. 1163–1177, Apr. 2017.

[10] G. Lee et al., “An IR-UWB CMOS transceiver for high-data-rate, low power, and short-range communication,” IEEE J. Solid-State Circuits, vol. 54, no. 8, pp. 2163–2174, Aug. 2019.

[11] V. Kopta and C. Enz, “A 4-GHz low-power, multi-user approximate zero-IF FM-UWB transceiver for IoT,” IEEE J. Solid-State Circuits, vol. 54, no. 9, pp. 2462–2474, Sep. 2019.

[12] H. Rahmani and A. Babakhani, “A wirelessly powered reconfigurable FDD radio with on-chip antennas for multi-site neural interfaces,” IEEE J. Solid-State Circuits, vol. 56, no. 10, pp. 3177–3190, Oct. 2021.

[13] S. A. Mirbizioi, H. Bahrami, M. Sawan, L. A. Ruch, and B. Gosselin, “A single-chip full-duplex high speed transceiver for multi-site stimulating and recording neural implants,” IEEE Trans. Biomed. Circuits Syst., vol. 10, no. 3, pp. 643–653, Jun. 2016.

[14] X. Liu et al., “A fully integrated wireless sensor-brain interface system to restore finger sensation,” in Proc. IEEE Int. Symp. Circuits Syst., 2017, pp. 1–4.

[15] N. Soltani, H. Kassiri, H. M. Jafari, K. Abdelhalim, and R. Genov, “0.13 um CMOS 230 Mbps 21pJ/b IR-UWB transmitter with 21.3% efficiency,” in Proc. 41st Eur. Solid-State Circuit Conf., 2015, pp. 352–355.

[16] G. Kraus et al., “Technology of deep brain stimulation: Current status and future directions,” Nat. Rev. Neurol., vol. 17, no. 2, pp. 75–87, 2021.

[17] G. de Streel, J. Krauss et al., “Technology of Deep Brain Stimulation: Current Status and Future Directions,” Science, vol. 369, no. 6505, pp. 989–995.

[18] “IEEE Standard for Safety Levels With Respect to Human Exposure to Electric, Magnetic, and Electromagnetic Fields, 0 Hz to 300 GHz,” IEEE Std C95.1-2019, 2021 edition.

[19] “IEEE Standard for Safety Levels With Respect to Human Exposure to Electric, Magnetic, and Electromagnetic Fields, 0 Hz to 300 GHz,” IEEE Std C95.1-2019/Cor 1-2019, Oct. 2019, pp. 1–312.
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