Propagation, Power Absorption, and Temperature Analysis of UWB Wireless Capsule Endoscopy Devices Operating in the Human Body

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Abstract—With the increasing use of wireless capsule endoscopy (WCE) devices in healthcare, it is of utmost importance to analyze the electromagnetic power absorption and thermal effects caused by in-body propagation of wireless signals from these devices. This paper studies the path loss, specific absorption rate (SAR), specific absorption (SA), and temperature variation of the human body caused by an impulse-radio ultra-wideband (UWB) based WCE operating inside the human abdomen. In addition, the design and in-body performance of an UWB antenna with dimensions of 11.85 $\times$ 9 $\times$ 1.27 mm and operating from 3.5 to 4.5 GHz is described in this paper. Path loss is evaluated using both experimental and simulation based methods to characterize the in-body propagation channel. The experimental setup uses a pig’s abdominal tissue samples to demonstrate the propagation characteristics of human tissue while a voxel model of the human body consisting of human tissue simulating materials is used in the simulations. The tissue properties, such as relative permittivity, are characterized according to the incident signal frequency and age of the tissue sample during simulations. The SAR and SA variations for different positions of the WCE device inside the colon and the small intestine of the human body model are analyzed using the finite integration technique as the discretization model. The dependency of the electromagnetic effects on the antenna positioning is investigated by using different positions of the antenna inside the human body.

Index Terms—Impulse-radio ultra-wideband (IR-UWB), path loss, specific absorption (SA), specific absorption rate (SAR), wireless endoscopy.

I. INTRODUCTION

WIRELESS capsule endoscopy (WCE) is a rapidly emerging technology used for diagnosis of diseases within the gastrointestinal (GI) tract [1]–[3]. Many narrowband based wireless capsules found in the literature suffer from limited battery life, which limits the operation time of the wireless capsule, and low image rate, which affects the clarity of the images taken from the device [4], [5]. Impulse-radio ultra-wideband (IR-UWB) wireless technology can provide high data rate for WCE devices while adhering to low-power and small form-factor requirements [6]–[8]. Ultra-wideband (UWB) signals have a fractional bandwidth larger than 0.2 or a bandwidth of at least 500 MHz and operate in the 3.1–10-GHz band with the indoor effective isotropic radiated power (EIRP) kept below $41.3 \text{ dBm/MHz}$ [9]. IR-UWB signals use short pulses to transmit data, as shown for the case of data transmission from a WCE device in Fig. 1. When implemented in CMOS technology, IR-UWB transmitters consume 30 times less power compared to narrowband transmitters [6]. Additionally, UWB-based transmitters can support image resolutions upwards from 640 $\times$ 480, while narrowband capsules operate at lower resolutions [5], [6]. Furthermore, UWB based capsules can support higher frame rates required by colonoscopy applications. Another main advantage of the UWB capsules is the fact that they can transmit images at a high data rate without...
any power-consuming image-compression algorithms that are used in narrowband capsules.

Path-loss variation defining the in-body propagation channel is a good indicator of the power absorption from the human body tissue. In-body path-loss variation for narrowband channels are studied in [10], [11]. The simulation results presented in [12] describe the variation of the path loss with the distance from a chest implanted UWB antenna using a human anatomical model. It mainly focuses on analyzing the variation of signal strength outside the human body with the varying implant depth. Another simulation-based study of the UWB propagation channel from outside-body to inside-body for chest implants is presented in [13]. The authors have used a UWB plane wave excitation signal with a particular polarization located outside the human chest as the UWB signal source for this study. Hence, it does not consider the complications that are introduced by an implanted antenna, as well as the presence of tissue reflections, which is an important factor in determining the power absorption for in-body propagation channels. Results presented in [8] analyze the path loss and bit error rate (BER) for IR-UWB communication using phantom liquid models and living animal experiments. Simulation and experimental based analyses of path loss for signals at various frequency bands emitted by WCE devices are carried out in [14]–[16].

The main objective of this paper is to analyze the power absorption characteristics of human tissue that is exposed to IR-UWB signals emitted by WCE devices operating inside the human body. In addition, this paper analyzes the in-body path loss for IR-UWB signals in the frequency range of 3.5–4.5 GHz using both experimental and simulation based approaches. A compact UWB antenna designed and fabricated for WCE devices that operates in the aforementioned frequency range is also described in this paper. Experiments are conducted using the fabricated antenna and a phantom model that uses a pig’s abdominal tissue samples in order to simulate human tissue. Porcine tissue provides a close approximation to human tissue in terms of dielectric properties at the frequencies of interest to this paper [17]–[19]. The dielectric properties of different tissue types used in the simulations carried out in this paper are taken from [20]. Simulation and experimental results for path loss are compared to validate the power absorption models that are used in the simulations that calculate specific absorption rate (SAR)/specific absorption (SA) and temperature increase of the human tissue that is exposed to IR-UWB signals. SAR and SA are used as key indices in measuring the electromagnetic power absorption by the human body in many international standards. For example, the International Council on Non-Ionizing Radiation Protection (ICNIRP) standard regulates the localized 10-g averaged SAR of the head and trunk to 2 W/kg for signals in the frequency range from 10 kHz to 10 GHz for the electromagnetic exposure of the general public [21]. In addition, pulse based transmissions are applied with a regulated SA value of 2 mJ/kg averaged over 10-g tissue weight by the ICNIRP standard [21] in order to prevent auditory effects. The SA has to be calculated over the duration of a single pulse according to the ICNIRP regulations. The electromagnetic power absorption results in a temperature increase in the human tissue. A temperature increase in excess of 1 ºC could prove harmful to the metabolic functions of the human body [21].

Recent studies analyze the electromagnetic effects of wireless implantable devices on the human body [22]–[25]. Many reported studies, such as [26] and [27], have used homogeneous human tissue models. Differences in human tissue properties, such as relative permittivity, for different tissue types and for different incident frequencies significantly affect the power absorption properties of the human tissue exposed to wideband signals [28]. Hence, it is not suitable to apply the human body models used in [26] and [27] for wideband applications. Also, the antenna properties, such as directivity, orientation, and antenna gain are key factors in determining the electromagnetic power absorption caused by the implanted wireless devices. The work presented in [23] has analyzed the SAR variation for WCE devices operating in medical implant communication service (MICS) and industrial–scientific–medical (ISM) bands, while [24] illustrates the SAR variation caused by an IR-UWB implant device inside the human stomach. The latter has not considered the effects of an antenna model in the simulations. Additionally, the Federal Communications Commission (FCC) regulations [9] that govern the UWB indoor propagation are not studied in [24].

The work carried out in this paper is an extension of the simulation-based analysis carried out in [29]. Extensive additions to the work presented in [29] are presented in this paper. These include design and fabrication of a miniaturized implantable or injectable antenna, experimental validation of the path-loss variations, improvements to the simulation scenarios using a 3-D Poynting vector instead of the 1-D method used in [29] for better representation of electromagnetic in-body power distribution, and calculating the SAR/SA for various positions and orientations of the implant antenna. Additionally, this work differs from the simulation based work presented in [28] due to the difference in in-body device location, addition of experimental and simulation based path-loss analysis, and experimental demonstration of the in-body IR-UWB communication system. A real-world hardware implementation of an IR-UWB based wireless capsule is presented in [30]. This device uses the implantable antenna presented in this paper for IR-UWB signal transmission. The device presented in [30] utilizes the power absorption analysis presented in this paper as the basis for its power control algorithm. A human anatomical body model developed using CST Studio is used to simulate the human tissue properties at UWB frequencies. SAR, SA, and temperature variation are analyzed for six different orientations of the WCE device that is inserted inside the colon and small intestine of the human body model. The simulated results are compared against the ICNIRP standard. This paper is organized as follows. Section II describes the design and fabrication of an UWB implant antenna and the WCE device positioning. Section III presents the in-body path-loss analysis. Section IV describes the SAR, SA calculation procedure, the bio-heat equation used to obtain the temperature variation in the human tissue, and the simulated results. Finally, Section V concludes this paper.
II. ANTENNA DESIGN AND WCE DEVICE POSITIONING

The UWB antenna used for the simulations and experiments described in this paper and its various orientations inside the human stomach are shown in Fig. 2. The antenna operates at a 4-GHz center frequency with a bandwidth of approximately 1 GHz. The 4-GHz center frequency is chosen to minimize the interference from other wireless technologies, such as the 5-GHz WiFi, in a practical scenario. The dimensions of the antenna are 11.85 \times 9 \times 1.27 \text{mm}. These dimensions are comparable with the commercially available capsule sizes used for WCE [2]. The antenna is fabricated on Rogers TMM 10 i high frequency material. The wide-slot antenna design with a U-shaped feed forms a magnetic dipole, which is less susceptible to variations in the near-field propagation environment. In the simulations, the antenna model is inserted in a capsule-shaped case with a diameter of 9.5 mm. The thickness of the capsule walls is negligible compared to the dimensions of the antenna. Using an insulating material between the dispersive tissue medium and the radiating element of the antenna improves the impedance-matching characteristics of an implantable antenna [28], [31]. The radiating element of the antenna used in the simulations, which occupies the lower half of the antenna, is inserted in glycerin for this purpose. Glycerin has a relative permittivity of 50, which is close to the relative permittivity of the surrounding tissue material, and hence, allows minimal reflections of the electromagnetic wave near the transitional boundaries between the tissue medium and the capsule.

The WCE device traveling inside the GI tract is free to move and rotate in any direction [31]. Six different positions and three main orientations of the WCE antenna [see Fig. 2(a) and (b)] are considered in order to represent these various positions for the SAR and temperature simulations described in the latter part of this paper. The objective of this is to analyze the dependency of the electromagnetic effects on antenna positioning.

A Cartesian coordinate system is used to define the WCE device positions in the simulations. The antenna orientation along the Z-axis (Fig. 2) is considered for four positions of the antenna, and is referred to as z-z orientation in the remainder of this paper. Similarly, antenna orientation along X-axis is referred to as x-x orientation and antenna orientation along Y-axis is referred to as y-y orientation. The antenna at position-A in Fig. 2 is used as the reference position in order to refer to the locations of all other positions. The antenna in position-A is inserted inside the small intestine, at a distance of 89 mm from the front surface of the stomach, 88 mm from the left side of the stomach, and 645 mm from the top of the head, as shown in Fig. 2. Table I summarizes the parameters related to the WCE locations used in the simulations that follow.

The experimental setup used for the power measurement experiments is shown in Fig. 3. Various abdominal tissue samples taken from young and well-developed porcine specimen is used...
in the setup. A high dielectric polycarbonate is used to make the container. The polycarbonate contains induced thallium chloride, which increases the dielectric constant of the material close to the average dielectric constant of tissue, and hence, minimizes the reflections at the separating walls. Glycerine, which has a dielectric constant close to the surrounding tissue, is applied on polycarbonate walls. This further minimizes the reflections that occur due to the polycarbonate walls

\[ G = 4\pi \frac{P_{\text{rad}} - P_{\text{tissue}}}{P_i} \]  

where \( G \) is the 3-D gain, \( P_{\text{rad}} \) is the power radiated per unit solid angle, \( P_{\text{tissue}} \) is the power absorbed by tissues within the unit solid angle, and \( P_i \) is the accepted power of the antenna after the antenna reflections. It should be noted that the antenna gain is high for antenna positions corresponding to low tissue absorptions. It can been seen from the far-field results that the 2-D antenna gain varies from \(-76\) to \(-40.8\) dBi among the six WCE positions. The minimum antenna gain of \(-76\) dBi was recorded for position-A, and hence, it corresponds to the WCE position with maximum tissue absorption.

Fig. 4 depicts the simulated S-parameters of the antenna for position-B and measured S-parameters when the UWB transmit antenna is inserted in the small intestine tissue sample of the experimental setup in Fig. 3. The 2-D polar plots of the simulated far-field gain of the antenna at 4 GHz for four important antenna orientations are shown in Fig. 5. The far-field gain is shown on the \( x-y \) plane that passes through the center of antenna. The 3-D far-field antenna gain is calculated using (1), and converted into the 2-D plots using the gain values at the intersections with the \( x-y \) plane.

It was observed from the simulations that the maximum 3-D gain is slightly higher than the 2-D gain, and lies in the same direction. Maximum simulated 3-D gain for each antenna position is mentioned in Fig. 5(a). Fig. 5(b) depicts measured far-field gain of the antenna together with the simulated far-field results for position-B. The antenna far field is measured in the following manner. Firstly, the UWB implant antenna is inserted in pig’s small intestine tissue sample, as shown in Fig. 3. The receiving horn antenna is positioned in the far field of the UWB implant antenna (at a distance of 8 cm from the UWB antenna in this case). The S-parameters (\( S_{11} \) and \( S_{21} \)) are recorded for this position. This measurement procedure is repeated for 12 positions on a circumference of a circle that is centered on the implanted UWB antenna. The gain is significantly low because of the large power absorption by the surrounding tissue mass. The recorded negative antenna gain after the tissue absorption means that a power level that is significantly higher than the FCC recommended spectral mask of \(-41.3\) dBm/MHz for indoor transmission of IR-UWB signals can be used for the antenna excitation, given that the regulations applied for SAR and SA are met [32], [33]. The delivered power to the antenna can be arranged such that the IR-UWB power level after the power loss due to tissue absorption lies within the FCC approved spectral mask.
III. PATH-LOSS VARIATION

A. Simulation Studies for Path Loss

In-body path-loss variation demonstrates the attenuation of signal power within the tissue medium. It demonstrates the incident power of tissues located at various depths from the WCE device. The Poynting vector at a certain point of an electromagnetic field is given by (2) [34],

\[
\overrightarrow{S}(t)_{x,y,z} = \overrightarrow{E}(t)_{x,y,z} \times \overrightarrow{H}(t)_{x,y,z} \quad \text{W/m}^2
\]

where \(\overrightarrow{S}(t)_{x,y,z}\) is the Poynting vector, \(\overrightarrow{E}(t)_{x,y,z}\) is the directional electric field intensity, and \(\overrightarrow{H}(t)_{x,y,z}\) is the directional magnetic field intensity at \((x, y, z)\) coordinates of the electromagnetic field. The corresponding energy flux can be calculated as

\[
F_{x,y,z} = \int \left| \overrightarrow{S}(t)_{x,y,z} \right| \, dt \quad \text{J/m}^2
\]

where \(t\) represents the observation time for the signal power at \((x, y, z)\). For the simulations presented in this paper, this time is defined as the time taken for the energy at \((x, y, z)\) to attenuate 80 dB from its peak value. The 80-dB attenuation value is chosen arbitrarily and it results in an observation time of 5 ns, which is a sufficient time for an IR-UWB pulse to propagate through surrounding tissue. The path loss between two points within an electromagnetic field is defined as the difference between received powers at those two particular points. The received power at a point in an electromagnetic field is proportional to the total received energy at that particular point. Hence, the path-loss variation between two points within an electromagnetic field can be obtained by (4),

\[
\text{PL}_{\text{dB}} = 10 \log \left( \frac{P_1}{P_0} \right) = 10 \log \left( \frac{\overline{\text{S}}_1}{\overline{\text{S}}_0} \right) = 10 \log \left( \frac{F_1}{F_0} \right)
\]

where \(\text{PL}_{\text{dB}}\) is the path loss in dB, \(P_0, \overline{\text{S}}_0,\) and \(F_0\) corresponds to the signal power, Poynting vector, and energy flux, respectively, at the reference point for the path-loss calculation, and \(P_1, \overline{\text{S}}_1,\) and \(F_1\) represents the signal power, Poynting vector, and energy flux, respectively, at the point at which the path loss should be evaluated. Simulations presented in this paper evaluates the time-varying Poynting vector at points in a 3-D space for the WCE device at position-A. It was observed from the simulations that the near field follows the same directional characteristics as the far field of the antenna. Hence, the Poynting vector observation points are defined in a 160 mm(W) × 90 mm(L) × 120 mm(H) cube that covers the 3-dB angular width of the far-field pattern of the WCE antenna, as shown in Fig. 6.

Fig. 7 depicts the simulated in-body path-loss variation. For the simulation results, only the data points that lie within the tissue medium are taken into consideration in order to plot the in-body channel characteristics. A dot in Fig. 7 represents the path loss at a certain point with respect to the reference point. It can be observed from simulation results that the path loss of the equidistant points from the reference point shows a scattering characteristic. This can be mainly attributed to different dielectric properties of different tissue types and multipath reflections of the electromagnetic signal.

A fitting curve for the scatter plot of the simulated path-loss values can be obtained in order to characterize the path loss using a generalized distance dependent path-loss equation, which is not affected by the tissue properties and multipaths. A fitting curve for the path-loss points in Fig. 7 is obtained using least square fitting and is plotted in Fig. 7. This fitting curve is characterized by the distance dependent fitting equation given in (5). The probability density function (PDF) for the scattering of the path-loss values around the fitted curve can be obtained from a normal distribution \(N\), as shown in (5),

\[
\text{PL}(d)_{\text{dB}} = \text{PL}(d_0)_{\text{dB}} + \alpha(d)^\gamma + N(\mu, \sigma^2)
\]

where \(\text{PL}(d)_{\text{dB}}\) is the path loss at a distance \(d\) in dB, \(\text{PL}(d_0)_{\text{dB}}\) is the path loss at the reference point, which is equal to 0 dB in this case, \(d\) is the distance to the point at which the path loss should be measured from the reference point in mm, \(\alpha = -5.828\) is the fitting constant, \(\gamma = 0.611\) is the path-loss exponent, \(\mu = 0.2684\) is the mean of the normal distribution \(N\), and \(\sigma = 8.7438\) is the standard deviation of \(N\). The histogram for the variation of the path loss from the least square fit line for the simulated data is plotted in Fig. 8 together with...
It should be noted that the total path loss is the mass density and 
\( (7) \)

transmit antenna insertion location inside meat. The free-space path-loss variation for various transmit–receive separations that are 1 cm apart from each other are measured by changing the transmit antenna insertion location inside meat. The free-space path-loss component, which is calculated during the initial measurement, is deducted from the path-loss values obtained from later measurements in order to calculate the in-body path-loss component according to the following equation:

\[
\text{PL}_{\text{in-body}} - \text{PL}_{\text{total}} - \text{PL}_{\text{free-space}}. 
\] (7)

The experimental path-loss measurements are plotted together with the simulated results in Fig. 7. It can be seen from Fig. 7 that the simulation results closely follow the experimental results that are obtained using the meat experiments. This close agreement between the results confirms the accuracy of the power absorption model used in the simulations, and hence, validates the SAR/SA and temperature results in the following section.

As a validation of the path-loss measurements conducted using power domain measurements, a time-domain experiment is carried out using the experimental setup shown in Fig. 9(a). A UWB transmitter and a UWB receiver front-end described in [36] and [37] are used for UWB pulse transmission and reception. The UWB receiver front-end amplifies the received signal by 48 dB using three low-noise amplifiers (LNAs) and a further 10 dB using a wideband amplifier [37]. Transmit and receive signals after the receiver front-end are depicted in Fig. 9(b) and (c), respectively. Transmit UWB pulses resulted in a frequency spectrum with a peak spectral amplitude of \(-12 \, \text{dBm}\) and lies in the frequency range of \(3.5\text{–}4.5 \, \text{GHz}\), as shown in Fig. 9(d). Received signals are obtained after an in-body propagation distance of 60 mm. The received power spectrum obtained after the receive antenna and first bandpass filter of the front-end [37] shows a peak spectral amplitude of \(-85 \, \text{dBm}\) [see Fig. 9(e)]. This is consistent with the calculated values considering an average measured path loss of 72 dB (see Fig. 7). Fig. 10 depicts the measured path-loss variation at 60-mm in-body propagation distance over the frequency range from 3.5 to 4.5 GHz, which is the target frequency range of the antenna. It should be noted that the path loss calculated using \(S_{21}\) is affected by different antenna patterns at various frequencies and differences in interactions between transmit and receive antennas at different frequencies.

### IV. SAR, SA, and Temperature Variation

#### A. SAR and SA Calculation Method

The SAR of a certain material exposed to an electromagnetic field can be derived from (8) [38],

\[
\text{SAR} = \frac{1}{2} \rho \sigma |E|^2 \] (8)

where \(E\) is the root mean square (rms) electric field strength (V/m), \(\rho\) is the mass density (kg/m\(^3\)), and \(\sigma\) is the conductivity of the tissue (S/m). The dependency of the electric field strength on the relative permittivity of the material affected by the electric field is demonstrated in Maxwell’s equations. Hence, the SAR is inherently dependent on the relative permittivity of a material [28]. The dependency of the relative permittivity of a particular material on the frequency of the incident signal is modeled by the so-called 4-Cole–Cole model approximation given in [28] and [39]. The tissue parameters required for this

#### B. Experimental Validation

The experimental setup shown in Fig. 3 is used for path-loss measurements. An UWB antenna is inserted in a plastic capsule shaped casing with its transmitting element immersed in a non-flowing glycerin based gelatin medium. This is then implanted inside a chamber that contains different tissue samples from a pig’s stomach, as shown in Fig. 3. A pig’s tissue is used in the experiments due to the similarities between pork and human tissue properties at microwave frequencies [21]–[23], [35]. A wide-band horn antenna with an internal ridged structure is used as the receiving antenna. The horn antenna has an operational bandwidth from 1.5 to 6 GHz. This horn antenna is used mainly due to its wideband nature and constant gain throughout the bandwidth from 3.5 to 4.5 GHz. Transmit and receive antennas are connected to a vector network analyzer (VNA). Path-loss measurements for various transmit–receive separation values are obtained using the following equation [10]:

\[
\text{PL}_{\text{dB}} = -|S_{21}|_{\text{dB}}. \] (6)

An input signal power of 0 dBm is used for the excitation of the transmit antenna. Initially, path loss for the free-space portion of the propagation channel is obtained by placing the transmit antenna near the skin/air interface inside the containing tank. Path-loss variation for various transmit–receive separations that are 1 cm apart from each other are measured by changing the transmit antenna insertion location inside meat.
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Fig. 9. (a) Experimental setup used for measurements. (b) Overall arrangement of the measurement setup. (c) Measured transmit-UWB pulse train. (d) Measured receive UWB pulse train after the front-end for an in-body propagation distance of 60 mm. (e) Measured transmit signal spectrum. (f) Measured receive signal spectrum after the horn antenna and the first bandpass filter.

Fig. 10. In-body path-loss variation over the target frequency range of the UWB implant antenna for an in-body propagation distance of 60 mm.

Fig. 11. IR-UWB excitation pulse.

content of tissue with age. The age of the human model used in the simulations is assumed to be 55 years old.

The finite integration technique (FIT) [42] is used for volume discretization in the described simulations. CST Microwave Studio, which is an FCC approved electromagnetic simulation software, is used for the finite-element simulations. SAR is defined as the power absorbed by the mass contained within a discretized volume element, as shown in (9) [21],

\[
\text{SAR} = \frac{d \left( \frac{\Delta W}{dV} \right)}{dt}
\]

where \(\Delta W\) is the power absorbed by the mass of the discretized volume element (W), \(dV\) is its incremental volume (m^3), \(\rho\) is the density of the absorbing material (kg/m^3), and \(dt\) is the incremental time (s). Simulations are carried out in order to calculate the 10-g averaged SAR so as to compare it with the ICNIRP specifications for pulse transmission [21]. The per pulse SA, which is used by the ICNIRP standard in order to introduce additional limitations for pulsed transmissions, can be computed using (10),

\[
\text{SA} = \text{SAR} \times T_p
\]

where \(T_p\) is the pulse duration.

B. Temperature Variation

The electromagnetic power absorption of the tissue material results in an increase in the tissue temperature. The simulations presented in this paper apply the bio heat equation given in
Fig. 12. (a)–(c) Side view and (d)–(f) top view of the simulated ten average SAR variations in the human voxel model for IR-UWB pulses with peak spectral input power limits of (a)–(d) 41.3 dBm/MHz, (b)–(e) 1.82 dBm/MHz, (c)–(f) 21.7 dBm/MHz and total in-band signal power of (a)–(d) 0.0024 mW, (b)–(e) 21.5 mW, and (c)–(f) 4.38 W.

Fig. 13. Side view of the simulated temperature variation in the human voxel model for IR-UWB pulses with peak spectral input power limit of: (a) 41.3 dBm/MHz with a total in-band signal power of 0.0024 mW and (b) 1.82 dBm/MHz with a total in-band signal power 21.5 mW.

[28] and [43] in order to analyze the temperature variation in the human tissue when it is exposed to IR-UWB transmission from the WCE device. The temperature dependency of the basal metabolic rate and blood perfusion is also considered in the simulations following the methods provided in [28], [43], and [44].

C. SAR, SA, and Temperature Variation Results

SAR and temperature simulations are carried out for two scenarios. In the first scenario, SAR, SA, and temperature variations are simulated for different accepted input power levels of the WCE antenna placed at a fixed position. In the second scenario, simulations are carried out in order to observe the SAR, SA, and temperature variations for different positions of the WCE device while keeping the antenna input power at a fixed value.

Position-A is chosen as the WCE device position for the first scenario as it corresponds to the antenna position with the minimum far-field gain (i.e., the maximum tissue absorption). The antenna is excited using an IR-UWB pulse with a center frequency of 4 GHz and a bandwidth of 1 GHz. The pulsewidth is set to be at 2 ns, as shown in Fig. 11. The pulse is extracted from a pulse train with a period of 50 ns. Simulations are carried out using both the FCC regulated IR-UWB pulse and an IR-UWB pulse with a power spectrum that is higher than the FCC spectral mask in order to assess and compare the SAR variations.

An IR-UWB pulselength of 2 ns (Fig. 11) is used for the calculation of the SA. The in-band power of the IR-UWB pulses for the three different scenarios is varied by changing the pulse amplitude. The total in-band power for each simulation is calculated by integrating the power spectrum of the IR-UWB pulse in the frequency band of 3.5–4.5 GHz. Fig. 12 presents the SAR variations for the first scenario. Fig. 12(a) shows the SAR variation for an IR-UWB pulse with a total in-band signal power of 0.0024 mW.

Table II

<table>
<thead>
<tr>
<th>Position</th>
<th>Max.10 g averaged SAR (W/kg)</th>
<th>Max.10 g averaged SA (nJ/kg)</th>
<th>Max.1 g averaged SAR (W/kg)</th>
<th>Max. localized temperature increase (°C)</th>
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<tr>
<td>A</td>
<td>2.00</td>
<td>4.00</td>
<td>6.65</td>
<td>0.354</td>
</tr>
<tr>
<td>B</td>
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<td>3.76</td>
<td>6.70</td>
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</tr>
<tr>
<td>C</td>
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<td>1.95</td>
<td>3.90</td>
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Fig. 14. Total SAR variations for different tissue types in each position.
– 41.3 dBm/MHz. The SAR variation in Fig. 12(b) corresponds to an IR-UWB pulse that causes a maximum 10-g averaged SAR value of 2 W/kg, which is the ICNIRP allowed SAR limit. Fig. 12(c) depicts the SAR variation for an IR-UWB pulse that results in a signal power level just outside the human body to lay within the FCC regulated spectral mask (refer to Section II). The maximum SA is calculated for each scenario using (4). The color scale in Fig. 12 is set to reflect the maximum SAR in all the scenarios, and is logarithmically marked to yield an acceptable resolution for low SAR values.

The results depicted in Fig. 12 show that the SAR variation for the third scenario, where the input signal power is adjusted in order to obtain a signal with a power level that lies within the FCC allowed spectral mask just outside the body, corresponds to an SAR variation that exceeds the ICNIRP allowed level of 2 W/kg. In other words, it can be observed from the results that it is the ICNIRP regulated SAR level of 2 W/kg that determines the maximum allowable delivered signal power to the antenna. This corresponds to an IR-UWB pulse with a total in-band signal power of 21.5 mW. Hence, this input power value is used as the reference input antenna power for the WCE device positions in the second scenario. The SAR variation shown in Fig. 12(a) is comparatively lower because of the small in-band power contained in the FCC regulated IR-UWB input pulse.

The temperature variations for the same input power scenarios used for Fig. 12(a) and (b) are shown in Fig. 13. An initial body temperature of 37°C is chosen in the simulations. Final temperature results are obtained after a period of 6 min according the ICNIRP standards. It can be observed from Fig. 13(a) that the temperature of the whole body has increased to 37.173°C from the initial body temperature of 37°C due to the metabolic activities of the body. In this case, the delivered signal power is not enough to cause a significant temperature increase in the tissue surrounding the WCE device. The blood perfusion used in the simulations regulates the minute temperature increase caused by the small power absorption in this scenario. It can be observed in Fig. 13(b) that the temperature of the tissues at a close proximity to the WCE device has increased up to a maximum of 37.354°C while the rest of the tissues showed a temperature of 37.173°C because of the metabolic activities.

Table II depicts power absorption and temperature variations for different positions of the WCE device for an IR-UWB signal with a total in-band input power of 21.5 mW. Maximum recorded 1-g averaged SAR value is also presented in the table in order to show the variations of a more localized SAR value for different WCE positions. According to the results in Table II, the maximum 10-g averaged SAR is reported for the position-A of the WCE device, and is equal to the ICNIRP regulated 10-g averaged SAR level of 2 W/kg. The maximum 10-g averaged SA value stays well below the ICNIRP regulated level of 4 mJ/kg for all the WCE positions. The SA value is low for these simulations mainly due to the nanosecond scale pulse duration. The 1-g averaged SAR value shows a maximum of 7.29 W/kg for the position-F of the WCE device. It should be noted that the 1-g averaged SAR uses a much lower averaging mass for SAR computation and represents a higher localized tissue power absorption. The maximum temperature increase of 0.438°C is reported for the position-C where the WCE device is inserted in the colon. The temperature increase depends on the physical properties of the surrounding tissue such as the density and the specific heat capacity, as well as the SAR value.

Fig. 14 depicts the total SAR variation for various tissue types exposed to the IR-UWB signals during the simulations. The scale is marked logarithmically to give a better resolution to lower SAR values. The total SAR is calculated by dividing the total power absorbed by each tissue type by the total mass of each tissue type present in the simulation domain. It can be observed from Fig. 14 that the tissue type nearest to the WCE device absorbs most of the electromagnetic energy. Table III compares the evaluated SAR values in this paper with related work in the literature. SAR due to the operation of a wireless capsule operating within the abdomen is presented in [23]. This application uses various narrowband signals with an input power of 25 mW. Reported SAR values in [23] are comparatively smaller than the SAR values presented in this work for scenarios that uses a total input power of 21.5 mW. This is mainly due to narrow bandwidth and lower incidence frequencies used in [23].

A SAR of 8.95 W/kg in abdominal tissues is reported in [24] for an IR-UWB signal operating around 8.75 GHz with a total power of 1 W. SA values ranging from 0.037 to 0.476 pJ/kg in human skin exposed to IR-UWB signals with a regulated spectral amplitude of –41.3 dBm/MHz is presented in [45]. These values are comparable with the SA values presented in this work for an IR-UWB signal with similar spectral amplitude.

<table>
<thead>
<tr>
<th>Reference</th>
<th>Scenario</th>
<th>Body part</th>
<th>Reference input power</th>
<th>Frequency</th>
<th>MAX 10g SAR/SA</th>
</tr>
</thead>
<tbody>
<tr>
<td>[23]</td>
<td>In-body</td>
<td>Abdomen</td>
<td>25 mW</td>
<td>2.4 GHz</td>
<td>0.37 W/kg (SAR)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>1.2 GHz</td>
<td>0.64 W/kg (SAR)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>800 MHz</td>
<td>0.66 W/kg (SAR)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>430 MHz</td>
<td>0.62 W/kg (SAR)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>400 MHz</td>
<td>0.54 W/kg (SAR)</td>
</tr>
<tr>
<td>[24]</td>
<td>In-body</td>
<td>Abdomen</td>
<td>1 W (IR-UWB)</td>
<td>8.75 GHz</td>
<td>8.95 W/kg (SAR)</td>
</tr>
<tr>
<td>[45]</td>
<td>On-body</td>
<td>Skin</td>
<td>-41.3 dBm/MHz regulated IR-UWB</td>
<td>3.1-10.6 GHz</td>
<td>0.037-0.476 pJ/kg (SA)</td>
</tr>
<tr>
<td>This work</td>
<td>In-body</td>
<td>Abdomen</td>
<td>-41.3 dBm/MHz regulated IR-UWB</td>
<td>3.5 – 4.5 GHz</td>
<td>0.2 mW/kg (SAR)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>-1.8 dBm/MHz (21.5 mW) IR-UWB</td>
<td></td>
<td>0.4 pJ/kg (SA)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>2 W/kg (SAR)</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>4 nJ/kg (SA)</td>
</tr>
</tbody>
</table>
V. CONCLUSION AND COMPARISON

This paper has analyzed the power absorption characteristics of human tissue exposed to IR-UWB signals emitted by a WCE device that operates inside the human body. In-body path loss has been analyzed using both experimental and simulation based methods. Both approaches have resulted in a path loss that varies between $-80$ and $-100$ dB for an in-body propagation distance of approximately 10 cm. The agreement between the experimental and simulation based results validates the power absorption model used in the simulations. The electromagnetic power absorption caused by IR-UWB based WCE devices has also been studied in this paper. The effects have been analyzed in terms of SAR, SA, and temperature increase considering the UWB communication regulation and safety standards.

Studies in this paper have shown that the maximum SAR value defined by the ICNIRP standard determines the maximum IR-UWB transmit power that can be utilized for a WCE device. The maximum allowable total in-band power per pulse was found to be 21.5 mW for the various positions of a WCE device investigated in this paper. The temperature increase caused by this transmit power level is well within the control of thermal regulatory mechanisms of the human body. A peak temperature increase of 0.438 °C above the original body temperature was recorded when the WCE device was operating inside the colon.

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The authors would like to thank Dr. T. Dissanayake for his help in designing the UWB antenna used for the high-frequency simulations. The authors would also like to thank the Monash MMARS Laboratory for providing the horn antenna.

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[3] T. D. Than, G. Alici, H. Zhou, and W. Li, “A review of localization based methods. Both approaches have resulted in a path loss that varies between $-80$ and $-100$ dB for an in-body propagation distance of approximately 10 cm. The agreement between the experimental and simulation based results validates the power absorption model used in the simulations. The electromagnetic power absorption caused by IR-UWB based WCE devices has also been studied in this paper. The effects have been analyzed in terms of SAR, SA, and temperature increase considering the UWB communication regulation and safety standards.

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REFERENCES

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