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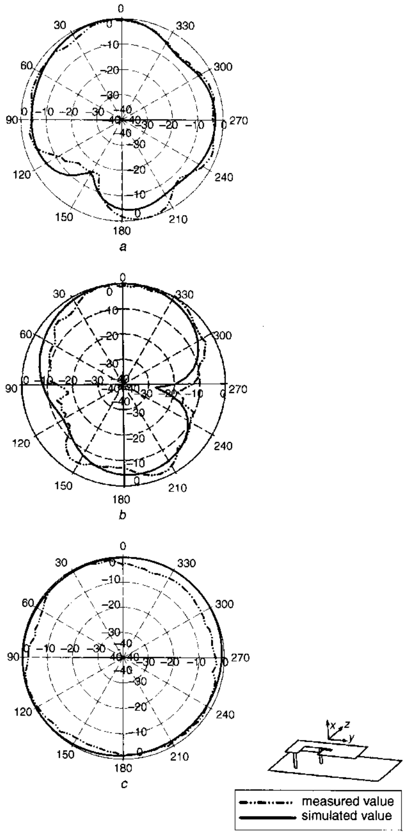

Fig. 5 *Measured and simulated radiation patterns at 5.5 GHz*
a x-y plane; *b x-z* plane; *c y-z* plane

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Path loss model for wireless narrowband communication above flat phantom

L. Roelens, S. Van den Bulcke, W. Joseph, G. Vermeeren and L. Martens

A new empirical path loss model for wireless communication at 2.4 GHz above a flat, lossy medium, representing human tissue, is presented. The model is valid for dipole antennas for heights up to 5 cm above the phantom and for distances up to 40 cm, and was applied to muscle and brain simulating media. For antennas placed close to the lossy medium, it was found that antenna height has a major influence on path loss. The model has been validated by measurements and simulations, which show excellent agreement.

Introduction: A wireless body area network (WBAN) is a network for which the nodes are located in the clothes on the body or under the skin of a person. These nodes are connected through a wireless communication channel and form a network that typically expands over the body of a person. According to the implementation, the nodes consist of sensors and actuators, placed in a star or multihop topology [1]. WBANs have many promising, new applications in medicine, multimedia and sports, all of which make use of the unconstrained freedom of movement a WBAN offers.

An important step in the development of a narrowband WBAN is the estimation of path loss (PL) between two nodes on the body. This requires detailed characterisation of the electromagnetic wave propagation near the human body. To date, only few attempts, which lack verification by measurements and do not investigate the influence of antenna height, have been made [2]. Therefore, we study the wave propagation above a lossy medium including brain and muscle tissue, using two commercial simulation tools and measurements. Our objective is to develop an empirical PL model, which takes the influence of antenna height into account. The simulations are performed with FEKO, a Method of Moments (MoM) program, and verified by SEMCAD, a finite-difference time-domain (FDTD) program. Simulations and measurements are performed at 2.4 GHz in the licence-free industrial, scientific and medical (ISM) band.

Method: Model – To model the PL between two dipole antennas above a specific lossy medium, we use the following semi-empirical formula, expressed in dB and based on the Friis formula in free space:

$$\left.\frac{P_{rec}}{P_{trans}}(d,h)\right|_{\mathrm{dB}} = P_0(h)|_{\mathrm{dB}} - n(h)d|_{\mathrm{dB}} = |S_{21}|_{\mathrm{dB}} \quad (1)$$

where P_{rec} and P_{trans} are received and transmitted power, respectively, and d is the distance between the transmitting antenna Tx and receiving antenna Rx. P_0 is path loss at a reference distance d_0 (here chosen to be 40 cm) and n is the so-called PL exponent which equals 2 in free space. Both P_0 and n depend on the height h of both antennas above the medium. The last part of (1) allows us to regard the setup as a two-port network for which we determine the S_{21}-parameter.

Because we want to investigate the influence of the medium on PL, we limit ourselves to a flat, conducting, dielectric and uniform medium, characterised by a specific relative permittivity ε_r and conductivity σ. Furthermore, to model the PL, we use elementary half-wavelength dipoles. The setup and parameters for determination of the PL are shown in Fig. 1.

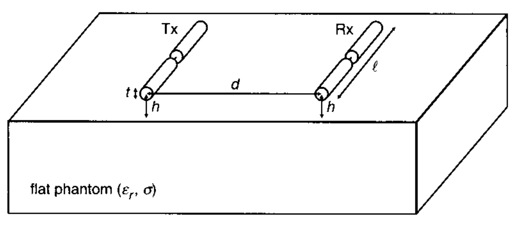

Fig. 1 *Setup for PL measurements and simulations*

Simulations and measurements – Using the MoM tool, we simulate the PL for distances between Tx and Rx up to 40 cm and antenna heights from 5 up to 50 mm above the lossy medium. Longer distances and heights are impractical for a multihop WBAN and are not considered. Both muscle ($\varepsilon_r = 53.57$ and $\sigma = 1.81$ S/m at 2.4 GHz) and brain tissue ($\varepsilon_r = 38.5$ and $\sigma = 1.9$ S/m at 2.4 GHz) are simulated. We make use of half-wavelength dipoles with length $\ell = 0.46\,\lambda = 5.75$ cm and a realistic diameter $t = 1$ mm (see Fig. 1). An applied E-field source model, which corresponds to an applied voltage over the source segment, is used and the length of the source segment is made equal to the length of the other segments. Furthermore, we can model the flat phantom by the built-in Green's function for planar, multilayer substrates. This greatly speeds up computation time compared to FDTD methods, since the influence of the presence of the flat phantom is taken into account implicitly.

For verification purposes, simulations were performed at an antenna height of 1 cm using the FDTD method. The flat phantom is now modelled by a rectangular box with a height of 75 mm. The FDTD cell size varies between 0.25 and 8.75 mm to ensure a maximum cell dimension around $\lambda/10 = 0.2$ mm in the medium and well below $\lambda/10 = 12.5$ mm in free space at 2.4 GHz. Because of these small cell sizes, the fields in a maximum grid size of 11 million cells have to be calculated for 10 periods of the excitation to obtain accurate results. This leads to long computation times compared to the MoM simulations.

To verify the simulations, we performed measurements with a network analyser (Rohde & Schwarz ZVR). Two half-wavelength dipoles with length of 5.75 cm and diameter of 1 mm are placed close to a rectangular box phantom recommended by the CENELEC standard EN50383 [3] with dimensions 80 × 50 × 20 cm. The phantom shell has a thickness of 1 cm (±1 mm) and is filled with brain simulating liquid ($\varepsilon_r = 38.5$ and $\sigma = 1.9$ S/m at 2.4 GHz) and was also modelled in the MoM and FDTD simulations.

The measurements are performed in a non-anechoic environment, resulting in undesired reflections. Therefore, we performed a de-embedding step [4]. At each position of the antennas, we measured $S_{21}(f)$ from 300 kHz up to 4 GHz. This frequency range of 3.9997 GHz is necessary to distinguish the direct and reflected waves in the time domain. Next, we took the inverse FFT to derive $S_{21}(t)$. Reflections in $S_{21}(t)$ are then mitigated by a tenth-order Butterworth digital bandpass filter, characterised by a flat passband and the absence of sidelobes. Finally, we took the FFT to obtain $S_{21\,filtered}(f)$, corrected for reflections.

Results: Validation – Fig. 2 shows simulation and measurement results for Tx and Rx both 1 cm above the phantom and distances up to 40 cm. We obtained very good agreement between MoM and FDTD and an average deviation of 1.8 dB between measurements and simulations. Contributing to this deviation is the difference of 1.4 dB found between the measured (2.8 dB) and simulated (4.2 dB) gain of the system consisting of both antennas in free space. Also, the effective thickness of the phantom shell and positional errors cause measurement inaccuracies. The small deviations between MoM and FDTD allow us to perform the further investigation in this Letter with the faster MoM tool.

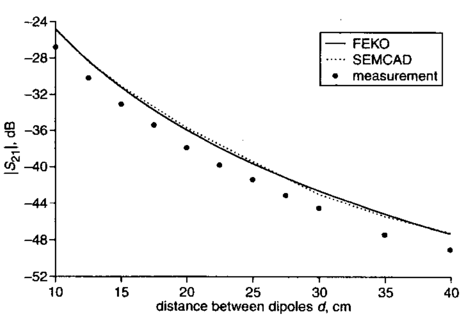

Fig. 2 *Influence of distance between Tx and Rx on $|S_{21}|$ (brain tissue)*

Influence of height – We performed a series of simulations with the MoM tool for varying antenna heights above the phantom, from 5 up to 50 mm, and distances up to 40 cm. The results for $|S_{21}|_{\rm dB}$ are shown in Fig. 3 for brain simulating tissue. It is clear that $|S_{21}|_{\rm dB}$ strongly depends on antenna height and drops quickly as the height decreases. The PL model of (1) is now fitted to the data obtained from these simulations. The results of these fits are shown in Fig. 4. The following PL model for wave propagation above brain tissue is obtained:

$$\begin{aligned} n_{brain}(h) &= -25.0h + 4.0 \\ P_{0,brain}(h)|_{\rm dB} &= 7.7\ln(h) - 11.9, \quad h \le 0.15\lambda \\ &= 388.7h - 49.4, \quad h > 0.15\lambda \end{aligned} \qquad (2)$$

where ln is the natural logarithm. For wave propagation above muscle simulating tissue we obtain:

$$\begin{aligned} n_{muscle}(h) &= -25.4h + 4.0 \\ P_{0,muscle}(h)|_{\rm dB} &= 7.7\ln(h) - 12.1, \quad h \le 0.15\lambda \\ &= 404.1h - 49.9, \quad h > 0.15\lambda \end{aligned} \qquad (3)$$

In (2) and (3), we use a linear approximation for the PL exponent n. For P_0 we use a logarithmic fit for heights below the breakpoint value 0.15 λ and a linear fit for heights above this breakpoint (see Fig. 4). Excellent agreement is obtained, with a maximal and average deviation of only 0.24 and 0.08 dB, respectively. From (2) and (3) it can be seen that brain and muscle tissue only result in small differences and that antenna height is the determining factor. Using these models, we are able to make an accurate estimation of the PL above a flat phantom.

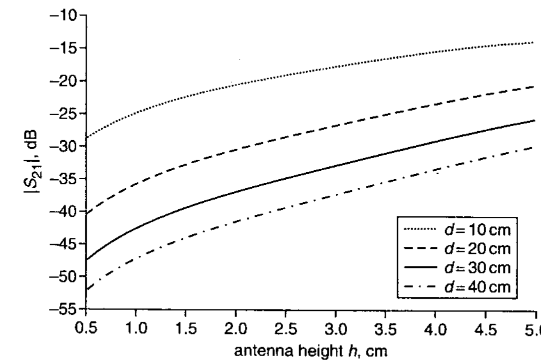

Fig. 3 *Influence of antenna height on PL for several distances (brain tissue)*

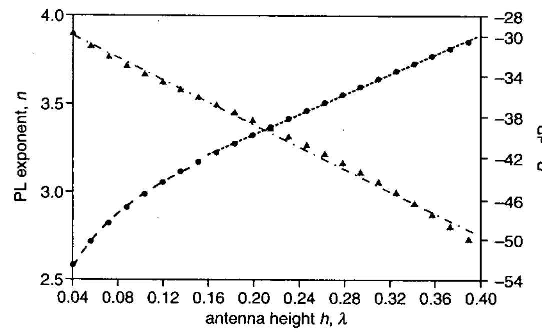

Fig. 4 *Results of fit for n(h) and $P_0(h)|_{dB}$ against antenna height (brain tissue)*

▲ FEKO simulations for n
······ linear fit for n
• FEKO simultions for P_0
- - - logarithmic fit for P_0
. . . . linear fit for P_0

Conclusions: We have presented a new, accurate model for the path loss near a flat and homogeneous phantom for both brain and muscle tissue at 2.4 GHz. The influence of antenna height was characterised and found to be very important. The model has been validated using simulations and measurements for which excellent agreement is reported.

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