A Scalable, 2.9 mW, 1 Mb/s e-Textiles Body Area Network Transceiver with Remotely-Powered Nodes and Bi-Directional Data Communication

Citation

As Published
http://dx.doi.org/10.1109/JSSC.2014.2328343

Publisher
Institute of Electrical and Electronics Engineers (IEEE)

Version
Author's final manuscript

Accessed
Sun Nov 25 18:57:44 EST 2018

Citable Link
http://hdl.handle.net/1721.1/98890

Terms of Use
Creative Commons Attribution-Noncommercial-Share Alike

Detailed Terms
http://creativecommons.org/licenses/by-nc-sa/4.0/
A Scalable, 2.9 mW, 1 Mb/s e-Textiles Body Area Network Transceiver with Remotely-Powered Nodes and Bi-Directional Data Communication

Nachiket Desai, Student Member, IEEE, Jerald Yoo, Member, IEEE, and Anantha P. Chandrakasan, Fellow, IEEE

Abstract—This paper presents transceivers and a wireless power delivery system for a Body-Area Network (BAN) that uses an e-textiles-based physical layer (PHY) capable of linking a diverse set of sensor nodes monitoring vital signs on the user’s body. A central base station in the network controls power delivery and communication resource allotment for every node using a general-purpose on-chip Node Network Interface (NNI). The architecture of the network ensures fault-tolerance, reconﬁgurability and ease of use through a dual wireless-wireline topology. The nodes are powered at a peak end-to-end efﬁciency of 1.2% and can transmit measured data at a peak rate of 1 Mb/s. Modulation schemes for communication in both directions have been chosen and a Medium Access and Control (MAC) protocol has been designed and implemented on chip to reduce complexity at the power-constrained nodes, and move it to the base station. While transferring power to a single node at maximum efﬁciency, the base station consumes 2.9 mW power and the node recovers 34 µW, of which 14 µW is used to power the network interface circuits while the rest can be used to power signal acquisition circuitry. Fabricated in 0.18 µm CMOS technology, the base station and the NNI occupy 2.95 mm^2 and 1.46 mm^2 area respectively.

Index Terms—Body-area networks, e-textiles, continuous health monitoring, wireless power delivery, inductive links, integrated medium access protocol

I. INTRODUCTION

BODY-area networks (BANs) allow continuous monitoring of vital signs by linking small, low-power sensors implanted in and placed on the surface of the body and transmitting the data they measure to healthcare providers [1]. Current state-of-the-art biomedical front-ends [2]–[4] can measure signals using only microwatts of power, making it possible to aggressively minimize their volume and the amount of energy stored locally. In order to realize their potential afforded by these advantageous traits, it makes sense to transfer the data using a two step process. Uncompressed data is first sent to a local base station (like a cellphone) using a link requiring minimal transmission effort by the sensor node. The local base station then processes the signals and sends them on to a healthcare provider using conventional data networks.

Wireless communication devices forming a BAN typically operate in the unlicensed bands, and have to contend with the high losses in those bands in the operating environment [5]. Another scheme for communication with the local base station is the use of body-coupled communication (BCC) [7]–[9]. BCC transceivers avoid a number of multiple-access problems when a large number people operate BANs in close proximity. However, the path loss in BCC channels is comparable to wireless channels. In contrast, textiles with electric routing (e-textiles) printed on top of the cloth or woven in as part of the yarn offer much lower path loss, and have long been a topic of research with the aim of building wearable computers [10]. They allow remote powering of the microwatt sensors and their associated telemetry circuits by a network base station [11]–[13]. Building on these advantages, the proposed BAN improves efficiency by using resonant power transfer between coupled inductors and combines this with electric routing on clothing to afford better aesthetics and greater comfort to the user by avoiding wires going all the way to sensors adhered to the body. This allows for cheap, disposable sensors that do not need an integrated battery or radio and can be easily attached to and removed from the user’s body depending on his/her specific requirements. The system architecture and the circuits used for resonant power transmission and recovery are presented in Sections II and III respectively. Modulation schemes for communication to and from the nodes have been chosen to lower the power requirements at the nodes and push complexity to the base station. These modulation schemes and the circuits implementing them are presented in Section IV.

Medium Access and Control (MAC) protocols used in commercially-deployed standards for wireless BANs, such as Bluetooth LE and IEEE 802.15.4/Zigbee [14] include significant protocol overhead at the sensor nodes and are not compatible with microwatt-sensors. Allowing sensors to sleep and requiring them to wake up after a preset amount of time [15] to check if the channel is unoccupied requires an accurate, power-hungry clock at the sensor that is not power-gated. A MAC protocol for e-textile based BANs that relies on Time-Division Multiple Access (TDMA) for resource allotment has been presented in [11]. A custom MAC protocol that moves all decision-making to the base station to reduce power consumption at the sensor nodes has been proposed and integrated on chip. The custom MAC protocol also allows sensors to be added and removed easily in order to keep the network...
reconfigurable and supports TDMA and Contention Access (CA) modes for sensors with wide variation in channel access and bandwidth requirements. Details of the implemented MAC are presented in Section V.

II. System Architecture

A. e-Textiles Network

Figure 1 shows an example of the proposed network implemented on a shirt with the user holding the local base station, whose size is limited by the test and debug structures on the board. Spiral inductors and their associated routing that connect them to a central base station have been screen-printed on fabric using the process described in [16]. Restricting the circuits on textiles to screen-printed passives avoids expensive procedures to bond silicon dies to the routing on fabric and keeps the costs of the textiles low. These inductors form near-field magnetic links with similar inductors on the sensor nodes, as shown in Figure 2. These nodes are of a band-aid patch form factor and are worn on the body. The local base station can be integrated in a cellphone linked to the shirt with a snap-button interface or a secondary inductive link (with additional losses), or onto a clip-on bluetooth-like device linked to the user’s cellphone. This hybrid wireless/wired architecture achieves better power efficiency compared to fully wireless- or BCC-based designs while being more comfortable to use than fully wired systems, such as the one presented in [11].

To maximize coverage and allow different varieties of sensors to operate in the same network, a large number of inductors must be placed on the e-textiles clothing. Additionally, e-textiles need to be designed keeping in mind that clothes are subject to harsher environments than most electronics used in daily life, for example inside a washing machine. In order to prevent network faults from arising due to a break (“opens”), redundant paths must be included in the network routing. Inductors are grouped to form sub-networks that can be turned on/off independently, as shown in Figure 3. Inductors in every sub-network are connected in parallel and the sub-network as a whole has multiple independent paths connecting it to the base station. This arrangement allows for a trade-off between routing complexity when each sub-network contains only one inductor and high power consumption when all inductors are in the same sub-network. Also, by not adhering to a strict arrangement like the (X,Y) grid in [12] or the linked list in [17], the proposed network architecture allows for greater freedom in covering sensors around the body based on user-specific sensor locations. An obvious arrangement would be to form sub-networks of inductors covering one part of the body. The only limitations would be the number of sub-networks that can be supported by the base station and the combined inductance of the parallelly-connected coils in each sub-network. For example, the shirt in Figure 1 can support 2-channel ECG while a cap made of similar material can function as an EEG helmet.

Power is transferred to the sensor nodes by resonant coupling between inductors at the base station and the sensor nodes. The network has a star topology, with all sensor nodes talking directly to the base station. Modulation schemes for data transmitted in both directions have been chosen to push most of the modulation and demodulation effort away from the sensor nodes towards the base station.

B. Base Station

A block diagram of the implemented base station for the network-on-shirt of Figure 1 is shown in Figure 4. It can support up to two sub-networks each containing up to two inductors. The base station has a separate analog interface for each of the two sub-networks it can support. Both sub-networks share the same digital back-end. The demodulated bitstreams from both the sub-networks are combined asynchronously in the digital domain and sent on to the digital back-end that implements the on-chip MAC and manages power delivery to the sensor nodes in the network. A regulator with four discrete output levels controls the amount of power transmitted to each sub-network. A digital on/off controller switches off a sub-network if it is devoid of active sensor nodes and doubles up as an on-off keying (OOK) modulator for downlink transmission. The analog interface to each sub-network consists of the front-end oscillators, power amplifiers and LC-tank filters for power transmission, and modulators and demodulators for data communication.
C. Sensor Nodes

Sensor nodes are partitioned into an application-specific biomedical data acquisition module and a general-purpose NNI, as shown in Figure 5. The on-chip NNI module supplies power to the sensor readout circuitry and transmits the data measured by it. The off-chip readout circuits consist primarily of a front-end amplifier followed by an ADC and can be any state-of-the-art microwatt-level sensor that measures signals on the human body [2]–[4]. An auto-commutative rectifier (ACR) minimizes the dead zone for rectification at low-amplitude ac inputs, and converts ac power transmitted by the base station to dc for use by all circuits at the sensor node. A low power, fully digital demodulator passes on instructions sent by the front-end and control how and when it is packetized and transmitted to the base station for interpretation. These instructions directly manage the data collected by the sensor front-end and control how and when it is packetized and transmitted to the base station.

III. Resonant Power Delivery Interface

A. Theoretical Overview

An RLC circuit model for power transfer across the inductive link is shown in Figure 6. The resonant near-field power transfer scheme implemented is similar to the one described in [18], except that a parallel LC-tank is employed on the primary (base station) side. The resistor $R_L$ in Figure 6 models the ac input resistance of the rectifier that supplies dc power to the sensor node.

The fabric inductors used have low quality factors ($Q \approx 6 – 8$) in the MHz range, hence the overall loaded $Q$ of the secondary resonator is dominated by the inductor losses and not the load. Also, the coupling coefficient across the inductive link, $k$, is approximately 0.1 for a separation as low as 5 mm between the primary and secondary inductors. In this regime of operation, the total efficiency of power transfer

1Assuming the rectifier supplying dc power to the sensor node consumes as much as 90 $\mu$A (p-p) current at 2.4 V (p-p) to supply 30 $\mu$A at 1.8 V dc, this yields $R_L=26.7$ k$\Omega$ and $Q_L \approx 100$. 

Fig. 3. Architecture of proposed BAN. Each sub-network can power and talk to a heterogeneous group of sensors. The NNIs can access a sub-network through any one or more of its constituent inductors.

Fig. 4. Block diagram of base station.

Fig. 5. Block diagram of a typical sensor node implementation. Only the Node Network Interface has been implemented on chip, while the application-dependent sensing and data conversion circuits are off-chip.

Fig. 6. Power transfer through inductively coupled resonators. The mutual inductance $M$ and the coupling coefficient $k$ are related as $M = k \sqrt{L_1 L_2}$. 

1Assuming the rectifier supplying dc power to the sensor node consumes as much as 90 $\mu$A (p-p) current at 2.4 V (p-p) to supply 30 $\mu$A at 1.8 V dc, this yields $R_L=26.7$ k$\Omega$ and $Q_L \approx 100$. 
across the inductive link can be approximated as

\[ \eta \approx \frac{k^2 Q_1 Q_2}{1 + k^2 Q_1 Q_2} \cdot \frac{Q_2}{Q_L} \tag{1} \]

where \( Q_L \), the quality factor of load, is defined as \( Q_L \triangleq \omega_0 C_2 R_L \) and \( Q_1 \) and \( Q_2 \) are the intrinsic quality factors of the primary and secondary inductors respectively. Equation (1) can be obtained from the expressions derived in [19].

The voltage transfer function from the primary to the secondary can be approximated as

\[ \left| \frac{V_s(j\omega_0)}{V_p(j\omega_0)} \right| \approx \frac{k \sqrt{Q_1 Q_2}}{1 + k^2 Q_1 Q_2} \sqrt{\frac{L_2}{L_1}} \tag{2} \]

Increasing the coupling coefficient and the \( Q \) of the resonators improves the efficiency of power transfer and the magnitude of the voltage transfer function, as shown by Equations (1) and (2). From the expressions for inductance in [16], a 6 turn spiral inductor with outer diameter of 30 mm has 1.6 \( \mu \)H inductance. This agrees with the measured values of 1.4 \( \mu \)H inductance. The measured self-resonant frequency of the inductors is 80 MHz, and the 27.12 MHz ISM band was chosen for operation since it is sufficiently below the self-resonant frequency for the inductors to remain dominantly inductive and the on-chip capacitors required are around 20 pF. These values lead to a link (ac-ac) efficiency of 1.7% and voltage transfer function 0.39 at the lowest power setting of the transmitter.

**B. Combined Oscillator-PA Transmitter**

The circuit used for resonant power transfer is shown in Figure 7. The oscillator uses a modified Colpitts’ feedback structure, which also doubles as the output filter. The filter is made of inductors on fabric and on-chip MIM capacitors. A parallel LC tank is used to avoid the large voltage stress that would arise from resonating on-chip capacitors in a series LC configuration. A class-D power amplifier is integrated into the oscillator feedback loop. Output power control is achieved by reducing the active device widths of the transistors in the PA to make them function as linear regulators in the lower power settings. This avoids both requiring two separate tuned networks in the MHz range and tuning those two networks to resonate at the same frequency. The on-chip capacitors employ two-level tuning in order to maintain operation at the desired frequency. Coarse tuning selects the number of capacitors to be connected in the network depending on the number of inductors connected in parallel in the sub-network. Fine tuning adjusts for variations in the inductance values of the fabric inductors; a one-time calibration step can tune out up to 20% static variation using four bits, which is sufficient based on inductance measurements made on 24 samples.

A small-signal model of the oscillator is shown in Figure 8. The resistance \( R_s \) is the small-signal output resistance of the class-D PA. The loop transfer function for the oscillator has three poles and no zeros, making the system oscillate above a minimum dc loop gain. This dc loop gain is provided by the small-signal amplification of the class-D PA, which is dc-biased around mid-rail by negative feedback. The small-signal model is obtained by linearizing the circuit at the dc bias point.

The oscillator oscillates at \( f_0 = (2\pi \sqrt{L C_{eq}})^{-1} \), where \( C_{eq} = C_1 || C_2 \). For a desired resonant frequency and given inductance value, the area of the on-chip capacitors is minimized when \( C_1 = C_2 \). This also leads to \( V_o \) and \( V_f \) oscillating between 0 and \( V_{DD} \) in opposite phase, maximizing the voltage across the fabric inductor.

**C. Auto-Commutative Rectifier**

AC power transmitted by the base station is rectified at the sensor nodes by using a self-synchronous topology similar to the one presented in [20], [21], which trades off efficiency for the capability to generate workable dc voltages from ac input amplitudes small enough to limit the transistors' performance by their threshold voltages. A schematic of the auto-commutative rectifier (ACR) used is shown in Figure 9. The gates of all transistors are biased such that the devices conduct strongly in the desired half-cycle and are strongly turned off in the other. The use of additional comparators and gate-drive circuits is avoided in order to extract net positive power in the \( \mu \)W region of operation. Additionally, no charge is stored on capacitors, which might be leaky or expensive to integrate, in order to overcome the threshold voltages of the transistors.

Simulation waveforms for the operation of the ACR are shown in Figure 10. The transistors \( M_{p1} \) (or \( M_{p2} \)) and \( M_{n1} \) (or \( M_{n2} \)) in Figure 9 are supposed to conduct when \( V_{AC_n} \) (or \( V_{AC_p} \)) is higher than \( V_{DC+} \) and \( V_{AC_n} \) (or \( V_{AC_p} \)) is lower than \( V_{DC-} \). Reverse conduction is possible during the initial phase.

\[ 2 \text{The value of } R_s \text{ for the highest and lowest power settings of the transmitter are approximately } 140 \, \Omega \text{ and } 480 \, \Omega \text{ respectively.} \]
of each half cycle and is kept lower than the forward current by the threshold voltages of the transistors. The net dc current supplied to the load in Figure 10 is 11 $\mu$A.

The transistors making up the ACR in Figure 9 need to be sized to carry currents much larger than the output load current supplied, as shown in Figure 10. However, compared to a conventional rectifier where the same transistors are diode-connected, the ACR supplies 2x higher output voltage than the conventional diode-connected rectifier in the low-input amplitude regime of operation. This is shown in Figure 11.

To generate the required dc output voltage, the four rectifiers in Figure 9 are connected in series in their dc path while their ac inputs are capacitively-coupled to the LC tank as shown in Figure 12 thus avoiding the need for additional voltage boost converters.

IV. BI-DIRECTIONAL DATA COMMUNICATION TRX

The capability to transmit data in both directions is necessary for the star network topology with the base station acting as the hub. The requirements for the links in both directions are highly asymmetric: the downlink from the base station to the sensor nodes needs to carry only small, low-duty cycle control instructions. On the other hand, the uplink from the sensor nodes to the base station needs to carry potentially large amounts of uncompressed data. An added constraint is the limited energy budget at the sensor node and the need to push complexity to the base station.

Various schemes for transferring data to inductively-coupled sensors have been proposed in [22]–[25], but they use modulation schemes such as FSK and OQPSK, which are more complex than OOK-based schemes, or use multiple inductors, or are data-only links. Since the sensor nodes have to be of a small form factor the data link required for e-textile BANs must use the same inductors for power and data transfer in both directions. A pulse-width modulated (PWM) OOK scheme is used on the downlink that can be demodulated asynchronously, eliminating the need for clock recovery circuits at the sensor node. On the uplink, impedance modulation is used which reduces the modulation circuitry to a switch across the inductor at the sensor node.

A. Downlink Communication

Binary coded messages to be transmitted to the sensor nodes are pulse-width coded digitally using LOW and HIGH times in a 25:75 ratio for transmitting a “1” and a 75:25 ratio for transmitting a “0”. This pulse-width-encoded signal is used to control a transmit on/off switch which generates the PWM-OOK waveform shown in Figure 13. In order to accommodate the startup/shutdown times of the oscillator, the data rate chosen is 80 kb/s, much lower than the local clock frequencies at the base station and sensor node.

Demodulation at the sensor node is done digitally using only the local 1 MHz clock. Two incrementers count the number of clock cycles for which the received waveform is 0 and 1, as shown in Figure 14. Since each PWM bit ends with a negative edge, the values of the counters can be compared against each other at the negative edge and the received bit can be latched. Fully digital demodulation avoids static power
dissipation, which is significant since downlink messages have a very low duty cycle. Additionally, the demodulator is self-referenced, i.e. it does not need a reference for the comparator, the setting of which would require accurate calibration. Sample waveforms for the case $T_{PWM} = 4T_{OOK}$ are shown in Figure 15.

### B. Uplink Communication

Uplink data is transmitted by short- and open-circuiting the switch across the sensor node LC tank in Figure 5, which changes the impedance reflected back to the base station inductor and can be sensed as voltage $V_1$ in Figure 5. During uplink transmission, the sensor node primarily relies on stored energy instead of rectifying the received ac power. The signal received at the base station, shown in Figure 13, is an ASK-modulated waveform with non-unity modulation index. Under the conditions outlined in Section III, the expressions for the modulation index of the received waveform $m$ derived in [19] can be approximated as

$$m \approx \frac{k^2 Q_1 Q_2}{1 + k^2 Q_1 Q_2} \cdot \frac{R_s}{R_s + Q_1 \omega_0 L_1} \tag{3}$$

The ASK modulation index at the base station evaluates to approximately 0.08 for the parameter values listed in Section III.

Demodulation is done differentially by extracting and amplifying both the positive and negative envelopes of the incoming signal shown in Figure 13. Messages arriving from different sub-networks need to be combined to have a single clock recovery circuit and digital back-end that handles data integrity and network access. Combining the impedance modulated ASK waveforms from the different sub-networks in the analog domain results in the received voltage signal being measured across an impedance that progressively decreases as the number of sub-networks increases. Instead, the received signal from each sub-network is demodulated separately as shown in Figure 4 and multiplexed in the digital domain. As it comes before the clock recovery circuit in the receive chain, the digital multiplexer needs to be asynchronous. Figure 16 shows the schematic of the asynchronous multiplexer. The output of each demodulator is “0” when there is no data being transmitted from that sub-network. The circuit shown in Figure 16 can easily be extended to more than two sub-networks.

### V. MEDIUM-ACCESS PROTOCOL

An application-independent health-monitoring BAN needs to be able to support different kinds of sensor nodes that have different channel access and bandwidth requirements. These could range from ECG and single-channel EEG recorders that need a time-averaged data rate of up to 10 kb/s, to temperature, blood pressure or blood oxygenation monitors that need infrequent access and a time-averaged data rate of only a few bits/s. In order to accommodate such a variety of sensor nodes, a hybrid MAC protocol has been designed and integrated on silicon along with the power and communication TRX circuits. The hybrid MAC uses both TDMA access and CA modes for nodes requiring frequent and infrequent access respectively. It also ensures that all decision-making is pushed to the base station, with the sensor nodes only following instructions sent to them.

Due to the architecture of the network, messages on the downlink are broadcast to all sensor nodes in a sub-network. Downlink messages are 31b long, and contain a 4b header that classifies the type of the message. The format of the following 27 bits varies based on the 4b header. For example, it could contain the addressed sensor node’s network ID in...
network access requirements. The base station then assigns a 16b pseudorandom value and responds with details about its configuration beacon, each sensor node backs off in time by a pseudorandom 16b value upon receiving the beacon. The end of the 4 second period (1 TDMA phase + 1 CA phase) is again broadcast to all sub-networks and the base station goes back to looping through all the TDMA sensors.

C. Contention Access Phase

After completion of the TDMA phase, the network moves to the CA phase. The start of CA is denoted by a beacon broadcast by the base station to all sub-networks. Since uplink data must cross two inductive hops to be heard by other sensor nodes, a CSMA-based protocol cannot be used for time multiplexing. Instead, CA nodes with data to transmit back off in time by a pseudorandom 16b value upon receiving the beacon. The end of the 4 second period (1 TDMA phase + 1 CA phase) is again broadcast to all sub-networks and the base station goes back to looping through all the TDMA sensors.

VI. MEASUREMENT RESULTS

The base station and the NNI for the e-textiles network have been fabricated in a 0.18 \( \mu \)m CMOS process. Die photographs are shown in Figure 20. A brief summary of the design characteristics are presented in Table II and performance comparisons with other recently reported solutions in the same application space are presented in Table III. The power consumption of the base station during transmit/receive-mode is 2.9 mW, which is comparable to the narrowband RF- and BCC-based systems in Table II. The energy per bit on the uplink is more important since the downlink only carries small, low duty-cycle packets. At 1 Mb/s, the base station expends 2.9 nJ per uplink bit. The 14 \( \mu \)W maximum power consumption by the communication and digital baseband circuits at 1 Mb/s translates to the sensor node’s expended energy per uplink bit approaching the pJ/bit.
TABLE I
<table>
<thead>
<tr>
<th>Summary of E-textiles Network Characteristics</th>
</tr>
</thead>
<tbody>
<tr>
<td>Base Station</td>
</tr>
<tr>
<td>Technology</td>
</tr>
<tr>
<td>Area</td>
</tr>
<tr>
<td>VDD,min</td>
</tr>
<tr>
<td>fCLK</td>
</tr>
<tr>
<td>Power Consumption</td>
</tr>
<tr>
<td>Frequency Band</td>
</tr>
<tr>
<td>Modulation Scheme</td>
</tr>
<tr>
<td>Max. Sensor Count</td>
</tr>
<tr>
<td>Multi Access</td>
</tr>
</tbody>
</table>

Figure 21 shows the efficiency of power transfer across resonant inductive link and rectifier output voltage for varying load current. The separation between the transmitter and receiver coils in this case was 5 mm and the transmitter PA was at its lowest power setting with 2 V supply.

range seen in fully wired systems [11], [17]. The higher power consumption at the base station, where energy is more readily available, trades off against greater comfort due to the wireless/wired hybrid architecture when compared against fully wired systems.

Figure 21 shows the efficiency of power transfer and the voltage at the output of the sensor node’s rectifier for varying load currents. As the load current increases beyond 20 µA, the rectifier operation is limited by the series ac-coupling capacitor reactances and the output voltage falls rapidly. At its lowest power setting, the power transmitter on the base station supplying one inductor consumes 2.7 mW with the PA operating from a 2 V supply. Owing to the weak inductive link, the power drawn by the transmitter varies negligibly with changes in the load current of the rectifier. The received power peaks at 20 µA load current and the peak dc-to-dc power transfer efficiency across the link is 1.2% with voltage transfer function of 0.29, which are smaller by approximately 25% than the values calculated in Section III since Equations (1) and (2) ignore the effects of the source resistance $R_s$ in Figure 6. Of a maximum 34 µW power recovered at the sensor node, a maximum of 14 µW is consumed by the communication and digital baseband circuits, leaving the remaining 20 µW for the biomedical signal acquisition front-end which is sufficient to power state-of-the-art biomedical data-acquisition circuits [2]–[4]. A plot of the measured efficiency and output power normalized to its maximum value with varying misalignment is presented in Figure 22. The base station increases transmitted power with increasing misalignment and is capable of supplying the node’s minimum 14 µW power requirement for misalignments up to half the radius of the primary inductor.

OOK and envelope-detected waveforms for a downlink PWM message are shown in Figure 23. The waveforms correspond to the initial beacon transmitted by the base station asking sensor nodes to join the network. Waveforms pertaining to the packet metadata from an uplink packet with simulated ECG data are shown in Figure 24. The measured modulation index of 0.11 agrees with the value calculated theoretically.
TABLE II
COMPARISON OF BODY-AREA NETWORK COMMUNICATION SYSTEMS

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Type</strong></td>
<td>Narrowband RF</td>
<td>BCC</td>
<td>Wired+Wireless</td>
<td>Wired e-textiles</td>
<td>Wired e-textiles</td>
<td>Wired+Wireless e-textiles</td>
</tr>
<tr>
<td><strong>Coverage</strong></td>
<td>Wide</td>
<td>Local</td>
<td>Local</td>
<td>Local</td>
<td>Wide</td>
<td>Wide</td>
</tr>
<tr>
<td><strong>Network Topology</strong></td>
<td>Star</td>
<td>Master-slave (X,Y) grid</td>
<td>Bus</td>
<td>Star</td>
<td>Star</td>
<td></td>
</tr>
<tr>
<td><strong>Base Station Power</strong></td>
<td>2.3 mW (^b)</td>
<td>3.2 mW (^b)</td>
<td>5.2 mW</td>
<td>75 µW</td>
<td>180 µW</td>
<td>2.9 mW</td>
</tr>
<tr>
<td><strong>Sensor Node Power</strong></td>
<td>2.7 mW (^b)</td>
<td>1.6 mW (^b)</td>
<td>12 µW</td>
<td>25 µW</td>
<td>20 µW</td>
<td>14 µW (^c)</td>
</tr>
<tr>
<td><strong>Data Rate</strong></td>
<td>50 kb/s</td>
<td>1.25 Mb/s</td>
<td>120 kb/s</td>
<td>20 Mb/s</td>
<td>10 Mb/s</td>
<td>1 Mb/s (uplink), 80 kb/s (downlink)</td>
</tr>
<tr>
<td><strong>Power Transfer Mechanism</strong></td>
<td>-</td>
<td>-</td>
<td>Inductive</td>
<td>Wired</td>
<td>Wired</td>
<td>Resonant inductive</td>
</tr>
<tr>
<td><strong>Power Transfer Efficiency</strong></td>
<td>0.2% (^d)</td>
<td>-</td>
<td>-</td>
<td>96%</td>
<td>1.2%</td>
<td></td>
</tr>
</tbody>
</table>

\(^a\) No remote power delivery to sensor nodes
\(^b\) Uplink communication power only
\(^c\) Power consumption of communication circuits and digital baseband circuits only
\(^d\) Based on power consumption values of network controller and supported sensor node

Fig. 22. Peak efficiency (dc-to-dc) of power transfer and peak output power for varying misalignment, measured as the center-center distance between the coils. The diameter of the primary inductor on clothing is 32 mm.

in Equation (3). The uplink message shown consists of an initial sequence of alternating 1s and 0s for clock recovery, followed by the random number ACK described in Section VI. The trailing bits of the message consist of the CRC checksum that ensures data integrity.

Figure 25 shows the network self-configuring upon startup. The example network consists of two sub-networks, each with one and two sensor nodes respectively. The base station sends the beacon shown in Figure 23 to each sub-network one at a time. Sensor nodes respond with messages similar to the one shown in Figure 24 after a random delay to avoid collisions. After waiting for 1 second for all sensor nodes in the sub-network to respond, the base station moves on to the next sub-network. In Figure 25, both sub-networks are fully configured at the end of 2 seconds.

VII. CONCLUSION

In order to address the problems of high path loss that BANs using radios and BCC face, a bi-directional communication TRX and wireless power delivery system based on e-textiles has been presented. The proposed e-textiles BAN has been designed across multiple layers to support a wide variety of sensors on the body. By delivering power wirelessly to state-of-the-art microwatt-level health monitors through resonant coupling, the problem of long-term energy storage can be addressed and their form factors can be reduced. Implemented in 0.18 µm CMOS, the base station, which acts as the hub in a star-connected network, consumes 2.9 mW per inductor it supplies power to. The network architecture has been chosen...
Voltage (V) 0 0.5 1 1.5 2
0 10 20 30 40 50 60 70 80

PLL Lock Sequence Preamble Random Number ACK

Time (µs)

0 10 20 30 40 50 60 70 80

Fig. 24. Measured waveforms for uplink packet header at input and output of base station demodulator.

Network Configuration Beacons

Sub-Network 0 Envelope Sub-Network 1 Envelope

SN_config1,0 SN_config0,0 SN_config0,1

2 SNs configured in sub-network 0 2 seconds 1 SN configured in sub-network 1

Fig. 25. Self-configuration of e-textiles BAN with two sub-networks. The bottom three traces track a bit that goes HIGH once the sensor node is configured.

to offer a good trade-off between flexibility in choosing any subset of inducators to power while reducing complexity on the clothing. A NNI that recovers power transmitted by the base station and handles data modulation and demodulation has also been implemented in 0.18 µm CMOS. It consumes 14 µW and harvests a maximum of 34 µW, leaving the remaining 20 µW for the biomedical front-end. Modulation schemes have been chosen keeping in mind the asymmetric nature of resource requirements for data flow in both directions, and such that complexity is pushed to the base station as much as possible. Finally, logic implementing a hybrid MAC protocol has also been integrated on chip, with all decisions being taken by the base station and the sensor nodes acting as slaves. The hybrid MAC uses different network access schemes for sensor nodes with different access and average data transfer rates.

ACKNOWLEDGMENT

The authors would like to thank members of the Semiconductor System Lab at KAIST, Daejeon, Korea for assistance with e-textiles fabrication and MOSIS for fabrication support.

REFERENCES


Nachiket Desai (S’10) received the Bachelor of Technology (B. Tech.) degree in electronics & electrical communication engineering from the Indian Institute of Technology, Kharagpur, India in 2010 and the S. M. degree in electrical engineering and computer science from the Massachusetts Institute of Technology, Cambridge, MA, USA in 2012, where he is currently working toward the Ph. D. degree.

Between June 2012 and August 2012, he was at Texas Instruments, Dallas, TX, USA, designing power converters for energy harvesting applications. His research interests include wireless power transfer for consumer devices and personal-area networks (PANs), circuits for near-field communication (NFC), and power converters.

Jerald Yoo (S05-M10) received the B.S., M.S., and Ph.D. degrees in electrical engineering from the Korea Advanced Institute of Science and Technology (KAIST), Daejeon, Korea, in 2002, 2007, and 2010, respectively.

In May 2010, he joined the faculty of Microsystems Engineering, Masdar Institute of Science and Technology, Abu Dhabi, United Arab Emirates, where he is currently an associate professor. During June 2010 to June 2011, he was also with Microsystems Technology Laboratory, Massachusetts Institute of Technology (MIT), Cambridge, MA, USA, as a visiting scholar. He developed low-energy Body Area Network (BAN) transceivers and wearable body sensor network using Planar-Fashionable Circuit Board (P-FCB) for continuous health monitoring system. His research focuses on low energy circuit technology for wearable bio signal sensors, wireless power transmission, SoC design to system realization for wearable healthcare applications, and energy-efficient biomedical circuits. He is an author of the book chapter in Biomedical CMOS IC’s (Springer, 2010).


Anantha P. Chandrakasan (M’95-SM’01-F’04) received the B.S, M.S. and Ph.D. degrees in Electrical Engineering and Computer Sciences from the University of California, Berkeley, in 1989, 1990, and 1994 respectively. Since September 1994, he has been with the Massachusetts Institute of Technology, Cambridge, where he is currently the Joseph F. and Nancy P. Keithley Professor of Electrical Engineering.

He was a co-recipient of several awards including the 1993 IEEE Communications Society’s Best Tutorial Paper Award, the IEEE Electron Devices Society’s 1997 Paul Rappaport Award for the Best Paper in an EDS publication during 1997, the 1999 DAC Design Contest Award, the 2004 DAC/ISSCC Student Design Contest Award, the 2007 ISSCC Beatrice Winner Award for Editorial Excellence and the ISSCC Jack Kilby Award for Outstanding Student Paper (2007, 2008, 2009). He received the 2009 Semiconductor Industry Association (SIA) University Researcher Award. He is the recipient of the 2013 IEEE Donald O. Pederson Award in Solid-State Circuits.


He has served as a technical program co-chair for the 1997 International Symposium on Low Power Electronics and Design (ISLPED), VLSI Design ‘98, and the 1998 IEEE Workshop on Signal Processing Systems. He was the Signal Processing Sub-committee Chair for ISSCC 1999-2001, the Program Vice-Chair for ISSCC 2002, the Program Chair for ISSCC 2003, the Technology Directions Sub-committee Chair for ISSCC 2004-2009, and the Conference Chair for ISSCC 2010-2014. He is the Conference Chair for ISSCC 2015. He was an Associate Editor for the IEEE Journal of Solid-State Circuits from 1998 to 2001. He served on SSCC AdCom from 2000 to 2007 and he was the meetings committee chair from 2004 to 2007. He was the Director of the MIT Microsystems Technology Laboratories from 2006 to 2011. Since July 2011, he is the Head of the MIT EECS Department.


