***In vivo* and *in situ* measurement and modelling of intra-body effective complex permittivity**

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Radio frequency tracking of medical micro-robots in minimally invasive medicine is usually investigated upon the assumption that the human body is a homogeneous propagation medium. In this Letter, the authors conducted various trial programs to measure and model the effective complex permittivity ϵ in terms of refraction ϵ' , absorption ϵ'' and their variations in gastrointestinal (GI) tract organs (i.e. oesophagus, stomach, small intestine and large intestine) and the porcine abdominal wall under *in vivo* and *in situ* conditions. They further investigated the effects of irregular and unsynchronised contractions and simulated peristaltic movements of the GI tract organs inside the abdominal cavity and in the presence of the abdominal wall on the measurements and variations of ϵ' and ϵ'' . They advanced the previous models of effective complex permittivity of a multilayer inhomogeneous medium, by estimating an analytical model that accounts for reflections between the layers and calculates the attenuation that the wave encounters as it traverses the GI tract and the abdominal wall. They observed that deviation from the specified nominal layer thicknesses due to non-geometric boundaries of GI tract morphometric variables has an impact on the performance of the authors' model. Therefore, they derived statistical-based models for ϵ' and ϵ'' using their experimental measurements.

1. Introduction: Analyses of the interaction between electromagnetic (EM) fields and human tissues typically take an approach in which the field is analysed under the assumption that the propagation medium is homogeneous. Furthermore, much of this work is restricted to the characterisation of human body as a medium in which refraction ϵ' is constant, and some of it further assumes that the absorption ϵ'' is negligible [1]. These simplifying assumptions significantly impact the use of medical micro-robots in minimally invasive medicine for targeted therapy (e.g. drug delivery) and brachytherapy where the knowledge of the exact location of the targeted area is crucial. Radio frequency (RF)-based tracking (navigation and localisation) of robot-assisted endoscopy, such as miniature wireless capsule endoscopy for gastrointestinal (GI) tract diagnosis, is a recent application in which the spatial resolution of the position and orientation highly depends on the model of ϵ' and ϵ'' [2–5].

The question of how variable conductivity (σ) and relative permittivity (ϵ) measurements of human or porcine tissues at single frequencies are and how these variations affect optimal RF communication have been recently addressed in [6–8], among others. Bao *et al.* [6], for instance, remarked that the standard deviations of dielectric values ranged from 4 to 16% for 12 samples of brain tissue. In Schmid *et al.* [7], similarly, it was observed that within 4 h after decease, measured values of both σ and ϵ of pig's tissues decreased by 10 and 4%, respectively. It was also noted in [7, 9, 10] that the dielectric properties for most tissues decrease with increasing age, the trend being generally in line with the decrease in water content of the tissue as the animal ages [7, 9–11]. Xu *et al.* [8] observed in their simulations the significant impact of σ and ϵ variations on RF communication so that higher σ results in an

increment of the maximum SAR value, while higher ϵ causes a decline. Furthermore, it was observed that variations of up to 20% in ϵ and σ , alone or simultaneously, lead to variations of up to 10% in SAR.

The first objective of this Letter was to measure ϵ' and ϵ'' (and their variations) of both the GI tract wall tissues and the porcine abdominal wall under *in vivo* conditions, in a live pig under anaesthetic, [hereafter referred to as dataset-*vivo* (D_v)]. We denote D_{v1} ($D_{v1} \subset D_v$) as measurements of ϵ' and ϵ'' of the wall of GI tract organs (i.e. oesophagus, stomach, small intestine and large intestine) that have been slightly displaced from their normal site towards the exterior of the abdominal wall, and remain in a steady position (Fig. 1a). The measurement process is shown in the video clip D_{v1} uploaded as the multimedia materials under 'supplementary files'. D_{v2} ($D_{v2} \subset D_v$) denotes the measurements of ϵ' and ϵ'' of the porcine abdominal wall in similar experimental conditions. We refer the reader to a video clip D_{v2} in the multimedia materials. The second objective of this Letter was to extend earlier works by Peyman *et al.* [10, 11] to measure ϵ' and ϵ'' (and their variations) of GI tract organs (same organs as in D_v), *in vivo* and *in situ*, i.e. organs located at their normal sites inside the abdominal cavity with the presence of the abdominal wall, hereafter referred to as dataset-*situ* (D_s). Therefore, in this Letter, *in situ* refers to inside a specimen at different anatomical sites along the GI tract of the live pig, with the organs at their normal locations, and in the presence of the porcine abdominal wall (Fig. 1b and video clip D_s in the multimedia materials). The third objective of this Letter was to analyse *in situ* and *in vivo*, the effect of irregular and unsynchronised contractions and simulated peristaltic movements of the GI tract organs inside the abdominal cavity

Figure 1 *In vivo and in situ measurements of dielectric properties of the GI tract*
 a D_v
 b D_s
 c D_m

(Fig. 1c), on the measurements of ϵ' and ϵ'' and their variations, hereafter referred to as dataset-motility (D_m). D_m contains the same information as D_s except for the fact that the porcine abdominal wall was rubbed and kneaded to mimic the GI tract motility. We refer the reader to video clip D_m in the multimedia materials. In this Letter, comprehensive examination of ϵ' and ϵ'' and their variations in different trial programs allows us to scrutinise the effects of both the GI tract and the abdominal wall on intra-body wave propagation during non-invasive *in vivo* and *in situ* procedures, such as wireless capsule endoscopy.

The final and main purpose of this Letter, featuring various experimental programs, refers to advancing the previous analytical models of the interaction between both the EM field and the GI tract tissues and the abdominal wall. These models assume the propagation medium as homogeneous bodies separated by homogeneous films discarding the continuous inhomogeneity. ϵ' and ϵ'' of the GI tract wall tissues (D_{v1}), have been compiled and modelled by 4-pole Cole–Cole equations [10, 11] and further simplified to a 2-pole Debye dispersion model [12, 13]. Firstly, these models only include a conductive loss term to capture the effect of single-layer medium on energy absorption. Secondly, in a multilayer structure, the effects of discontinuity and reflections between the layers are discarded, which leads to shortcomings in modelling dielectric properties of inhomogeneous multilayer media. In light of the above, the fourth specific objective of this Letter was to estimate an analytical model for D_s that accounts for: (i) the effects of lossy (non-magnetic) multilayer structure of the GI tract and the porcine abdominal wall, and (ii) the reflections between the layers. In this Letter, we calculate the attenuation that the wave encounters as it traverses the GI tract tissue and the abdominal wall by obtaining the incident and reflected fields at each interface.

The GI tract organs and the abdominal wall are modelled as dielectric slabs, i.e. lumen, tissue wall, fascia, muscle, fat and skin, with specific thicknesses and dielectric properties. Furthermore, we obtain the overall reflection response recursively using propagation of impedances and reflection responses at the interfaces between slabs. Finally, we show that the estimated total effective permittivity of our model is an accurate approximation of the measured D_s , although sensitive to variations of layer thicknesses.

The results and the model obtained for the steady lossy multilayer structure D_s are arguably inapplicable to the more general case D_m . Firstly, the assumption of dielectric slabs with specific thicknesses must be refuted due to the GI tract motility. Secondly, the simplifying assumption that dielectric properties change discretely at the material interfaces while being constant within each material region is overlooked. Bearing in mind the challenges above, we argued in some of our previous theoretical studies, namely [14, 15], that ϵ' and ϵ'' of inhomogeneous multilayer dielectric media must be treated as random variables with Gaussian probability density function (pdf). This hypothesis is now tested and the statistical properties of D_s and D_m are studied on the ground of our experiment.

In Section 2, we briefly describe the experimental setup of this Letter. In Section 3, we present the performance of the 2-pole Debye dispersion model of [12, 13] tested on our measured D_{v1} . In Section 4, we derive an analytical model of D_s using multilayer dielectric slabs by calculating the incident and reflected fields at each interface. We further study the statistical properties of both D_s and D_m . Our future works and conclusions are briefly summarised and discussed in Section 5.

2. *In vivo* experimental setup and measurements: Different trial programs in this Letter were performed at the Animal Medicine

Figure 2 *Experimental setup to measure ϵ' and ϵ''*

Table 1 Mean and standard deviations of the measurements in D_v , D_s and D_m

Location ^a	D_v				D_s		D_m	
	D_{v1}		D_{v2}		μ_ϵ	σ_ϵ	μ_ϵ	σ_ϵ
	μ_ϵ	σ_ϵ	μ_ϵ	σ_ϵ				
A	n/a ^b	n/a	n/a	n/a	71.53 – 47.19j	1.71	n/a	n/a
B	92.69 – 63.96j	0.40	80.52 – 62.49j	3.25	92.10 – 63.05j	1.36	92.08 – 62.99j	1.00
C	79.22 – 46.30j	0.96	83.21 – 67.46j	0.46	82.15 – 57.49j	0.88	82.63 – 58.36j	0.68
D	77.61 – 38.94j	1.40	83.62 – 66.54j	3.00	77.62 – 39.36j	0.47	78.81 – 40.46j	2.15
E	80.61 – 39.68j	0.42	83.38 – 66.63j	1.12	80.64 – 41.31j	0.35	73.29 – 33.28j	1.04
F	77.10 – 38.06j	0.47	82.46 – 66.87j	1.33	73.46 – 31.96j	0.29	71.88 – 31.83j	0.40
G	65.09 – 20.95j	0.22	85.30 – 69.37j	0.28	67.03 – 23.37j	0.23	66.61 – 23.91j	0.38
H	n/a	n/a	83.65 – 66.12j	0.72	66.74 – 44.34j	0.27	n/a	n/a

^aLocations are identified in Fig. 2.^bNo measurements were taken.

Laboratory, Medical Research Center at the University of Southern Denmark with the approval of Association for Assessment and Accreditation of Laboratory Animal Care International accreditation. The work has been carried out in accordance with the EU Directive 2010/63/EU for animal experiments. The *in vivo* experiment was carried out on a pig of approximately 60 kg (≈ 100 days old) under anaesthetic. Measurements of complex permittivities during different trial programs, i.e. D_v , D_s and D_m were carried out using an N9923A FieldFox vector network analyser and a KeySight Technologies (Agilent) performance dielectric probe kit (85070E) designed for medical applications (Fig. 2). The system provides non-destructive measurements in liquids, semi-solids and flat-surface solids within the frequency range of 10 MHz–50 GHz [16]. Throughout the experiment, the aperture of the probe was held as perpendicular to the tissue surface as possible, leaving no air gaps between the probe and the target tissue. Each dataset was measured twice containing 101 measurements, i.e. D_{v1} , D_{v2} , D_s and $D_m \in \mathbb{C}^{101 \times 2}$. The three-term calibration process was carried out in air, shorting block and water at 25°C and validated by 91% isopropyl alcohol. Details of the precision of such measurement techniques can further be found in [16].

The choice of carrier frequency in capsule endoscopy communication (≈ 434 MHz) is substantially confined by factors, such as high attenuation (above 1 GHz), high bandwidth, antennas effectiveness and transmission power loss due to antenna frequency mismatch. Given the above constraints and bearing in mind that the measurements of $\epsilon'(\omega)$ and $\epsilon''(\omega)$ should be valuable and relevant for future studies on capsule endoscopy localisation and navigation, ω was set to 101 linearly spaced frequencies in the interval [433–434] MHz ISM band. The experiment was run twice and the measurement datasets D_v , D_s and D_m were collected at eight locations (marked by [A–H] in Fig. 2). The corresponding mean and standard deviations of the measurements are reported in Table 1. The *in vivo* and *in situ* conditions of these experiments guaranteed that they were virtually identical to those expected in a human body [17].

3. n -Pole Debye dispersion model of D_v : The n -pole Debye dispersion model of $\epsilon(\omega)$ proposed by Fujii *et al.* [12, 13] at a given angular frequency ω with a conductive loss term is as follows

$$\epsilon(\omega) = \epsilon_\infty + \sum_{i=1}^n \frac{\delta\epsilon_i}{1 + j\omega\delta\tau_i} + \frac{\sigma}{j\omega\epsilon_0} \quad (1)$$

where ϵ_0 denotes the permittivity of vacuum and ϵ_∞ and σ denote the permittivity at infinite frequency and the electrical conductivity at DC, respectively. $\delta\epsilon$ and $\delta\tau$ are the strengths and the relaxation times of the Debye dispersion and $j = \sqrt{-1}$. By

applying the Newton's method and least squares fitting, Peyman *et al.* [10, 11] estimated the parameters of an optimal 2-pole Debye dispersion model for the measured data of [17, 18]. In our study, we compared the modelled $\epsilon(\omega)$ in (1) to the expected value of our measured $\epsilon(\omega)$ in dataset D_{v1} . The results showed that the expected value of the measured values is generally larger than modelled data (Table 2). This difference could be attributed to the fact that unlike the works of [17, 18], our experiment was carried out *in vivo*, including the existence of animal's pulse and respiration. The variation in transmission loss, correlated with heartbeat and proportional to the ventricular volume changes in the heart, has been thoroughly studied by Johnson and Guy [19].

4. Lossy multilayer structure: The general case of an arbitrary number of dielectric slabs, each of them having arbitrary thicknesses, is shown in Fig. 3. This setting consists of M slabs, $M + 1$ interfaces and $M + 2$ dielectric media, including the left and right semi-infinite media η_a and η_b . The incident and reflected fields are considered at the left of each interface. The overall reflection response, $\Gamma_{T_1} = E_{T_1-}/E_{T_1+}$, can be obtained recursively using the propagation of the impedances and reflection responses.

Following the lead of [20, 21], the reflection responses Γ_{T_i} and the wave impedances Z_{T_i} satisfy the following recursions initialised at $\Gamma_{T,M+1} = \rho_{T,M+1}$ and $Z_{T,M+1} = \eta_{T_b}$, respectively,

$$\Gamma_{T_i} = \frac{\rho_{T_i} + \Gamma_{T_{i+1}} \exp(-2j\delta_i)}{1 + \rho_{T_i} \Gamma_{T_{i+1}} \exp(-2j\delta_i)} \quad (2)$$

$$Z_{T_i} = \eta_{T_i} \frac{Z_{T_{i+1}} + j\eta_{T_i} \tan \delta_i}{\eta_{T_i} + jZ_{T_{i+1}} \tan \delta_i} \quad (3)$$

$$i = M, M - 1, \dots, 1$$

In (2) and (3), the complex propagation phase thickness δ_i of each layer is defined as $\delta_i = l_i \omega / c_0 \sqrt{n_i^2 - n_a^2 \sin^2 \theta_a}$, $i = 1, 2, \dots, M$ where the refractive index, $n_i(\omega) = \sqrt{\epsilon_i(\omega)/\epsilon_0}$, θ denotes the

Table 2 2-pole Debye dispersion model versus measurements in D_{v1}

$E[\epsilon(\omega)]^\dagger$	2-pole Debye model	Measure
Stomach	66.83 – 42.89j	92.69 – 63.96j
Small intestine	64.01 – 81.33j	79.22 – 46.30j
Colon	61.35 – 37.58j	80.61 – 39.68j
Spiral colon	61.35 – 37.58j	77.10 – 38.06j

[†] $E[\cdot]$: expected value.

Figure 3 Lossy multilayer dielectric slab
a general case with $M+2$ layers
b porcine abdominal wall

incident angle and $\eta_{T_i} = \eta_0/n_{T_i}$ (η_0 : characteristic impedance of vacuum). Furthermore, we define the transverse reflection coefficients at the $M+1$ interfaces as $\rho_{T_i} = (n_{T_{i-1}} - n_{T_i})/(n_{T_{i-1}} + n_{T_i})$ for $i = a, 1, 2, \dots, M, b$ initialised at $n_{T_0} = n_{T_a}$ and $n_{T_{M+1}} = n_{T_b}$. The transverse refractive indices in each medium are defined by $n_{T_i} = n_i \cos \theta_i$ for the perpendicular s -polarisation mode. Given (2) and (3), the total effective permittivity of the lossy multilayer structure in terms of the input impedance Z can be readily derived as $\epsilon_{\text{eff}} = (\eta_a/Z)^2$.

4.1. Multilayer Model-VH and Model-AW of D_s : We investigate the performance of (2) and (3) for estimating the total effective permittivity of the inhomogeneous dielectric media D_s using two different multilayer structures. The first structure, hereafter referred to as Model-VH, is based on the cross-sectional layer model (transverse slice $a-vml610$) of the Visible Human Project, NIH [22], comprised of six sequential slabs, i.e. lumen, tissue wall (e.g. stomach or intestine), fascia, muscle, fat and skin. The thickness and dielectric properties of each slab are reported in several studies, including [21, 23]. These values (except ϵ' and ϵ'' of the tissue wall obtained from D_{v1}) are used in this Letter to derive the total effective ϵ' and ϵ'' of the multilayer structure. The second structure is a 3-layer structure consisting of the lumen, tissue wall (e.g. stomach or intestine) and the porcine abdominal wall, hereafter referred to as Model-AW. ϵ' and ϵ'' of the lumen and the thicknesses of the slabs in Model-AW are equal to those of Model-VH, while ϵ' and ϵ'' of the tissue wall and the porcine abdominal wall are set based on our measurements D_{v1} and D_{v2} , respectively. The performance of Model-VH and Model-AW in estimating the total effective ϵ'

and ϵ'' of the portion of D_s corresponding to stomach data is presented in Figs. 4a and b. The superior performance of Model-AW can be readily inferred from the analysis of our results. We, however, observed that deviation from the specified nominal layer thicknesses has an impact on the performance of the model and therefore, sensitivity analysis of both Model-VH and Model-AW are performed in the following section.

4.2. Sensitivity analysis of Model-VH and Model-AW: The first-order sensitivity of ϵ_{eff} in the multilayer model to the perturbations of the layer thickness l_i is evaluated as the ratio of the change in ϵ_{eff} to the change of l_i for a small incremental variation from the nominal value and is computed as $S_{l_i}^{\epsilon_{\text{eff}}} = (\partial \epsilon_{\text{eff}} / \partial l_i)(l_i / \epsilon_{\text{eff}})$. Recall from (2) and (3) that the perturbation of l_i leads to a variation of δ_i which in turn will affect Z_i . Variations of Z_k will result in changes in Z_{k-1} for $k = i, i-1, \dots, 2$. Finally, perturbations in Z_1 affect ϵ_{eff} . By computing $S_{l_i}^{\delta_i}, S_{\delta_i}^{Z_i}, S_{Z_i}^{Z_{i-1}}, \dots, S_{Z_2}^{\epsilon_{\text{eff}}}$ and $S_{Z_1}^{\epsilon_{\text{eff}}}$ and invoking the chain rule in sensitivity analysis, $S_{l_i}^{\epsilon_{\text{eff}}}$ can be calculated

$$S_{l_i}^{\epsilon_{\text{eff}}} = S_{Z_1}^{\epsilon_{\text{eff}}} S_{Z_2}^{Z_1} S_{Z_3}^{Z_2} \dots S_{Z_{i-1}}^{Z_{i-2}} S_{\delta_i}^{Z_i} S_{l_i}^{\delta_i} \quad (4)$$

given

$$S_{Z_1}^{\epsilon_{\text{eff}}} = -2 \quad (5)$$

$$S_{Z_{k-1}}^{Z_k} = \frac{\eta_{k-1} Z_k}{\cos^2 \delta_{k-1} (\eta_{k-1} + j Z_k \tan \delta_{k-1}) (Z_k + j \eta_{k-1} \tan \delta_{k-1})} \quad (6)$$

$$S_{\delta_k}^{Z_k} = \frac{j \delta_k (\eta_k^2 - Z_{k+1}^2)}{\cos^2 \delta_k (\eta_k + j Z_{k+1} \tan \delta_k) (Z_{k+1} + j \eta_k \tan \delta_k)} \quad (7)$$

Figure 4 Analytical model of total effective $\epsilon'(\omega)$ and $\epsilon''(\omega)$ in a lossy multilayer structure

Figure 5 Magnitude of S_i^{eff} for Model-AW and Model-HV (fat and muscle thickness variation)

$$S_{l_i}^{\delta_k} = 1 \quad (8)$$

$$S_{l_i}^{\epsilon_{\text{eff}}} = \frac{-2j\delta_i(\eta_i^2 - Z_{i+1}^2)}{\eta_i Z_{i+1}} \times \prod_{k=1}^i \frac{\eta_k Z_{k+1}}{\cos^2 \delta_k (\eta_k + jZ_{k+1} \tan \delta_k)(Z_{k+1} + j\eta_k \tan \delta_k)} \quad (9)$$

The sensitivity of ϵ_{eff} to the perturbations of l_i for the Model-AW and Model-VH are presented in Figs. 5a–c, respectively. It can be observed that generally, the magnitude of $S_{l_i}^{\epsilon_{\text{eff}}}$ is smaller in the Model-AW compared to that of the Model-VH. Furthermore, variations of the thickness of muscle results in a larger $S_{l_i}^{\epsilon_{\text{eff}}}$ compared to that of fat which can be contributed to the significant difference in the magnitude of the tissues' ϵ .

4.3. Statistical model of D_s and D_m : Under the null hypothesis (H_0), the sample measurements of ϵ' and ϵ'' in D_s and D_m are normally distributed, $\mathcal{N}(\mu_{\epsilon'}, \sigma_{\epsilon'})$ and $\mathcal{N}(\mu_{\epsilon''}, \sigma_{\epsilon''})$, while H_1 represents the alternative hypothesis. To accept or reject H_0 , we derive the hypothesis testing decision values following the Kolmogorov–Smirnov (K–S) test statistics [24]

$$D_n = \sqrt{n} \sup_{x \in \mathbb{R}} |(F_n(x) - G(x))| \quad (10)$$

where $F_n(\cdot)$ is the empirical CDF of the analysed sample and $G(\cdot)$ is the CDF of a Gaussian random variable. To test H_0 , the decision rule is considered as following

$$\delta = \begin{cases} H_0: & D_n \leq c \\ H_1: & D_n > c \end{cases}$$

The threshold c depends on the significance level α found from the

condition $\alpha = \mathbb{P}(D_n \geq c | H_0)$. A p -value larger than the critical value of $\alpha = 0.05$, i.e. the probability to have a D_n -value less or equal than the one obtained if $F_n(\cdot)$ and $G(\cdot)$ were sampled from identical populations (below the critical maximal D_n -value of 0.136) in the K–S test is referred to as failing to reject H_0 .

These results (Table 3) confirm our previous theoretical studies [14, 15], as the hypothesis that ϵ' and ϵ'' of D_s are normally distributed cannot be rejected. Unlike in D_s , the results for D_m showed that the null hypothesis (normal distribution) should be rejected. The distribution functions for ϵ' and ϵ'' in D_m will be presented in our future work.

5. Future works and concluding remarks: The significant impacts of variations of complex permittivity and conductivity on intra-body wave propagation measures (such as specific absorption rate, time of flight and strength attenuation) have been emphasised in the literature but inadequately addressed. This is the first study to present *in vivo* and *in situ* measurement and modelling of effective complex permittivity of the GI tract organs and the porcine abdominal wall. To perform precise measurements of ϵ' and ϵ'' , the experimental conditions in our trials were closely observed, however, few sources of system drift and measurement inaccuracy associated with instrumentation and procedure were still present. The primary source of measurement uncertainty was the instability of the RF cable which was occasionally flexed between the calibration and the measurement process, due to the practical difficulties of the *in vivo* and *in situ* experiment. Features, such as automated electronic calibration refreshing, that recalibrate the system before each measurement to avoid system drift errors are not applicable in trials where the subject, in this case the pig, is immobilised. The secondary source of inaccuracy was the potential formation of air gaps between the probe and the tissue wall, in particular during the course of D_m measurements. The final cause of imprecision is related to the fact that the coaxial probe method was deployed in

Table 3 Statistical model and H_0 testing of D_s and D_m

Location	D_s						D_m					
	ϵ'			ϵ''			ϵ'			ϵ''		
	D -value	p -value	H_0 -reject	D -value	p -value	H_0 -reject	D -value	p -value	H_0 -reject	D -value	p -value	H_0 -reject
A	0.10	0.32	no	0.10	0.24	no	n/a	n/a	n/a	n/a	n/a	n/a
B	0.13	0.08	no	0.10	0.22	no	0.16	0.01	yes	0.19	0.01	yes
C	0.10	0.30	no	0.10	0.22	no	0.24	0.01	yes	0.16	0.02	yes
D	0.08	0.47	no	0.08	0.50	no	0.21	0.01	yes	0.24	0.01	yes
E	0.08	0.58	no	0.07	0.69	no	0.27	0.02	yes	0.28	0.01	yes
F	0.09	0.38	no	0.08	0.48	no	0.15	0.02	yes	0.10	0.44	no
G	0.07	0.67	no	0.07	0.72	no	0.14	0.04	yes	0.07	0.70	no
H	0.10	0.31	no	0.11	0.20	no	n/a	n/a	n/a	n/a	n/a	n/a

inhomogeneous media, where the measured ϵ' and ϵ'' are average values weighted by the intensity of the E -field, being highest at the centre conductor of the probe tip. The necessary condition for minimum semi-infinite sample thickness of $20/\sqrt{|\epsilon_{\text{eff}}|}$ mm to avoid commensuration was fulfilled for all measurements of D_v , D_s and D_m .

Overall, our study shows that the simplifying assumption of discrete dielectric change at the material interfaces while being constant within each material region, needs to be revisited by a more realistic scenario of continuous variation of dielectric permittivities across phase interfaces. This can be inferred by the superior performance of the Model-AW compared to the Model-VH which can be correlated to (i) fewer number of slabs and thin intermediate layers; (ii) more realistic layered model of the propagation medium; (iii) smoother variations of ϵ' and ϵ'' at the interfaces between layers (Table 1); and (iv) smaller magnitude of S_{ii}^{eff} . In light of the above, our next study is targeted at measuring s_{11} and s_{12} scattering parameters of intra-body propagation *in vivo* and *in situ*, to measure and derive a mathematical model for path-loss exponent in the Friis channel model.

Large inter-individual variability and non-geometric boundaries of GI tract morphometric variables in a human population pose a challenge to the application of ray tracing techniques outlined in Section 4. Moreover, visceral motility and irregular and unsynchronised contractions of the GI tract hinder the analysis of transmitted and reflected waves in intra-body propagation. Both for these reasons and also bearing in mind other relevant variables, such as peristaltic movement and body motion, dielectric properties at a certain frequency must be regarded as functions of aforementioned variables rather than constant values. To cope with the shortcoming of ray tracing techniques and to avoid complex anatomical and expensive numerical models, in our study, we derived the statistical models of ϵ' and ϵ'' where the hypotheses of the Gaussian pdf for these variables were not rejected. Future work will include conducting a more extensive study on the statistical properties of ϵ' , ϵ'' and path-loss exponent for precise intra-body localisation of a wireless endoscopic capsule.

6. Multimedia materials as supporting document: Video clip D_{v1} demonstrates the procedure to measure ϵ' and ϵ'' of the wall of small intestine (duodenum) and large intestine that have been slightly displaced from their normal sites towards the exterior of the abdominal wall, and remain in a steady position. Video clip D_{v2} presents the procedure to measure ϵ' and ϵ'' of the porcine abdominal wall. Video clip D_s shows the process to measure ϵ' and ϵ'' of the organs located at their normal sites inside the abdominal cavity with the presence of the abdominal wall. Video clip D_m demonstrates the procedure to collect the same information as in D_s except for the fact that the porcine abdominal wall was rubbed and kneaded to mimic the GI tract motility.

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8 References

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