Free Battery–based Energy Harvesting Techniques for Medical Devices

Mustafa F. Mahmood, Saleem Lateef Mohammed, and Sadik Kamel Gharghan

Department of Medical Instrumentation Techniques Engineering, Electrical Engineering Technical College, Middle Technical University, Baghdad, Iraq
musiraq86@gmail.com(M.F.M.), saleem_lateef_mohammed@mtu.edu.iq(S.L.M), sadik.gharghan@mtu.edu.iq (S.K.G.)

Abstract. Many wearables or portable medical devices have limited battery energy. Such batteries cannot operate for a long time and require recharging or periodic replacement. A piezoelectric transducer (PZT), ultrasonic sensor (USS), and magnetic resonator coupling (MRC) are potential technologies for solving this problem, being promising technologies that can be used to generate free power for low-power medical applications. The USS and MRC optimize transfer power, efficiency, and distance between the transmitter and receiver. These three technologies can generate power to wearable and implantable medical devices (IMDs). To validate the proposed PZT, USS, and MRC, we supplied electromyograph (EMG) sensor, a heart rate sensor, and oxygen saturation (SpO2) sensor with adequate power to measure the subject’s muscle activity, heart rate (beats per minute, bpm), and SpO2 rate, respectively. The proposed system consists of four parts: power system, measurement part, wireless transmitter, and monitoring part. We found that 5 V could be used for charging 0.25, 0.33, 0.5, and 1 Farad supercapacitors based on the PZT at duration. Furthermore, the 0.25 F supercapacitor was fully charged in 41 min; compared with previous closed-circuit studies, it achieved high power of 197 μW at resistive load 15 kΩ. In addition, USS-based transfer efficiencies and powers could be used with 1, 4, and 8 F supercapacitors. The system had transfer efficiency and power of 69.4% and 0.318 mW, respectively, at 4 cm when 4 F was adopted. Furthermore, the MRC system had transfer efficiency and power of 21.14% and 2.079 W, respectively, at 7 cm at resistive load 70 Ω. Our results show that the PZT, MRC, and USS in the present study outperformed previous works in terms of power generation, transfer power, and efficiency.

Keywords: EMG, free battery, heart rate, magnetic resonator coupling, piezoelectric transducer, ultrasonic sensor

1. Introduction

Modern-day technologies provide the global community with a sustainable environment and a future without power interruptions. Currently, there are many resources for producing free power, such as biomass, wind, radiofrequency, solar, hydro, vibration, and heat energy. Studies have highlighted several methods of power production, optimization transfer power, efficiency, and distance between transmitter and receiver sensors for operating implantable biomedical systems and medical sensors, such as electromyography (EMG), blood pressure observers, pacemakers (PMs), thermometers, heart rate sensors, and neural stimulators, which are frequently employed to enhance the lifestyle of millions of
the ill people. Researchers have examined yield methods based on the following harvesting techniques. Some have studied far-field or near-field energy transmission. Near-field wireless power transfer (WPT) includes inductive coupling (IC), capacitive coupling (CC), magnetic resonator coupling (MRC), and ultrasonic transducers. Radiofrequency uses far-field transfer power. In addition, other research has used vibration based on the piezoelectric transducer. Besides, other research has used vibration based on the piezoelectric transducer.

In [1], a CC method was used to optimize transfer power for modeling tissue loss. Several studies have used an IC technique to optimize transfer distance, power, and efficiency, such as in [2,3,4,5], for different applications such as electrocardiograms (ECG), PMs, retinal prostheses, and others. In addition, MRC has been used for optimizing transfer distances and efficiencies [6,7,8,9], involving special components between experimental settings and simulations, which each had its respective advantages and disadvantages for applications such as capsule endoscopy, left ventricular assist devices (LVAD), PMs, implantable cardioverter defibrillators, and others. This will be discussed in Section 2.

To develop implant devices, many examiners [10, 11] were employed ultrasonic signals to improve transfer power and efficiency and to improve transfer air gap in “percutaneous coronary intervention (PCI)”. In this context, the authors in research work [12], the operating frequency of the system was modified to improve transfer distances between transmitter and receiver sensors. In [13], microelectronics components were used to improve transfer efficiency and compare them with WPT relay on IC technique.

Ultrasonic signal–based transfer energy is a different technique for active wearable biomedical transducers relative to the WPT method. Supplementary components were added to biomedical implant system, such as matching layers, to optimize transfer power and efficiency [14]. Several investigators [15] have used piezoelectricity to generate power in wearable biomedical devices and implantable devices such as heart rate monitors. Additional components, such as MOSFET (metal-oxide-semiconductor field-effect transistor) bridge rectifiers, have been used in medical implant devices to improve power [16]. In [17], a special-design piezoelectric transducer such as a piezoelectric polymer polyvinylidene fluoride (PPF)–trifluoroethylene, was used to optimize power. One study used a storage capacitor to operate a device [18]. Others have used program simulations to generate power for open and closed circuits [19,20,21,22,23,24].

The present study involved three designs; a piezoelectric transducer (PZT), ultrasonic sensor (USS), and MRC. These systems will be presented in Section 3.

The main aim was optimizing the transfer distance, transfer efficiency, and transfer power. The outcome of this research work can be figured out as follow:

1. Prototype of PZT, USS, and MRC were designed and implemented.
2. The power performance of the proposed systems (i.e., PZT, USS, and MRC) was achieved based on three low power sensors (i.e., heart rate, EMG, and SpO2).
3. The results of the three systems were identified and compared, including transfer distance and DC output voltage.
4. The performance metrics of three systems were validated relative to previous articles, which contained transfer efficiency and power.

2. Related works

This section investigates previous studies using CC, IC, MRC, piezoelectric sensors, radiofrequency, and a USS for generating power for patient vital sign monitoring systems. Jegadeesan et al. [1] carried out a wireless power transfer (WPT) using CC for a biomedical implanted device for modeling tissue losses operating at 402 MHz. The system contained a transmitter plate and receiver plate. Three skin thicknesses were tested. The experimental results showed that transfer power efficiencies differed according to the skin thickness of 0.3, 0.4, and 0.5 cm. Transfer efficiencies of 68.3%, 67.2%, and 67% were achieved at inter-plate distances of 0.46, 0.56, and 0.66 cm, respectively. Le et al. [2] proposed a WPT using IC to monitor firefighters’ survival using ECG. The system consisted of an ECG sensor and Bluetooth data communication. The operating power involved IC through a couples of twin planar spiral antennas installed in the external unit and the base layer. The selected operating frequency was approximately 1 MHz. The maximum power transfer of the device for non-contact electrode (NCE)
ECG was 0.28 W obtained at 1 cm. However, the system could be used at a maximum transfer distance of 4 cm. Das et al. [3] implemented a WPT using IC for optimizing transfer efficiency of a PM using a Yagi-Uda antenna. The system consisted of a Yagi-Uda antenna, Tx, Rx, the Remcom XFtdl program, and a PM device. The experimental results showed that a different part of the Yagi-Uda antenna component could be used. Maximum transfer efficiency of 68% was achieved based on MTMs for all parts. The maximum transfer efficiency was 43% and 57% based on MTMs as a reflector and as a director, respectively. However, a maximum transfer efficiency of 35% could be attained without MTMs.

Kyungmin et al. [4] designed a WPT using MRC for biomedical capsule endoscopy operating at 16.47 MHz oscillation frequency. The system consisted of the vision, control, data transmission components, and the MRC circuit. The MRC consisted of receiver, transmitter, and load. The realistic results showed that the transfer power efficiency was improved by 0.71% inside the patients at a distance of 7 cm. The transfer efficiency was investigated relative to three parameters: distance, angle, and axial misalignment between coils. Monti et al. [5] presented a WPT using MRC to supply implantable medical devices (IMDs) such as a PM with power at 403 MHz. The system consisted of a Tx, Rx, rectifier, and PM device. The experimental results revealed that the PM power consumption was 0.01 to 1 mW. The maximum transfer efficiency of 5.24% was accomplished between the Tx and Rx at an airgap of 10 cm. However, it was large. Campi et al. [6] implemented a WPT using MRC for IMDs such as LVADs, PMs, and implantable “cardioverter defibrillators” at 300 kHz. The system consisted of a Tx, Rx, and load. The practical results exposed that the transferred power in the Rx was 1 W. The transfer efficiency was almost 47% at 3 cm, whereas it was about 7% at 6 cm. Campi et al. [7] tested a WPT using MRC for an implantable medical PM device for battery recharging at 300 kHz and 13.56 MHz. The systems consisted of a source coil, load coil, battery, and load. Two configurations were tested. The experimental results revealed that series/primary (SP) coil, series/secondary (SS) coil, a series/primary (SP) coil, and parallel/secondary (PS) coil had greater transfer efficiencies at 13.56 MHz than at 300 kHz; the comparison was based on the number of turns. The transfer efficiency of the PS configuration was better at 300 kHz, i.e., about 31%, at 3 cm compared with that at 13.56 MHz, which was 27% at the same distance. The SS configuration had better transfer efficiency at 13.56 MHz, i.e., about 10%, at 5 cm compared with that at 300 kHz, which was 1% at the same distance. The output power of the secondary circuit was 1 W.

Islam et al. [8] presented a power system-based vibration sensor to enhance transfer power efficiency for implantable biomedical devices in PCI at 14 MHz. The proposed system comprises an ultrasonic transducer, PPF, autonomous active stent sensor, and rectifier circuit. The experimental outcomes indicated that the transfer efficiency rely on active stent sensor and PPF was around 14%, whereas transfer efficiency of 11.5% was accomplished with “lead zirconate titanate” (LZT). When low resonant frequency and short distance of 2.5 mm is adopted, the obtained voltage was 200 mV. Comparison the transfer power efficiency between PPF and ASS with LZT At the same distance (i.e. 2.5 mm) revealed that the system employing PPF and ASS was superior the LZT system. Shi et al. [9] improved transfer efficiency and power of a vibration-based piezoelectric sensor for medical implant devices. The operating frequency of the system was 240–250 kHz. The platform consists of load and ultrasonic receiver and transmitter. Based on the piezoelectric ultrasonic transducer (PU), the DC output power was developed from 12 to 77 nW at an airgap of 10 mm. The realistic results disclosed that the obtained output power was decreased as the distance increased and vice versa.

Vihvelin et al. [10] vibration-based piezoelectric sensor for MIDs were highlighted to increase transmission efficiency. Thereby, the MID was supplied by energy depending on the energy scavenging method (i.e. ultrasonic transducers) at operating frequency between 1.2–1.4 MHz. The system comprises ultrasonic receiver, ultrasonic transmitter, and load. In this research, the biomedical device was tested in two kinds of medium. The practical outcomes disclosed that the transmission distance and efficiency between receiver and transmitter varied based on medium and frequency. In water media, the transfer efficiency was observed of 45% at distance of 6.1 and 5.9 mm at 1.29 and 1.35 MHz, respectively. Whereas, the efficiency was 48% at airgap media of 4 mm in the frequency range of 1–1.6 MHz. The experimental results demonstrated that the deliver efficiency was changing between 35–100% for varying frequencies, while it was in the range of 8–25% for constant frequencies. Meng et al. [11]
optimized transfer power efficiency of ultrasound sensor for implantable biomedical device. Energy harvesting technique based on vibration-based piezoelectric sensor used to provide the biomedical devices by power at different frequencies such as 1, 10, and 15 MHz. The system includes an ultrasonic receiver and transmitter, rectifier, regulator, and load. The above frequencies were practically implemented. The results illustrated that the transmission efficiencies were 0.03%, 0.15%, and 6% at 3 cm and 0.043%, 0.06%, and 5% at 8 cm for 10, 15, 1 MHz, respectively. The authors deduced that the transfer efficiency for inductive method was outperformed the ultrasonic method at 3 cm, where the transfer efficiency was 0.27% (inductive method) and 0.03% (ultrasonic method). However, the transfer efficiency was deteriorated to 0.0025% for inductive method compare with ultrasonic method (0.04%) when transfer distance increased to 8 cm. Miao et al. [12] vibration-based piezoelectric sensor for the implantable microelectronic device were proposed. The aim of the work was to optimize the transfer efficiency at different frequency operation (i.e., 1.1–1.8 MHz) of ultrasonic transducer. The system contains a piezoelectric receiver and transmitter, regulator, rectifier, and load.

Ansari et al. [13] presented a piezoelectric sensor for wearable microelectronic devices such as heart rates at less than 20 Hz. The system includes a piezo-element, supercapacitor, and load. Two kinds of piezoelectric sensor were utilized in their real experiment. Experimental results revealed that using a small-scale of piezoelectric is adequate to generate a power source for PM device. Taeho et al. [14] implemented a vibration in wearable sensors to produce a voltage based on the piezoelectric element. The system includes the piezoelectric, rectifier, regulator, and load. Experimental results revealed that the output power was 10.7 μW by using MOSFETs bridge rectifier involves of two p-channel and two n-channel. The result showed an output voltage with input voltage was 694 mV and 703 mV at resistive load 45 kΩ, respectively. Toprak et al. [15] designed a piezoelectric transducer-based on vibration to generate power in a wearable sensor at 1.74 kHz. The system includes piezoelectric energy harvesters using piezoelectric polymer polyvinylidene fluoride–trifluoroethylene, buffer circuit, and load. Experimental results disclosed that the power and DC output voltage were 35pW and 33.3 mV at 4.3 MΩ with high displacement, respectively. Zhang et al. [16] presented a vibration-based on harvesting technique piezoelectric material sensor for wearable devices at 100–170 Hz. The system includes the piezoelectric material, rectifier, storage capacitor, and load. Experimental results revealed that the output power was 2.22 μW at 160 Hz with acceleration 10.5 m/s². Janusas et al. [17] presented wearable devices by using vibrational-based on piezoelectric material to generate power at 50 Hz. The system includes the piezoelectric sensor and load. Experimental results showed that the DC output voltage was up to 80 μV. Saadon et al. [18] designed and implemented a vibration based on a piezoelectric plate to generate power for different applications such as wearable sensor at 67–70 Hz. The system used Coventorware simulation program for studied an output power and voltage at acceleration. The practical results demonstrated that the DC output voltage and output power were 0.4 V and 6.8 μW with acceleration 0.2–1.3 g at 20 kΩ, respectively.

Safaie et al. [19] implanted a piezoelectric transducer-based on vibration in knee implants to generate power to wearable device at 25 Hz. The system used three program OpenSim modeling software, ANSYS software, and MATLAB software for simulation a knee, design a piezoelectric transducer, and calculate power and voltage. Experimental results revealed that output power and voltage was 12 μW and 2.3 V at load 1MΩ. Pillatsch et al. [20] presented a vibration by using piezoelectric to generate power for wearable devices at 0.5–4 Hz. The system contains piezoelectric and load. Practical results deduced that high output power was obtained at high acceleration. The output power was 0.02 mW at 2 Hz for acceleration 20 m/s². The maximum output voltage at 150 kΩ loads was 12 V. Ghosh et al. [21] presented a vibration-based piezoelectric element to generate power for wearable devices at 0.1–1000 kHz. The system contains piezoelectric, rectifier, and load. The results showed that the output voltage for the closed-circuit is 4 V with 1.5 μA, the maximum power was 1.14μW. The output power at 13 MΩ was 0.47μW. Yu et al. [22] presented a vibration-based on harvesting technique piezoelectric material sensor to generate power for wearable devices at rang frequencies 180–230 Hz. The system includes piezoelectric material and load. Experimental results revealed that the output power was 1.78 μW at 210 Hz with acceleration 0.6 m/s² by using COMSOL program. The output power was increased with acceleration.
The aforementioned scholars were briefed in Table 1. This table presents the comparison of the performance metrics of these articles including the DC output power, transfer efficiency, transfer distance, type of harvesting technique, operating frequency, voltage for each research work. In addition, the implementation environments, objective of each study, and application in medical field were also highlighted.

| Ref | Objective | Operating Frequency (MHz) | Harvesting Technique | Implementation environment | Application | Transfer distance (cm) | Transfer efficiency (%) | Voltage (V) | Power (W) |
|-----|-----------|--------------------------|----------------------|----------------------------|-------------|------------------------|------------------------|------------|----------|
| [1] | Optimize transfer efficiency at distance | 402 | CC | Experimental | MTL | 0.66 | 67 | --- | --- |
| [2] | Optimize the power | 1 | IC | Experimental | ECG | 4 | --- | --- | 0.28 |
| [3] | Optimize transfer efficiency | 402-405 | IC | Simulation | PM | --- | 68 | --- | 0.001 |
| [4] | Optimization transfer efficiency | 16.47 | MRC | Experimental | CE | 7 | 0.71 | --- | --- |
| [5] | Optimize transfer efficiency at distance | 403 | MRC | Experimental | PM | 10 | 5.24 | --- | 10×10^6 |
| [6] | Optimize transfer efficiency at distance | 0.003 | MRC | Experimental | ICD | 6 | 7 | --- | 1 |
| [7] | Battery-recharge | 0.003–13.56 | MRC | Experimental | PM | 5 | 10 | --- | 1 |
| [8] | Optimize transfer efficiency at distance | 14 | Vibration | Experimental | PCI | 3 | 14.8 | --- | --- |
| [9] | Optimize transfer power | 0.24–0.25 | Vibration | Experimental | ID | 1 | --- | --- | 7.71×10^6 |
| [10] | Improve transfer efficiency | 1.2–1.4 | Vibration | Experimental | ID | 0.61 | 45 | --- | --- |
| [11] | Improve transfer efficiency | 1, 10, 15 | Vibration | Experimental | ID | 4 | 5.5 @ 1 cm, 0.035 @ 10 cm, 0.087 @ 15 cm | --- | --- |
| [12] | Improve transfer efficiency | 1.1–1.8 | Vibration | Experimental | ID | 4 | 1.14 | --- | --- |
| [13] | Generate power | 0.02 | PZT | Experimental | HRs | --- | --- | --- | 16×10^6 |
| [14] | Generate power | 0.2 | PZT | Experimental | WS | --- | --- | 0.694 | 10.7×10^6 |
| [15] | Generate voltage | 1.074 | PZT | Experimental | WD | --- | --- | 0.033 | 0.703 |
| [16] | Generate power | 0.160 | PZT | Experimental | WD | --- | --- | --- | 0.018×10^6 |
| [17] | Generate voltage | 0.05 | PZT | Experimental | WD | --- | --- | 8×10^4 | --- |
| [18] | Generate voltage | 0.067 | PZT | Simulation | WD | --- | --- | 0.45 | 6.8×10^6 |
| [19] | Generate voltage | 0.025 | PZT | Simulation | WD | --- | --- | 2.3 | 12×10^6 |
| [20] | Generate power | 0.002 | PZT | Simulation | WD | --- | --- | 12 | 20×10^6 |
| [21] | Generate power | 0.1–1000 | PZT | Simulation | WD | --- | --- | 4 | 0.47x10^6 |
| [22] | Generate power | 0.21 | PZT | Simulation | WD | --- | --- | 1.78×10^6 | --- |

Modelling tissue losses (MTL); Pacemaker (PM); Retinal prostheses (RP); Capsule endoscopy (CE); Implantable cardioverter defibrillators (ICD); Percutaneous coronary intervention (PCI); Heart rates (HRs); Wearable sensor (WS); WD - Wearable devices; ID - Implant devices; P - program

3. System models

The proposed power systems related to PZT, USS, and MRC for running medical sensors were presented in this section as in the following subsections. The powers, measurements, and monitors units for three systems also were described in this section.

3.1. Piezoelectric transducer (PZT)

The suggested PZT consists of three units: power, measurement, and monitoring as shown in Figure 1. The power unit includes a PZT, bridge rectifier, storage unit, ON–OFF switch, DC-DC converter, and load (i.e., supercapacitor). The measurement unit consists of microcontroller-based Arduino Nano, MyoWare muscle sensor, and wireless communication module (i.e., nRF24L01). This radio frequency module is small, ultra–low power (ULP) consumption, cost-effective, and transmits data up to 100 meters in outdoor environment. The model size is 28.5×15.2 mm².
The monitoring components included microcontroller-based Arduino Mega 2560, nRF24L01 module, and a laptop supported by MakerPlot software to monitor the muscle activity data. In this design, the generated power of PZT was mostly depend on the subject’s or patient’s weight.

3.2. Ultrasonic sensor (USS)

The suggested USS consists of three parts: power, measurement, and monitoring as shown in Figure 2. The power unit includes ultrasonic receiver, DC-DC converter, bridge rectifier, and supercapacitor. The measurement part comprises of a microcontroller using Arduino Nano, nRF24L01 wireless module, and heart rate sensor. The monitoring part included a microcontroller-based Arduino Mega 2560, nRF24L01 module, and a laptop with the MakerPlot software to monitor the heart rate data. The USS operation was based on air gap between the transmitter and receiver sensors.
3.3. Magnetic resonator coupling (MRC)

The suggested MRC consists of three units: power, measurement, and monitoring as shown in Figure 3. The power unit included a transmitter comprising an oscillator, power supply, and transmitter coil ($T_x$). The receiver comprised a receiver coil ($R_x$), capacitor resonator, bridge rectifier, and regulator. The measurement unit consisted of microcontroller-based Arduino Nano, SpO2 sensor, and nRF24L01 module. The monitoring unit included microcontroller-based Arduino Mega 2560, nRF24L01 module, and a laptop supported by MakerPlot software to monitor the SpO2 data. The MRC operation was based on distance between the transmitter and receiver coils.

![Figure 3. Block diagram of using an MRC-based SpO2 sensor.](image)

3.4. Power unit

Power unit for PZT/USS systems uses piezoelectric transducers/ultrasonic sensor, bridge rectifier, DC-DC converter and bank power (supercapacitor). PZT converted mechanical signals to electrical signals; the signal has an alternate signal. USS in the transmitter side converts ultrasonic signals from electrical signals. A bridge rectifier converts AC to DC signal. The DC-DC converter uses DC boost model (based on MT3608 chip), which cost-effective, small, and high boosting efficiency of 93% [25]. Whereas power unit for MRC system uses an oscillator, transmitter coil, receiver coil, bridge rectifier, regulator, capacitor resonator and load.

3.5. Measurement units

The measurement units of the PZT, USS, and MRC all contained an Arduino Nano microcontroller, sensors, and wireless protocol nRF24L01. The Arduino Nano microcontroller is small, low-cost, lightweight, programmed using C++ language and targeted at low power consumption [26-28]. The PZT system is connected directly, based on the subject’s weight, to measure the subject activities of muscle. The USS and MRC systems are based on the airgap or distance between the sensors and coils, respectively. The measurement sensor in the PZT system is a MyoWare muscle sensor, an analog sensor that is low-cost and lightweight [29, 30]. The USS uses a heart rate sensor linked to low voltage for decreased power consumption, is small, i.e., 16-mm diameter, and low-cost [31-33]. The MRC uses a SpO2 sensor and is small, low-cost, and lightweight [34]. The nRF24L01 wireless protocol was adopted in three systems to send the measured data of the sensors to the monitoring unit. It is low-power consumption, low-cost, lightweight, and can send data over long distances of up to 100 meters [35, 36]. It uses 2.4 GHz radio frequency. It has many applications, such as wireless computer peripherals, wireless data communication, industrial sensors, and ULP sensor networks. In the present module, we
enabled control of media access, address, information, and error detections and retries for securing and receiving data [37, 38].

3.6. Monitoring units
The monitoring unit for all system includes a microcontroller-based Arduino Mega 2560, nRF24L01, and a laptop. The Arduino Mega 2560 is lightweight, low cost, and programmed using C++ language [39]. Wireless protocol nRF24L01 was employed to receive data from the measurement unit. It was described in the measurement part. The laptop supported by MakerPlot software was used to display the measured data of the different sensors on serial monitoring.

4. Experiment configuration
In this section, the experiments for the PZT, USS, and MRC have been described as follows.

4.1. Experiment of Piezoelectric Transducer
The piezoelectric ceramic transducer uses a 35mm diameter of copper [40]. The circuit diagram for the system shown in Figure 4a. First, a piezoelectric transducer is used 12 pieces. These are implanted up and down the plantillas; those are connected in parallel to increase a current value. The PZT convert a mechanical signal to an electrical signal at a low frequency about < 3 Hz based on the force applied it. Secondly, convert AC to DC so using bridge rectifier, it is using a Schottky diode (SR260) model [41] because the Schottky diode has low voltage drop about 0.18 V. The output voltage from the bridge is filtered by a capacitor about 100 nF at 50 V. Third, the supercapacitor was charged when the ON–OFF switch was closed. Forth, the DC-DC converter was used in this experiment to increase the DC output voltage where the produced voltage form piezoelectric transducers was low and inadequate to supply the biomedical device by power. In the current experiment, the DC-DC converter dimensions are 36 × 17 × 14 mm³ [25]. The DC-DC converter uses for increases voltage from 1.8 V to 4.2 V initially based on ON–OFF switch to charge supercapacitors. Supercapacitors value are used in experiment 0.25, 0.33, 0.5, and 1 Farad (F). The mechanism to charge a supercapacitor at a time based on Equations (1), (2), and (3). The resistor for charging a supercapacitor is 10 Ω and capacitance for capacitors depend on uses a supercapacitor [42].

\[
I_c = I_s [1-e^{-t/\tau}]
\]
\[
V_c = V_s [1-e^{-t/\tau}]
\]
\[
\tau = re
\]

where \(I_s\) and \(V_s\) are the current and voltage, respectively, at the source. \(I_c\) and \(V_c\) are the current and voltage, at a particular time and \(\tau\) is the time constant, it can be obtained by multiplying capacitor \(c\) times resistor \(r\).

Measurement units, Arduino Nano microcontroller have based a supercapacitor to operate a device. A supercapacitor is connected on voltage and ground pin. Hence, the Myoware muscle sensor and wireless protocol nRF24L01 are connected at Arduino Nano. The Myoware muscle sensor is connected on pins A0 and 3.3 V and GND of the microcontroller. The wireless protocol nRF24L01 is connecting on pin 3.3 V, GND, D7, D8, D13, D11, and D12 of the microcontroller. In addition, the chip-enabled activation of transmitter and receiver mode (CE) and SPI chip-select (SCN) of the nRF24L01 were connected to digital pins D7 and D8 of the microcontroller, respectively. In Figure 4b, shows a snapshot of the PZT-based system for generating adequate voltage to operate a wearable device and the wireless protocol for the soleus muscle, depending on the person’s weight.
4.2. Experiment of Ultrasonic Sensor

The USS system uses ultrasonic transmitter sensor (Tx) is convert electrical signal to ultrasound signal as shown in Figure 5a and b. The dimension of USS transmitter and receiver is 45×20×15 mm³. The input voltage to this transmitter is 10 Vpp square wave at 40 kHz operating frequency feed by function generator (VC2002 model). Ultrasonic receiver sensor (Rx) converts the ultrasound wave to an electrical square waveform. The transfer power of USS depend on airgap between transmitter and receiver. Next stage, bridge rectifier was used to convert AC to DC voltage by adopting the rectifier (LTC3588 model). The dimension is 2×1.25 cm². The receiver side receives frequency and voltage of the USS transmitter. Then fasten to bridge rectified by input element at (PZ1 and PZ2) and output voltage from element at (VIN). This voltage was low, therefore DC-DC converter was used to raise up voltage. The DC-DC converter (MT3608 model) [25] was used in this experiment has board dimension of 36×17×14 mm3, small size, and low cost. The measurement part includes the USS receiver provides the Arduino Nano by power-based capacitor 4F that connect on V_IN and GND pin. The microcontroller is supplying a wireless protocol and heart rate sensor at 3.3V, as explained previously in Figure 2. While the monitoring part is supplying by the power at USB adaptor from PC or laptop to supply Arduino Mega 2560 and nRF24L01 module. The experiment was designed and practically implemented to transfer power between ultrasonic transmitter (Tx) and receiver (Rx). 4 cm transfer distance was considered to charge supercapacitor 4F. Supercapacitor is required time for fully charging relate to Equations 1, 2 and 3. The ultrasonic receiver, Rx is detecting the transmitted signal of Tx at 4 cm. In this experiment, storage oscilloscope (UTD2025CL model) was used to display the Rx signal. Then, the AC signal is converted into DC by a bridge rectifier. Then DC-DC converter and apply on supercapacitor. Supercapacitor is supplying power to the microcontroller of the measurement part. Arduino Nano at data rate is 9,600 bps was selected so as for optimizing a transmission time between microcontroller and nRF24L01 wireless protocol. Moreover, Arduino mega 2560 in monitoring part has the same data rate. The wireless protocol nRF24L01 data rate is 250 kbps for transfer data between the two models. Finally, laptop using for observing a result in serial monitoring port by adopting Arduino IDE software version 1.8.8. While calculating the heart rate in beat rate per minute based on Equation (4) [43].

\[
BPM = \left(\frac{1}{\text{Signal}}\right) \times 60.0 \times \text{frequency}
\]  

(4)

where \(BPM\) is a beat per minute, \(\text{signal}\) is an input signal from A0 of the microcontroller, and \(\text{frequency}\) is a sampling frequency.
Figure 5. Photos of the ultrasonic transfer power (a) system under test using supercapacitor 4F at airgap 4 cm and (b) monitoring part to display the heart rate data.

4.3. Experiment of Magnetic Resonator Coupling

The MRC experiment shown in Figure 6a was used the ZVS oscillator (zero voltage switching) to convert AC from DC at a high frequency of about 13.6 kHz at 19 V. The Tx_c converts the electrical field to the magnetic field. The diameter of the transmitted coil is 18 cm at 21 American wire gauge (AWG) for 695 μH as shown in Figure 6b. The Rx_c converts the magnetic field to the electrical signal with a specific amplitude (depend of distance between transmitter and receiver coils) and frequency. The diameter is 12 cm at 21 AWG for 208 μH as shown in Figure 6c. The resonant frequency of the transmitter and receiver coils can be computed based on Equation (5). Table 2 summarizes the information on the coils.

\[
F = \frac{1}{2\pi\sqrt{LC}}
\]  

where \( F \) is resonant frequency, \( L \) is the inductance value, and \( C \) is the capacitor value for the coupling resonator.

Figure 6. Experiment of MRC (a) A snapshot of the MRC system at 7 cm, (b) transmitter coil, and (c) receiver coil.

Table 2. Specification for coils in the transmitter and receiver units.

| Parameters                      | Unit | Value for transmitter coil \((Tx_c)\) | Value for receiver coil \((Rx_c)\) |
|---------------------------------|------|--------------------------------------|----------------------------------|
| DC Input voltage                | V    | 29                                   | -----                            |
| AC out voltage                  | V    | 19                                   | 19                               |
| Operating frequency             | kHz  | 13.6                                 | 13.6                             |
| American wire gage (AWG)        | -----| 21                                   | 21                               |
| Inductance                      | μH   | 695                                  | 208                              |
| Number of turns                 | turns| 91                                   | 62                               |
| Diameter of turn                | cm   | 18                                   | 12                               |
| Compensating capacitor          | pF   | 300                                  | 300                              |
The MRC depends on electrical and Electromagnetic Field, shape of coils, and airgap between the Tx and Rx. The transmitter side of system includes of power supply, oscillation, and Tx. The receiver side includes an Rx, capacitor resonator, bridge rectifier, regulator, and load. The experimental setting distance was 7 cm for operating the measurement unit. In a receiver side, the Rx, detected a signal at frequency from Tx. In the next stage, the signal into a capacitor resonator to have a maximum transfer power between the coils at a distance about 300 pF. In addition, convert AC to DC utilizing a bridge rectifier circuit. Then, to a voltage regulator is modified to +5V using 7805 model. The output voltage from regulator is supplies power to the Arduino Nano microcontroller by connecting the regulator at the VIN pin and ground pins. We used a voltage regulator because at a short distance a voltage was high due to damage for electrical component for instance Arduino. Simultaneously, a sensor and the radio communication part (i.e., nRF24L01) are provided by 3.3 V voltage as seen in Figure 3. The nRF24L01 using to transfer data between measurement and monitoring parts. A SpO2 sensor is joined at low voltage (3.3 V) for decreased current consumption. In monitoring unit, Power is supplied to the Arduino Mega and wireless protocol via a USB cable from the laptop show in Figure 6a. The Arduino Nano microcontroller and radio frequency nRF24L01 model, configured at a bit data rate of 9,600 bps, were selected for improving the processing time between the microcontroller and wireless communication model. Furthermore, the monitoring part, using a microcontroller-based Arduino Mega 2560, has the identical bit data rate as the microcontroller in monitoring part. The wireless nRF24L01 model has data transmission rate is 250 kbps for transferring data between two Arduino in the measurement and monitoring units. In the transmitter model, the microcontroller is connected at an input voltage of 3.3 V. The monitoring unit, Arduino Mega 2560 microcontroller is based on the USB cable to supply power from the laptop. Hence, it supplies a wireless protocol nRF24L01 to receive data from measurement units. In addition, uses a MakerPlot program to display a result.

5. Results and Discussion

5.1. Charging supercapacitors, supplying power, and EMG signal based on PZT

In the present experiment, used four different supercapacitors with charging at different times. The different capacitor charging categories are 0.25, 0.33, 0.5, and 1 F at different times. The chart is acting full charging supercapacitors at a time in min. It is shown a time sharply increase at large value of capacity, show Figure 7a. The solid blue line indicates the charging time at 0.25 F at 40.75 min, the dashed orange line indicates that for 0.33 F at 61.42 min, the dash-dotted green line indicates that for 0.5 F at 148.96 min, and the long dash-dotted black line indicates that for 1 F at 295.26 min, respectively based on Equations (1), (2), and (3). We performed an experiment for calculating an output power at different loads based on PZT. The bar chart is acting a resistance load in (kΩ) at power in (μW). The maximum power is 197 μW at 15 kΩ, whereas the minimum power is 18 μW and 27 μW at 330 kΩ and 1.5 MΩ, respectively, we chose yield load between 0.33–1500 kΩ show in Figure 7b. The measurement part measures a muscle activity and sends this result via nRF24L01. A sensor is position in the soleus muscle. The monitoring part receives data via nRF24L01 show in Figure 7c.
5.2. Charging supercapacitors and transferring powers based on USS

The charging yield categories of capacitors are 1, 4, and 8 F for five voltage at diverse transfer airgap, in Figure 9. Using 3 different supercapacitors are charging at air gap between sensors. The x-axis represents distances in centimeter and the y-axis represents time in a minute, in Figure 8a. Full charging time at 5 cm is 273, 1128 and 3303 min for 1, 4 and 8 F, respectively. Whereas, the charging time is gradually increased to 1813, 3078 and 21419 min at 10 cm. The resistor value required to charge the capacitor is 10 Ω. Where the current (Equation 1) and voltage (Equation 2) values depend on charging time which in turns rely on the capacitor and resistor values. Figure 8b presents the output power plotted on the y-axis and the transfer distance on the x-axis. The output power were measured for different distances for USS at different loads. The output power was 0.56, 0.57 and 0.21 mW at 2 cm. Whereas, it was gradually decreasing to 0.012, 0.07 and 0.034 mW for 1, 1.2 and 15 KΩ, at 10 cm, respectively. In Figure 8c, the transfer efficiency were measured for several distances at different loads. The transfer efficiency were plotted on the y-axis and the transfer distance were plotted on the x-axis. The transfer efficiency was 78.03%, 73.16%, and 58.54% at 2 cm. Whereas, it was gradually decreasing at high 2 cm, at 10 cm 39.39%, 36.8%, and 38.91% at 1, 1.2 and 15 KΩ, respectively.

5.3. Signal wave and transfer power efficiency based on MRC

When the experiment configuration was completed, the output voltage signal of the transmitter and the receiver with the frequency spectrum was observed using storage oscilloscope (UTD2025CL). The output voltage signal of the Tx was a sinusoidal signal (Figure 9a). However, the generated frequency of the oscillator circuit was $13.6$ kHz. At the receiver circuit, Rx the signal was also sinusoidal signal
with single pure tone, but its amplitude lower than the amplitude of the transmitted signal (Figure 9b). Figure 10b disclosed that the detected signal by the Rx, had the same frequency of the transmitted signal (i.e., 13.6 kHz) at 7 cm airgap between transmitter and receiver coils. Figures 9a and 9b show the frequency spectrum (red shape) of the transmitter and receiver circuit as a single tone without noise or interference signals. At the receiver circuit, the signal was adapted from AC to DC voltage based on bridge rectifier. Consequently, the DC output voltage of 3.5 was noticed at 7 cm with load systems shown in Figure 9c. The y-axis indicates the amplitude of the signal and x-axis demonstrates the time.

Figure 9. The voltage waveforms based on MRC at 7 cm for (a) transmitter coil, (b) receiver coil, (c) voltage load.

Figure 10a shows the transfer power at distance for different loads. The bar chart shows the relationship between distance (cm) and transfer power (W). The transfer power was 1.386, 1.71, and 2.08 W at 7 cm. However, it gradually decreased at distances over 7 cm. The transfer power was 0.075, 0.0476, and 0.084 W for 40, 50, and 70 Ω at 15 cm, respectively. Figure 10b shows the transfer efficiency at distance for different loads. The bar chart shows the relationship between distance (cm) and transfer efficiency (%). The transfer efficiencies were 20%, 20.5%, and 21% at 7 cm; however, it gradually decreased at distances over 7 cm. At 15 cm, the transfer efficiency was 9.19%, 8.74%, and 6.27% for 40, 50, and 70 Ω, respectively. The transfer efficiency was calculated using Equation 6 [8].

\[
\text{Transfer efficiency} \ (\eta) = \frac{\text{output power}}{\text{input power}} \times 100\%
\]  

Figure 10. The MRC performance at different transfer distance loads (a) transfer power and (b) transfer efficiency.

6. Comparison results

Table 3 summarizes the present work, comparing the performance metrics of the present study involving PZT, USS, and MRC in terms of system transfer distance, voltage before/after DC boost, size, cost, wearable device, power generation, and supercapacitor. The table shows that the three systems share the three parameters of size, cost, and wearable device. The PZT system generated power without requiring an external source, whereas the USS and MRC required external sources such as a function generator and a power supply. The USS and PZT required a supercapacitor to operate the device, whereas the MRC did not because it operated directly. The PZT was connected directly with the circuit, so there was no transfer distance, whereas the USS and MRC had a transfer distance. The USS and MRC system
transferred power at a transfer distance between the receiver and transmitter units. Hence, the wire-free operation is preferable and does require external force as the PZT would. The MRC system did not require a DC boost, whereas the PZT and USS did. The MRC could be used at a long distance, whereas the USS could be used at a short distance.

Table 3. Comparison of results of PZT, USS, and MRC.

| Parameter         | PZT system | USS system | MRC system |
|-------------------|------------|------------|------------|
| Transfer distance | ------------ | @10 cm     | @25 cm     |
| Voltage before DC Boost | 1.6 V      | 1.8 V@4 cm | ------------ |
| DC output voltage | 4.7 V      | 4.2 V@4 cm | 3.5 V @ 7 cm |
| Size              | Small      | Small      | Small      |
| Cost              | Low        | Low        | Low        |
| Wearable device   | Yes        | Yes        | Yes        |
| Power generation  | 44 µW @ 1 kΩ | No     | No         |
|                   | 196 µW @ 15 kΩ |        |            |
|                   | 116 µW @ 100 kΩ |        |            |
| Supercapacitor    | Yes        | Yes        | No         |

7. Comparison with previous works

Figure 11a shows the power (µW) based on the PZT system in the present and previous studies. In [13, 20], the power was 16 and 20 µW, respectively, whereas it was 196 µW in the current work. Figure 11b demonstrates the transfer efficiency (%) of the USS technique in the previous and current studies. In [11, 12], the transfer efficiency was 14.8%, 45%, 6% and 1.14%, respectively, it was 69% at 4 cm in the present study. Figure 11c displays the transfer efficiency (%) based on the MRC system in the current and previous studies. In [6, 7], the transfer efficiency was 7% and 10% at 6 and 5 cm, respectively, whereas the transfer efficiency in [5] was 5.24% at 10 cm. The transfer efficiency in the present study was 21% compared with that of [4], which was 0.71% at 7 cm. Hence, the present study has superior power generation, transfer efficiency, and transfer the power of the PZT, USS, and MRC systems as compared to that of the previous studies.

8. Conclusions

Here, we introduce the design and implementation of the PZT, USS, and MRC approaches for supplying an EMG sensor, heart rate sensor, and SpO2 sensor, respectively. The PZT system was tested by charging 0.25, 0.33, 0.5, and 1 F supercapacitors. We evaluated several supercapacitors to identify the best capacitor for yielding a fast charge time with good results compared with the other capacitor values. At 0.25 F, the PZT system was charged in 41 min. In addition, high power was generated at about 197 µW. At different supercapacitor values, the capacitor charging time increased with supercapacitor capacity. Therefore, we recommend using the proposed low value, i.e., the 0.25 F supercapacitor, to obtain adequate voltage quickly. The USS system was tested by charging 8, 4, and 1 F supercapacitors. At 4 F, the USS system has accomplished transfer power and efficiency of 0.318 mW and 69.4%,
respectively, at a 4-cm air gap between sensors. Furthermore, there was adequate power for charging a supercapacitor to operate a medical device. The MRC system was tested for operating a medical device directly without requiring any supercapacitors at an inter-coil air gap of 7 cm. Transfer efficiency and power were 21.14% and 2.079 W, respectively. The PZT, USS, and MRC systems were outperformed the methods described in preceding works in regard to the power generation, transfer efficiency, power, and distance. Future work will concentrate on developing all parameters for the PZT, USS, and MRC systems. In addition, a standalone microcontroller for reducing the energy consumption of the measurement part is planned.

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9. References

[1] Jegadeesan R, Guo Y X, and Je M, 2013 "Electric near-field coupling for wireless power transfer in biomedical applications," in 2013 IEEE MTT-S International Microwave Workshop Series on RF and Wireless Technologies for Biomedical and Healthcare Applications (IMWS-BIO), pp. 1-3.

[2] Le T, Huerta M H, Moravec A, and Cao H, 2018 "Wireless Passive Monitoring of Electrocardiogram in Firefighters," in 2018 IEEE International Microwave Biomedical Conference (IMBioC), pp. 121-123.

[3] Das R and Yoo H, 2015 "Wireless power transfer to a pacemaker by using metamaterials and yagi-uda antenna concept," in 2015 International Workshop on Antenna Technology (IWAT), pp. 353-354.

[4] Na K, Jang H, Ma H, and Bien F, 2014 "Tracking optimal efficiency of magnetic resonance wireless power transfer system for biomedical capsule endoscopy," IEEE Transactions on Microwave Theory and Techniques, vol. 63, pp. 295-304.

[5] Monti G, Arcuti P, and Tarricone L, 2015 "Resonant inductive link for remote powering of pacemakers," IEEE Transactions on Microwave Theory and Techniques, vol. 63, pp. 3814-3822.

[6] Campi T, Cruciani S, Feliziani M, and Hirata A, 2014 "Wireless power transfer system applied to an active implantable medical device," in 2014 IEEE Wireless Power Transfer Conference, pp. 134-137.

[7] Campi T, Cruciani S, Palandrani F, De Santis V, Hirata A, and Feliziani M, 2016 "Wireless power transfer charging system for AIMDs and pacemakers," IEEE transactions on microwave theory and techniques, vol. 64, pp. 633-642.

[8] Islam S and Kim A, 2018 "Ultrasonic Energy Harvesting Scheme for Implantable Active Stent," in 2018 IEEE International Microwave Biomedical Conference (IMBioC), pp. 70-72.

[9] Shi Q, Wang T, and Lee C, 2016 "MEMS Based Broadband Piezoelectric Ultrasonic Energy Harvester (PUEH) for Enabling Self-Powered Implantable Biomedical Devices," Scientific Reports, vol. 6, p. 24946.

[10] Vihvelin H, Leadbetter R J, Bance M, Brown A J, and Adamson B R, 2016 "Compensating for tissue changes in an ultrasonic power link for implanted medical devices," IEEE transactions on biomedical circuits and systems, vol.10, pp. 404-411.

[11] Meng M, Ibrahim A, and Kiani M, 2015 "Design considerations for ultrasonic power transmission to millimeter-sized implantable microelectronics devices," in 2015 IEEE Biomedical Circuits and Systems Conference (BioCAS), pp. 1-4.

[12] Meng M and Kiani M, 2017 "Design and optimization of ultrasonic wireless power transmission links for millimeter-sized biomedical implants," IEEE transactions on biomedical circuits and systems, vol 11, pp. 98-107.
[13] Ansari M and Karami A M, 2017 "Experimental investigation of fan-folded piezoelectric energy harvesters for powering pacemakers," *Smart Materials and Structures*, vol. 26, p. 065001.

[14] Oh T, Islam S, Mahfouz M, and To G, 2017 "A Low-Power CMOS Piezoelectric Transducer Based Energy Harvesting Circuit for Wearable Sensors for Medical Applications," *Journal of Low Power Electronics and Applications*, vol. 7, p. 33.

[15] Toprak A and Tigli O, 2015 "MEMS scale PVDF-TrFE-based piezoelectric energy harvesters," *Journal of Microelectromechanical Systems*, vol. 24, pp. 1989-1997.

[16] Zhang Y, Wang T, Luo A, Hu Y, Li X, and Wang F, 2018 "Micro electrostatic energy harvester with both broad bandwidth and high normalized power density," *Applied energy*, vol 212, pp. 362-371.

[17] Janusas G, Poneyte S, Brunius A, Guobiene A, Prosycevas L, Vilkauskas A, and Palevicius A, 2015 "Periodical microstructures based on novel piezoelectric material for biomedical applications," *Sensors*, vol. 15, pp. 31699-31708.

[18] Saadon S and Sidek O, 2015 "Micro-electro-mechanical system (MEMS)-based piezoelectric energy harvester for ambient vibrations," *Procedia-Social and Behavioral Sciences*, vol. 195, pp. 2353-2362.

[19] Safaei M, Meneghini M R, and Anton R S, 2018 "Energy harvesting and sensing with embedded piezoelectric ceramics in knee implants," *IEEE/ASME Transactions on Mechatronics*, vol 23, pp. 864-874.

[20] Pillatsch P, Yeatman M E, and Holmes S A, 2014 "A piezoelectric frequency up-converting energy harvester with rotating proof mass for human body applications," *Sensors and Actuators A: Physical*, vol. 206, pp. 178-185.

[21] K. Ghosh S and Mandal D, 2016 "High-performance bio-piezoelectric nanogenerator made with fish scale," *Applied Physics Letters*, vol. 109, pp. 103701.

[22] Jia Y and Seshia A A, 2015 "Power optimization by mass tuning for MEMS piezoelectric cantilever vibration energy harvesting," *Journal of Microelectromechanical Systems*, vol. 25, pp. 108-117.

[23] Li X, Tsui Y C, and Ki H W, 2015 "A 13.56 MHz wireless power transfer system with reconfigurable resonant regulating rectifier and wireless power control for implantable medical devices," *IEEE Journal of Solid-State Circuits*, vol. 50, pp. 978-989.

[24] Thangasamy V, Kamsani N A, Thiruchelvam V, Hamidon M N, Hashim S J, Bukhor M F i, and Yusoff Z, 2015 "Wireless power transfer with on-chip inductor and class-E power amplifier for implantable medical device applications," in *2015 IEEE Student Conference on Research and Development (SCOReD)*, pp. 422-426.

[25] A. Technology. MT3608 DC Voltage Regulator Module Available: https://prom-electric.ru/media/MT3608 (accessed on 20th March. 2019).

[26] Warudkar S, Deshmukh R, and Parihar V, 2018 "Power Monitoring System Using Microcontroller for Optimum Power Utility in homes," *Reinvention International: An International Journal of Thesis Projects and Dissertation*, vol. 1, pp. 96-112.

[27] Mnati M, Van den Bossche A, and Chisab R, 2017 "A smart voltage and current monitoring system for three phase inverters using an android smartphone application," *Sensors*, vol 17, pp. 872.

[28] Furter J S and Hauser P C, 2018 "Interactive control of purpose built analytical instruments with Fouth on microcontrollers-A tutorial," *Analytica chimica acta*, pp.18-28.

[29] Artanto D, Sulistyanto M P, Pranowo I D, and Pramesta E E, 2017 "Drowsiness detection system based on eye-closure using a low-cost EMG and ESP8266," in *2017 2nd International conferences on Information Technology, Information Systems and Electrical Engineering (ICITISEE)*, pp. 235-238.

[30] Pancholi S and Joshi A M, 2018 "Portable EMG data acquisition module for upper limb prosthesis application," *IEEE Sensors Journal*, vol. 18, pp. 3436-3443.
[31] Thomas S S, Saraswat A, Shashwat A, and Bharti V, 2016 "Sensing heart beat and body temperature digitally using Arduino," in 2016 International Conference on Signal Processing, Communication, Power and Embedded System (SCOPES), pp. 1721-1724.

[32] Valliappan S, Mohan B P R, and Kumar S R, 2017 "Design of low-cost, wearable remote health monitoring and alert system for elderly heart patients," in 2017 International Conference on IoT and Application (ICIOT), pp. 1-7.

[33] Narváez R B, Villacís D M, Chalen T M, and Velásquez W, 2017 "Heart rhythm monitoring system and iot device for people with heart problems," in 2017 International Symposium on Networks, Computers and Communications (ISNCC), pp. 1-5.

[34] Watthanawisuth N, Lomas T, Wisitsoraat A, and Tuantranont A, 2010 "Wireless wearable pulse oximeter for health monitoring using ZigBee wireless sensor network," in ECTI-CON2010: The 2010 ECTI International Conference on Electrical Engineering/Electronics, Computer, Telecommunications and Information Technology, pp. 575-579.

[35] Ahmed H S and Ali A A, 2016 "Smart intensive care unit design based on wireless sensor network and internet of things," in 2016 Al-Sadeq International Conference on Multidisciplinary in IT and Communication Science and Applications (AIC-MITCSA), pp. 1-6.

[36] ThoiThoi S, Kodur K C, and Ari’f W, 2016 "Quaternion based wireless AHRS data transfer using nRF24L01 and HC-05," in 2016 International Conference on Microelectronics, Computing and Communications (MicroCom), pp. 1-6.

[37] Nordic Semiconductor. Available: https://www.nordicsemi.com/Products/Low-power-short-range-wireless/nRF24-series (accessed on 19 March. 2019).

[38] Linsangan N B, Bagtas D L P, Itchon R T, Reyes J M L, Suarez G M, and Maramba R G, 2017 "Offline-based clicker using Gaussian frequency shift keying and arduino NRF24L01," in 2017IEEE 9th International Conference on Humanoid, Nanotechnology, Information Technology, Communication and Control, Environment and Management (HNICEM), pp. 1-5.

[39] Mahzan N N, Omar A M, Rimon L, Mohammad Noor S Z, and Rosselan M Z, 2017 "Design and development of an arduino based data logger for photovoltaic monitoring system," Int. J. Simul. Syst. Sci. Technol, vol. 17, pp. 15.1-15.5.

[40] Murata. Available: https://www.murata.com/en-global/products/sound/diaphragm (accessed on 5 February 2019 ).

[41] Silicon diode. SCHOTTKY BARRIER RECTIFIERS. Available: http://pdf.datasheetcatalog.com/datasheet/bytes/SR260.pdf (accessed on 6 February 2019 ).

[42] Mahmood M F, Mohammed S L, and Gharghan S K, 2019 "Ultrasound Sensor-Based Wireless Power Transfer for Low-Power Medical Devices," Journal of Low Power Electronics and Applications, vol. 9, pp. 20.

[43] Das S, Pal S, and Mitra M, 2016 "Real time heart rate detection from PPG signal in noisy environment," in 2016 International Conference on Intelligent Control Power and Instrumentation (ICICPI), pp. 70-73.