Abstract—Wireless body area network (WBAN) provides a means for seamless individual health monitoring without imposing restrictive limitations on normal daily routines. To date, Radio Frequency (RF) transceivers have been the technology of choice, however, drawbacks such as vulnerability to body shadowing effects, higher power consumption due to omnidirectional radiation and security concerns, have prompted the adoption of transceivers that use the human body channel for communication. In this paper, a vital signal monitoring transceiver prototype based on the human body channel communication (HBC), using commercially available chipsets is presented. RF and HBC communications are briefly reviewed and compared, and different schemes of HBC are introduced. A circuit model that represents the human body channel is then discussed and simulations are presented to illustrate the influence of the return path capacitance and receiver terminations on the path loss. The architecture of the transceiver prototype is then introduced where it is designed at a 21 MHz IEEE 802.15.6 standard-compliant carrier frequency. Finally, the performance of the transceiver, including the bit error rate (BER) and power efficiency, are characterized. Path loss is measured for two different scenarios, where variations of up to 5 dB were observed due to environmental effects. Energy efficiency measured at a maximum data-rate of 1.3 Mbps was found to be 8.3 nJ/b.

Index Terms—Human body communication, Internet of Bodies, wearable prototype, path loss simulation, vital signal monitoring

I. INTRODUCTION

An unprecedented number of physical objects and devices are being connected to the Internet by the Internet of Things (IoT) to share information and coordinate decisions. IoT devices have been widely applied in smart homes, smart cities and business applications [1]. A growing segment of IoT are IoT-connected devices that combine software, algorithms and computing capabilities to monitor an individual’s health conditions. These systems are referred to as the Internet of Bodies (IoB) [2]. IoT devices are usually flexible, swallowable, digestible, or injectable with a very small form factor, which can provide patients with a more comfortable experience while offering holistic healthcare. Compared with the traditional medical monitoring approach, IoT devices enable patients to be monitored easily while away from hospital settings. Patients can move freely and keep their normal daily routines. Fig. 1 illustrates the myriad of applications enabled by IoT devices.

Generally speaking, wearable health monitoring systems need to operate at extremely low power to extend battery life and reduce complexity. This is an area where HBC transceivers excel. HBC transceivers couple the signal directly to the human body, thus eliminating the need for complex and power hungry RF front ends, resulting in significant reduction in power consumption. Fig. 1 shows a comparison between the energy efficiency of recently published RF and HBC transceivers, where energy efficiency is defined as the energy needed in Joules to transmit and receive one bit. The figure shows that most RF transceivers operate between 10 nJ/b and 100 nJ/b, while state-of-the-art HBC transceivers operate between 0.01 nJ/b to 0.1 nJ/b. Therefore, the state-of-the-art HBC communication exhibits up to three order of magnitude better energy efficiency, enabling ultra low-power transceivers. Furthermore, due to the omnidirectional nature of RF communication, it is easily eavesdropped on by attackers so data encryption is necessary, which increases demand on energy. HBC, in comparison, transmits the signal in a more secure way since attackers have less opportunity to overhear the signal coupled through the human body.

In this paper, a low power wearable prototype for vital signal monitoring based on HBC and commercial-off-the-shelf (COTS) chipsets are introduced. The contributions of this paper can be summarized as follows:

- The influence of return path capacitance and receiver impedance on the channel loss is studied and quantified using Advanced Design System (ADS) tools. Experimental measurements are then used to confirm simulations results and illustrate the effect of the environment on the HBC channel path loss.
- The architecture of a healthcare monitoring transceiver prototype based on the HBC with COTS chipsets is
presented. The system supports up to 1.3 Mbps data rate.

- Electrocardiogram (ECG) transmission is successfully demonstrated as an exemplar application by using the proposed HBC transceivers. The system delivered a 8.3 nJ/bit energy efficiency. Even though the proposed system does not achieve sub 1 nJ/b as custom fabricated HBC transceivers do, it demonstrates that COTS HBC implementations can compete with custom fabricated RF solutions.

The remainder of the paper is arranged as follows: In section II, HBC modalities and prior work are presented. Section III presents the circuit model and ADS simulations on HBC characteristics to show the factors influencing the path loss of capacitive coupling HBC (CC-HBC); Section IV introduces the circuit and system design with the IEEE 802.15.6 standard 21 MHz carrier frequency; Section V presents the characterization of the designed system and illustration of an exemplar ECG signal transmission; Section VI concludes the paper.

II. HBC MODALITIES AND PRIOR WORK

There are two forms of [3]: the electrical field based HBC (eHBC), and magnetic field based HBC (mHBC). Depending on how the signal is electrically coupled to the body, eHBC can be further classified into galvanic coupling HBC (GC-HBC) and CC-HBC. In the GC-HBC, both signal and ground electrodes are in contact with the skin, thereby coupling both forward and return signals to the body and not affected by environmental changes. However, the GC-HBC operates best at low frequency and shorter distance due to path loss. On the other hand, CC-HBC applies the signal electrode to the skin while keeping the ground electrode floating in air. In this case, the return path is susceptible to the external environment and user postures. However, it can operate at higher frequencies, thus enabling high data rates as well as longer distance. Finally, the mHBC was proposed in 2014 [4], where the signal is transmitted by the magnetic field generated by a TX coil. When the hand touches the signal electrode and the foot electrode, an effective loop is formed, which forms a communication channel between the transmitter and the receiver. When either of them is detached, the loop is broken.

Numerous wearable, injectable, or digestible sensors have been invented for WBAN applications, including ECG, electromyography (EMG), electrooculography (EOG), blood pressure, and heart rate, etc [5]. Typically, these WBAN systems use traditional RF communications, for instance, Bluetooth, Zigbee, etc. To address the shortcomings of RF transceivers listed earlier, HBC transceivers have been proposed and studied. The HBC channels are characterized by numerical/physical body phantoms, equivalent circuit models, and empirical channel measurement campaigns. We refer interested readers to [3] for a in-depth survey of channel characterization issues. HBC transceivers based on COTS microcontrollers have been introduced. Fahier et al. [6] presented a real time ECG monitoring system with ADS1298 ECG sensor and FPGA platform. At the transmitter side, the data is encoded with Manchester code and the signal is modulated using On-Off-Keying (OOK) with a 30 MHz carrier. At the receiver side, the signal is amplified with a bandpass Sallenkey filter amplifier and digitized with a comparator before being decoded with FPGA. However, the FPGA platform is power-consuming, which is not suitable for healthcare systems that demand long battery life. Maity et al. [7] presented a wearable health monitoring device with a communication module, processing module, memory, power source, sensor and an interface with the human body. The system is based on the TM4C123G LaunchPad evaluation kit. The received signal is demodulated with a 12-bit Analog to Digital Converter (ADC) for logic decision. It achieves a low BER and 8.2X power efficiency improvement over the RF wireless system, but has a limited data rate at 500 kbps. Moreover, both of the designs lead to a high cost and large area when using development kits. There are several other ECG monitoring device based on HBC. Table. I shows a comparison of power consumption versus data rate of the previous devices.

III. VOLTAGE-MODE BODY CHANNEL SIMULATION

The circuit model of the voltage-mode body channel is shown in Fig. 2 [8]. This circuit includes an AC voltage source $V_{in}$, a source resistor $R_s$, the transmitter/receiver termination capacitances $C_{TX}/C_{RX}$, return path capacitance $C_{ret}$, the coupling between the signal/ground $C_{TX,grd}/R_{RX,grd}$ electrodes, skin-electrode contact impedance $R_{hand}$, skin voltage $V_{skin}$, skin impedance $R_{skin}$, body impedance $R_{body}$, as well as the receiver termination resistance $R_t$ [8].

A. Simulation Setup

For a voltage-mode transceiver the path loss in dB is given by $L = 20\log\left(\frac{V_{out}}{V_{in}}\right)$. This ratio is mainly influenced by the termination impedance at the receiver end, the return path capacitance, as well as the carrier frequency applied. In the ADS simulation system, we assume that there is no environmental noises since noises cannot be included at the source and the noise sources can only be used for the device noise analysis. In this study, we focus on the effect of return path capacitance, which is mainly influenced by the distance between the ground electrodes of transceivers and the receiver termination. Two copper plates with dimensions 25 mm × 25 mm × 3.5 mm are placed face to face in a 3 m × 4 m × 3 m room filled with air, where Ansys Maxwell simulation suite is used to find the return path capacitance.

The ground plane is placed on the bottom of the room, and the vertical distance between the copper plates and the ground...
Fig. 3. Simulated path loss values various: a) receiver termination with 1 pF $C_{\text{ret}}$, b) receiver termination with 20 pF $C_{\text{ret}}$ and c) return path capacitances plane is 1 m. The distance between the two copper plates are set from 0.2 mm to 700 mm. When the two ground electrodes are placed close to each other, the capacitance reaches as large as 17 nF, resulting in a low impedance at higher frequencies, and the transmitter and receiver can be regarded as common-ground. When the two ground electrodes are placed farther apart, the capacitance reduces. When the distance is 70 cm, the capacitance is as low as 1 pF, resulting in high impedance, even at high frequencies, leading to a poor return path. Based on this discussion, and to emulate real-life scenarios, the return path capacitance used is from 1 pF to 20 pF.

B. Simulation results

Firstly, we simulated the path loss under different receiver terminations ranging between 50 Ω and 1 MΩ. The carrier frequency is 21 MHz and two different return path capacitances are used: 1 pF and 20 pF. For the 20 pF case, as shown in Fig. 3(b), increasing receiver resistance decreases the path loss. When the receiver resistance exceeds 100 kΩ, the path loss saturates due to the high receiver impedance. For the 1 pF case, as shown in Fig. 3(a), the receiver resistance does not play an important role, instead, the return path capacitance plays the dominant role in the channel path loss.

Figs. 3(a) and 3(b) illustrate the channel path loss while sweeping the return path capacitance between 1 pF and 20 pF. Two receiver resistances are used, 50 Ω and 1 MΩ. When increasing the return path capacitance, the path loss reduces rapidly in both cases. The results clearly illustrate the effect of return path capacitance on the path loss. As the return path capacitance increases (impedance reduces), the path loss reduces, resulting in a higher signal-to-noise ratio (SNR). When the return path capacitance reaches 20 pF, the 50 Ω termination case has 4 dB higher path loss than that of 1 MΩ case. This is because a larger load can lead to more voltage applied at the receiver, yielding a lower path loss. Taking these simulation results as our reference, the next section introduces the system architecture, performance analysis, and application to ECG signal monitoring.

IV. SYSTEM DESIGN

A. System overview

Fig. 4 shows the system architecture. The transmitter is composed of an Atmega328PU microcontroller with an external oscillator, a modulator, and signal and ground electrodes. It receives data from the sensor and converts it to the universal asynchronous receiver transmitter (UART) output of the microcontroller. The UART data is transmitted through the human body, and is modulated by an on-off-keying (OOK) modulator to 21 MHz. The modulated signal is then connected to the signal electrodes and injected into the human body. The receiver is composed of a passive low pass filter (LPF), an envelope detector, a comparator, and electrodes. The received signal from the signal electrode is filtered with a low pass filter to remove the high frequency noise. Then, an envelope detector is used to demodulate the signal. The output of the envelope detector is compared with a signal level that is generated by lowpass filtering the envelope detector output so that the comparator recovers the transmitted bits. After the data is extracted by the comparator, it is transmitted to a laptop using a bluetooth low power module, where Matlab is used to display the signal. The following subsections introduce specifically how these blocks are constructed and how the functions are realized.

B. Transmitter

Microcontroller: The microcontroller used in the transmitter is Atmega328PU, which is a 8-bit high-performance, low power system (~7.5 mA), with external clock speed of 16 MHz. It has two UART interfaces, one inter-integrated circuit (I²C) interface, a master-slave Serial Peripheral Interface (SPI), and 8-channel 10-bit analog-to-digital converters (ADC).

Modulator: The modulator is composed of an oscillator SG8018CE, with 21 MHz oscillation frequency according to the IEEE 802.15.6 standard and a single-channel low-power AND gate SN74LV1G08 (~10µA) for OOK modulation with a ~24 mA output drive at 3.3 V. A low-dropout (LDO) voltage
regulator TPS782 with 3.3 V output voltage is used to provide the power for the oscillator.

C. Receiver

The receiver is composed of an envelope detector and a comparator for OOK demodulation. For the envelope detector, the low-cost and low-power (~17 mW) RF detector MAX9933 is used with a input resistance of 30 MΩ and input capacitance of 0.5 pF, where logarithmic amplifiers are designed to detect RF power levels. There is a 40 dB fixed gain and a logarithmic amplifier in this chip. It has a dynamic range of 45 dB, an operation frequency from 2 MHz to 1.6 GHz, and an input signal range from -58 dBV to -13 dBV. A 150 pF capacitor is used at the CLPF for the purpose of lowpass filtering.

The comparator is an MAX9031 chip with a current consumption of ~35 µA with 4 mV comparator hysteresis. The maximum output frequency of the comparator is 2 MHz. The positive input is connected to the output of the envelope detector while the negative input is connected to the high-pass filtered enveloped detector output, whose level is at the middle of the envelope output signal. Then the output of the comparator is the transistor-transistor-logic (TTL) signal with logic 0s and 1s. These bits are then transmitted to the computer with a Bluetooth module for analysis.

V. NUMERICAL RESULTS

This section introduces the experimental results of the HBC prototype, including the signal transfer illustration, path loss, power efficiency, and the prototype used in ECG monitoring.

A. Path Loss

Fig. 5 shows the signal transfer of the HBC system. Signals at four points are observed: a) AD8232 UART output, b) output of the SN74LVC1G08 AND gate, c) output of the MAX9393 envelope detector and d) output of the MAX9331 voltage comparator. A 18650 battery with nominal 3.7V voltage is used. The specific battery used in this experiment was measured at 4V.

The path loss of the capacitive HBC system is mainly influenced by the return path capacitance. According to the definition of the capacitance, the return path capacitance is mainly determined by distance between the ground electrodes of the transmitter and receiver. In the path loss measurement, two different scenarios were considered: a) both devices are placed on a table, and b) both devices were worn on the arm. For both scenarios, the distance between the two electrodes was incremented as follows: 5 cm, 10 cm, 15 cm, 20 cm, and 25 cm, as shown in Fig. 6(a). The experiments were implemented in the lab environment where there are metal furniture and shelves around. Note that in the table scenario, the BCC channel length does not change, rather the return path is incremented. The table is made of wood with alloy frames. The experimental results are shown in Fig. 6(b). At the same distance, the path loss when wearing the devices on the body is larger than that observed when the devices are placed on a table because the latter has a return path of lower impedance. Simulations and experimental results agree closer when the electrodes are placed on the human body. For the table scenario, simulation and experimental results diverge more showing the significant effect that the environment has on return capacitance and overall path loss, where a difference of up to 5 dB between simulation results and measurements were observed. For the on body scenario, Fig. 6(d) depicts the BER vs SNR for both theoretical based OOK and BER measurements on the prototype showing a close fit with minimal divergence occurring towards higher SNR values. Finally, Fig. 6(c) shows the transmitted and received ECG waveform using an AD8232 ECG sensor at the transmitter side.

B. Power efficiency

To quantify the energy efficiency of the communications sub-system of the proposed prototype, the total current draw for both TX and RX were measured during active transmission. It is important to note that this current supplies both the micro-controller and the communication subsystem.

The highest data rate that could be supported on the HBC system was found to be 1.3 Mbps. When operated at 1.3 Mbps, the TX circuit consumes a current of 12.14 mA in total including the micro-controller, sensor and communication sub-system. Per the data sheet, the micro-controller draws a typical current of 7.5 mA when clocked at 16 MHz at a supply voltage of 4.5 V. Based on this, once can conclude that excluding

<table>
<thead>
<tr>
<th>Reference</th>
<th>[1]</th>
<th>[2]</th>
<th>[8]</th>
<th>[7]</th>
<th>Proposed</th>
</tr>
</thead>
<tbody>
<tr>
<td>Frequency (kHz)</td>
<td>100~1000</td>
<td>1000~6000</td>
<td>30000</td>
<td>Baseline</td>
<td>21 kHz</td>
</tr>
<tr>
<td>Total power (mW)</td>
<td>151</td>
<td>4.8</td>
<td>–</td>
<td>Around 5</td>
<td>67.8</td>
</tr>
<tr>
<td>Transmit power (mW)</td>
<td>–</td>
<td>–</td>
<td>–</td>
<td>–</td>
<td>17.1</td>
</tr>
<tr>
<td>Data rate (kbps)</td>
<td>15</td>
<td>1250</td>
<td>468</td>
<td>9.6</td>
<td>1 Mbps</td>
</tr>
<tr>
<td>BER</td>
<td>–</td>
<td>–</td>
<td>–</td>
<td>–</td>
<td>Minimum</td>
</tr>
<tr>
<td>Device</td>
<td>Chips</td>
<td>Chips</td>
<td>FPGA kit</td>
<td>TM4C123G</td>
<td>Chip</td>
</tr>
<tr>
<td>Application</td>
<td>ECG</td>
<td>ECG</td>
<td>ECG</td>
<td>ECG</td>
<td>Pulse</td>
</tr>
</tbody>
</table>

Fig. 5. Signal transfer illustration (a) shows the components in the transmitter and the receiver and the experiment setup. (b) shows the waveforms after each stage. The yellow wave shows the UART output of the AD8232 sensor, the green one shows the modulated signal with 21 MHz carrier wave, the blue one shows the output of the envelope detector, and the purple one shows the output of the comparator with Vcc amplitude. (c) shows the signal transfer latency is 300 ns.
the micro-controller, and neglecting the current drawn by the sensor, the communication subsystem consumes 4.64 mA. Thus the performance of the communication subsystem shows a 10X reduction in power when compared to the HC-06 Bluetooth module with a typical current draw of 30-40 mA. According to the guideline from International Commission on Non-Ionizing Radiation Protection (ICNIRP), the threshold of contact current is 10 mA for children and 20 mA for adults and this current of the communication subsystem is far lower than the threshold. Moreover, the RX circuit was measured at 6.2 mA current draw. Thus the total current draw of the communication sub-system is 10.84 mA. Thus for the maximum data rate of 1.3 Mbps, the minimum energy efficiency can be calculated as 8.3 nJ/bit. This potentially sheds light on design of ultra-low power, low cost and small-sized sensors by lowering the output current of its backend transceiver, or ultra-low power multi-parameter healthcare monitoring system.

VI. CONCLUSION

In this paper, a wearable healthcare monitoring system based on human body communication is proposed using COTS components. The system shows 10× reduction as compared to a Bluetooth transmitter such as HC-06, with an energy efficiency of 8.3 nJ/b. This work can be potentially extended to other vital sign monitoring applications, such as EMG, EOG, etc., and sensors with all kinds of serial ports, including SPI, I2C, and UART.

ACKNOWLEDGEMENT

This work has been funded by King Abdullah University of Science and Technology (KAUST) and the Smart Health Initiative (SHI) at KAUST. The authors gratefully acknowledge Yuxiang Liu for contributions to system design and implementation.

REFERENCES