On the UWB medical radars working principles

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Abstract: In the last 20 years, a lot of research activity has been carried out on UWB radar in medicine. Over the years, many implementations of medical UWB radars have been presented in literature. An interesting medical application for these devices is the wireless monitoring of vital signs. Heart and breath rate were successfully detected with such devices but in spite of the many working examples presented, the authors are still convinced that a solid explanation of the operation of such radars is lacking. UWB radar output signals are indubitably correlated with the respiration and heart activities but where do they come from? The classic explanation as per the McEwan’s patent of 1996 proposes that the signal is due to deep echoes reflections from the heart wall and blood. This fact does not seem a realistic explanation of the phenomenon.

Keywords: medical radar; UWB; attenuation model; microwaves; human tissues; HT; FDTD.


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1 Introduction

In a classical medical UWB radar application (Staderini, 2002), repetitive UWB pulses are generated and sent toward the body in which electromagnetic (EM) energy propagates to reach an internal organ like the heart. While trespassing each tissue interface, part of the radiated energy is reflected backward and so it can be detected from the outside. It has been already proven that such medical radars are able to detect internal organ movements using very short UWB pulses (Staderini, 2001; McEwan, 1996, 1998; Varotto and Staderini, 2007b) but there is not yet an exhaustive description of the interaction which makes possible to retrieve a simple heart-rate correlated signal.

Many scientists claim to have developed a UWB radar prototype which can detect both respiration and heart rate using very short EM pulses (100 to 400 ps) emitted from a broad-band antenna. Most of these systems embed fast samplers or low noise correlators in order to detect the back scattered echoes in a synchronous way, so generating a low frequency signal for the processing unit.

In this paper, we want to analyse all the possible physical mechanisms which may be implicated in this phenomenon in order to better understand it and to give a contribution to the amelioration, or fundamental modification, of UWB radar design. We think that all the possible interactions may be ranked as in the block diagram of Figure 1. There are three main hypotheses to analyse:

- Far field operation. This case concerns two different options: deep echoes coming from the lung-heart wall (the classic explanation) and early echoes from the skin, modulated by the superficial perfusion and/or skin elastic displacement.
- Natural emission of the human skin considered as a black body radiator and modulated by the peripheral perfusion (both in far and near field).
• Near field disturbance (NFD) or the modulation of the near field due to the direct coupling between the antenna and the skin and/or deeper tissues.

**Figure 1** Block diagram of the UWB radar’s signal possible generating mechanisms

![Block diagram of the UWB radar’s signal possible generating mechanisms](image)

### 2 Far field operation

In this paragraph, we consider the classical explanation of UWB radar working principle. A short EM pulse is sent to the target and a backward echo is detected into a very narrow time-window exactly positioned in regard to the time of flight required by the EM pulses to cover the return path between the pulse emitter (typically the Tx antenna) and the target (heart or other internal moving organ).

This is the case of far field working condition, which means that the target is positioned at a distance from the Tx module where the EM field can be considered completely decoupled from the antenna. Only in this situation, it is possible to simply consider pulse propagation and return echoes.

The radiated pulse will suffer a geometrical attenuation due to the propagation in the air before reaching the body and a further attenuation due to energy adsorption in the biological tissues considered as dispersive dielectric materials.

As a first step to evaluate the deep echoes hypothesis, we use a model (Varotto and Staderini, 2007a) to estimate the transmission and reflection components of a generic EM wave propagating into a multilayer dielectric material. Such a model becomes of crucial importance for the prediction of echoes magnitudes coming from deep human tissues (HT) interfaces and therefore to evaluate the required performance for a real pulse echo detection system.

#### 2.1 Propagation multilayer model

Our proposed model is a simple transmission line based, time domain reflectometry mutated, multilayer dielectric sandwich (Christ et al., 2006; Bilich, 2006; Varotto and Staderini, 2007a). Using Gabriel et al.’s (1996a, 1996b, 1996c) fitting functions for the relative dielectric permeability $\varepsilon'_r$ and the electric conductivity $\sigma$ of HT (http://www.fcc.gov/fcc-bin/dielec.sh) it is possible to express the complex dielectric permittivity $\varepsilon_r$ as in (1) and any further electric parameter related to each biological layer, like the characteristic impedance $Z_n$, the reflection coefficient $\rho_n$ and the adsorption rate $\alpha$ in the broad frequency range 10 Hz to 20 GHz.

$$\varepsilon_r = \varepsilon'_r - j\frac{\sigma}{\varepsilon_0 \omega} \quad (1)$$

From the definition of complex refraction index $n'$, we can link $\varepsilon_r$ to the attenuation factor $\alpha$ (where $c$ is the speed of light in vacuum and $\omega = 2\pi f$):

$$n' = n - j\frac{\alpha \cdot c}{2\omega} \equiv \sqrt{\varepsilon_r} \quad (2)$$

By further developing (2) we can found the direct expression (3) for $\alpha$:

$$\alpha = -\frac{2\omega \cdot \Im\sqrt{\varepsilon_r}}{c} \quad (3)$$

Furthermore, the real part $n$ of the complex refraction index $n'$ can be expressed as a function of the phase velocity $v_k$ of the EM wave propagating into the specific layer $k$:

$$n = \frac{c}{v_k} \equiv \Re\sqrt{\varepsilon_r} \quad (4)$$

Using (4) we can calculate $v_k$ in each different layer and therefore predict the return echo delay.

The characteristic impedance $Z_k$ and the reflection coefficient $\rho_k$ relative to the interface between dielectric $k$ and $k + 1$ are expressed by the following (5).

$$Z_k = Z_0 \sqrt{\varepsilon_r} \quad \rho_k = \frac{Z_k - Z_{k+1}}{Z_k + Z_{k+1}} \quad (5)$$

Using geometric information from the ‘Visible Human Project’ (Ackerman, 1998; http://visiblehuman.epfl.ch), we can define a sandwich-model in the thorax zone following a direct path from the skin to the inner heart wall as in Table 1.

<table>
<thead>
<tr>
<th>Human tissues sandwich model</th>
</tr>
</thead>
<tbody>
<tr>
<td>Skin</td>
</tr>
<tr>
<td>1.3</td>
</tr>
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</table>

Note: Each layer thickness is reported in mm.

This simple model considers each layer as a transmission line with proper characteristic impedance and attenuation factor. Therefore, it is possible to calculate reflections at each interface and attenuations due to EM propagation into the dielectric. For this model, we do not take into account the geometric attenuation due to free space propagation; that
is why this model is more similar to standard time-domain cable reflectometry.

2.2 Multilayer model results

An EM wave sent toward the heart crosses all the layers and the interfaces indicated in Table 1. At each interface, a reflection occurs because of the impedance mismatch between the two adjacent tissues. In this very simple case, we can calculate the echo magnitude coming from each inner interface in the frequency range 10 Hz to 20 GHz [Figure 2(b)].

**Figure 2** (a) Inner interfaces echo power attenuation at 3.1 GHz (b) echo power attenuation over the UWB spectrum (see online version for colours)

EM wave propagation velocity is different in each layer (4) and it may vary with frequency as well. Therefore, the return echo delay does not depend on the layer geometric configuration only. Figure 3(a) shows the delay of return echoes from inner tissue interfaces corresponding to the model of Table 1. We can notice how echoes from air-skin, skin-fat and fat-muscle (early interfaces) falls into a time window of 250 ps, while deep echoes are much more delayed. As shown in Figure 3(b), early echoes delays are not very much influenced by frequency, while deeper echoes are affected by quite linear variations of up to 250 ps over the UWB spectrum.

**Figure 3** (a) Inner interfaces echo return delay at 3.1 GHz (b) return delay variation over the entire UWB spectrum (see online version for colours)

Also from Figure 2(a), we can remark how the heart-blood interface echo is attenuated of 93 dB still at 3.1 GHz while Figure 2(b) shows how drastically the attenuation rises with frequency (especially for deep return echoes). To have an idea of the receivable power coming back from the heart-blood interface we have to consider the FCC –41 dBm/MHz limit (0.5 mW of maximum power over the whole UWB spectrum). Given the average attenuation of 96 dB [Figure 2(b)] the expectable return power is –99 dBm in the most optimistic prediction. Therefore, it seems not so reasonable to expect observable UWB echoes from deep tissues interfaces.

This first result does not agree with the well-known classical explanations (McEwan, 1996, 1998) where the inventor claims to be able to monitor the heart activity thanks to the impedance mismatching between heart muscle and blood.
What would be more reasonable is detecting the lung-heart interface (as the heart left ventricle is in part covered by the inferior lobe of left lung). At 3.1 GHz it generates a weak –60 dB (–63 dBm) echo [Figure 2(a)], even if the attenuation rapidly increases to prohibitive levels at frequencies higher than a few GHz.

It may be interesting to study how much the lung-heart echo attenuation varies when the heart moves modifying the model geometry. Supposing that a displacement $ds$ of the heart wall would be mainly adsorbed by the lung elasticity, then the lung layer would change its thickness by $ds$ for each heart beating.

**Figure 4** Dynamic power attenuation of heart-lung interface, (a) absolute variation (b) differential variation (see online version for colours)

![Figure 4](image)

Figures 4(a) and 4(b) plot the heart-lung attenuation dynamic as a function of frequency in the case $ds = 6$ mm.

This effect could be exploited to detect the heart rate at UWB frequencies. However, Figure 4(a) and 4(b) show that while the dynamic increases, the average attenuation steps up as well and therefore a trade-off should be searched.

Figure 5 shows a simulation obtained with Remcom XFDTD 3D EM analysis software. A numeric model of a double-ridged horn antenna (Burns et al., 2003; Kujalowicz et al., 2006) has been designed in order to simulate the radiation of UWB pulses. A false colour scale is used to show magnitudes. From red to dark blue the scale represents a 35 dB dynamic range. The antenna has been excited with a Gaussian derivative signal (400 ps duration and 1 V amplitude) and the short red pulse of Figure 5(a) [5(d)] is radiated in air, followed by a weaker oscillating tail (in green).

In Figure 5(b) [5(e)], the pulse reaches a 50 × 50 cm plane made of two simulated layers: a skin layer of 0.5 cm thickness and a fat layer of 10 cm thickness. Most of the energy is reflected back and only a little part propagates inside the obstacle.

In Figure 5(c) [5(f)], we can easily identify the early echo from the skin and a secondary echo from the following tail reflection.

**Figure 5** Remcom XFDTD simulations [in (d), (e) and (f) images of horn and tissues were removed for clarity purposes] (a) and (d) ridged horn antenna emitting a Gaussian derivative pulse (400ps) (b) to (c) and (e) to (f) pulse rebounds and penetrate on a plane skin layer (10 mm) followed by a thicker fat layer (100 mm) (see online version for colours)

![Figure 5](image)

No evidence for the skin-fat echo can be seen in Figure 5(c) [5(f)], even if the expected echo attenuation calculated with our model is about 25 dB in average over the UWB spectrum. However, the predicted return echo delay is in the order of 250 ps and therefore, we can expect that the skin-fat interface echo is confused into the stronger echo associated to the main pulse’s tail.

### 2.3 EM echoes from deep Interfaces

From the considerations in the previous paragraph it seems not very realistic to get echoes reflected from deep tissues, with the FCC limitation of –41.3 dBm/MHz over the UWB spectrum that is a maximum power of 0.5 mW.

As a matter of fact, in the optimistic case of 100 dB of return echo attenuation, the system should be able to resolve power levels in the order of 0.05 nW, which is not far from
the thermal noise power (kT) at $T = 310\,^\circ\text{K}$ in the same band (0.03 nW). However, if the emitted pulse is not narrow enough, then it is possible that those deep return echoes are superimposed on stronger ones coming from more superficial earlier tissue interfaces. So the possibility to time-resolve the echoes will be strongly impaired as well.

The received signal should be sampled into a very stable, narrow and synchronous time window in order to enhance the incoming signal to noise ratio (SNR) by time averaging techniques (process gain). In this case, the jitter noise becomes a critical problem because the sampling window must be more stable than the echo duration. This means that using pulses of 100 ps duration, we should have the jitter uncertainty less than 10 ps, a quite restrictive limit, indeed.

The situation becomes even more complicated if we consider that the target should be placed at a fixed distance from the radar. The least parasitic movement between the radar and the subject under test would lead to the migration of the target outside the sampling window. Even the breath of the patient would easily saturate the front-end radar and the subject under test would lead to the migration from the radar. The least parasitic movement between the operation as reported in the literature.

Explanations should be investigated for actual UWBsimplicity and low power consumption; therefore, different echoes processing without losing its main advantages of medical radar can hardly work on the principle of deep

indeed.

The amplitude of an EM pulse reflected by the skin surface may vary as much as 4.8% over the UWB spectrum between the two previous extreme cases. This important result might finally give an explanation of the working principle of UWB medical radar operation in the far-field mode.

If the early echoes are directly modulated by the blood perfusion then we can figure out to simply integrate the incoming pulses into a temporal window wide enough to include at least one pulse. The sampling window position is no more a critical parameter because deep echoes are negligible if compared to early ones. The system can simply integrate all received echoes as often proposed in the literature (McEwan, 1996, 1998).

The reflection coefficient only depends on the characteristic impedance of the layer. All those parameters can be calculated using the propagation multilayer model. Figure 6(a) shows the characteristic impedances of a skin and blood layer respectively. The variation of reflection coefficient between the previously mentioned cases is illustrated in Figure 6(b).

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The target displacement may result in a sudden change in amplitude of the incoming signal, because reflections occur at a different distance and thus the attenuation path changes. This is probably the trivial explanation why UWB radars are able to detect the respiration rate so simply. During respiration, the thorax changes its volume and the distance between radar and thorax varies depending on the breathing rate. If the patient were obliged to push against a panel placed in front of the radar (constraining the distance between radar and patient), the respiration rate would

2.4 EM echoes from superficial layers

Tissues perfusion through the capillary network is certainly correlated with the heart rate. Each time heart pumps on, arterial dilatation occurs and the vascular network receives new oxygenated blood. This mechanism takes place into tissues like skin and muscles as well. A further consequence is the cyclic variation of local blood concentration in the tissue volume. If skin would be transparent (to visible light) this phenomenon would appear as cyclic colour changing between two red tones at the heart rate frequency.

If the electric properties of the skin would significantly vary with the perfusion cycle, we could obtain a heart rate correlated signal by simply analysing the echoes coming from the tissues on the body surface (like the skin).

Echoes coming from early reflections on the skin-fat-muscle interfaces can be easily processed by UWB radars. From Figure 2(b), we can see that the air-skin echo has a quite stable magnitude of about –6 dB over the entire spectrum, while the skin-fat and fat-muscle echoes are over –20 dB into the UWB range.

Even if previous considerations regarding the averaging and sampling window are still valid for whatever far field hypothesis, the much more important SNR of such early echoes would open more possibilities from the design point of view.

To understand if the perfusion can really modulate early return echoes we can simulate two extreme cases:

- EM pulse reflection on the skin surface
- EM pulse reflection on a pure blood surface.

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Figure 6 (a) Characteristic impedance module of skin surface in back and blood surface in red (b) difference between the blood and skin reflection (see online version for colours)

2.5 EM echoes modulation by skin elastic displacement

The target displacement may result in a sudden change in amplitude of the incoming signal, because reflections occur at a different distance and thus the attenuation path changes. This is probably the trivial explanation why UWB radars are able to detect the respiration rate so simply. During respiration, the thorax changes its volume and the distance between radar and thorax varies depending on the breathing rate. If the patient were obliged to push against a panel placed in front of the radar (constraining the distance between radar and patient), the respiration rate would
probably not be visible because the skin surface does not move.

The effect of target displacement which makes possible to monitor the respiration activity with UWB radars could give an alternative explanation on how the UWB radar retrieves a heart rate correlated signal in far field working conditions.

We wanted theoretically verify if the elastic skin deformations induced by the heart mechanical activity could modulate early echoes coming from the skin surface. The first point to investigate is the skin displacement dynamic range and if it is wide enough to produce an effect on the return echoes.

\[ \frac{(R + d)^4}{R^4} - 1 \approx \frac{4d}{R} \]  

Now assuming \( R = 10 \text{ cm} \) we get a weak variation of 0.24% at receiver side. Even considering the phase modulation contribution, we find a maximum shift of \( \frac{d}{c} = 0.2 \text{ ps} \) which is normally negligible.

Finally, we can simply discard the hypothesis of a contribution of the skin displacement induced by the cardiac activity to the modulation of the early echoes. In the best case, this effect can be considered as a minor contribution to the much more important one induced by the skin perfusion. Nevertheless, at the recent ‘2nd International MELODY Workshop’ held at the Oslo University Hospital on 17–18 October 2011, Svein-Erik Hamran showed an FMCW radar being able to detect heart rate in the far field region. He reported an estimation of a few tens of a millimeter for the movement of the target which is in accordance with these findings.

### 3 Natural radiating effect

The thermal radiation from objects strongly depends on their temperature gradient with respect to the ambient temperature and their emissivity coefficient. The human body can be considered as a black body and its radiation power can be calculated over the entire spectrum. Whenever the peripheral blood perfusion could modulate the skin temperature then the radiated power would be modulated as well following the heart rate period. We wanted to theoretically investigate if the black body emitted power itself could be detected by UWB radar receivers (no matter the echoes) as already studied by TimeDerivative, Inc. (2005).

#### 3.1 Human radiation model

The human body can be approximated to a black body emitter across all frequencies, including the UWB spectrum. The human body radiates at its temperature \( T_b \) and absorbs at the ambient temperature \( T_a \).

The total radiated power can be predicted using the Stephan-Boltzmann’s law for a black body radiator (7).

\[ P_{\text{tot}} = \sigma A \epsilon \left( T_b^4 - T_a^4 \right) W \]

where \( \sigma \) is the Stephan-Boltzmann’s constant, \( A \) is the emitting area which is close to \( 1 \text{ m}^2 \) for the combined torso and head body parts and the emissivity of the human body is known to be \( \epsilon = 0.98 \) (Dziuban, 2002).

At the ambient temperature of \( T_a = 288.15 \text{ K} \) (15°C) and the body temperature of \( T_b = 310.15 \text{ K} \) (37°C) the total power body emission is 131 W distributed over the entire spectrum.

The black body spectral radiation power distribution at given temperature \( T \) is predicted by the Plank’s law (8).
where $h$ is the Planck’s constant, $K$ the Boltzmann’s constant and $c$ the speed of light. The radiation peak can be calculated with the Wien’s law (9).

$$f_{\text{max}} = \frac{\alpha h}{K T} \text{ Hz}$$  \hspace{1cm} (9)

where $\alpha$ is a constant. The peak radiation frequency occurs close to 32 THz.

The radiated power spectral density (PSD) from the human body can be calculated as follow:

$$P_b(f) = A [P(T_h, f) - P(T_a, f)] \text{ W/Hz}$$  \hspace{1cm} (10)

The curve related to (10) is shown in black in Figure 8. The human PSD can be linearly approximated in a large frequency band below the peak frequency by the following (11):

$$P_b(f) = 20 \log(f_{\text{GHz}}) - 106.7 \text{ dBm/MHz}$$  \hspace{1cm} (11)

Figure 8 also shows the UWB interval and the FCC outdoor mask which regulates the emission limit. The average emission level for a human body in the UWB is –90.75 dBm/MHz; well below the –41.3 dBm/MHz FCC limit. The total power emitted in the UWB is the blue area in Figure 7 and it results about 8.2 nW (Table 2), still in the limit. The total power emitted in the UWB is the blue area in Figure 7 and it results about 8.2 nW (Table 2), still in the limit. The total power emitted in the UWB is the blue area in Figure 7 and it results about 8.2 nW (Table 2), still in the limit.

**Table 2** Total emitted power in the UWB spectrum at different body temperatures

<table>
<thead>
<tr>
<th>Temperature 1</th>
<th>Temperature 2</th>
<th>Power (nW)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$T_a = 15^\circ C$</td>
<td>$T_b = 36.9^\circ C$</td>
<td>$P_{\text{ewb}} = 8.182048$ nW</td>
</tr>
<tr>
<td>$T_a = 15^\circ C$</td>
<td>$T_b = 37^\circ C$</td>
<td>$P_{\text{ewb}} = 8.219398$ nW</td>
</tr>
<tr>
<td>$T_a = 15^\circ C$</td>
<td>$T_b = 37.1^\circ C$</td>
<td>$P_{\text{ewb}} = 8.256759$ nW</td>
</tr>
</tbody>
</table>

### 4 Near field operation

In this last paragraph, we want to analyse the hypothesis of NFD to explain the UWB medical radar operation in the case of strong proximity between receiving UWB antenna and moving target. According to ATIS Telecom Glossary (2010):

“A restricted radiation device which establishes a radio frequency field in its vicinity and detects changes in that field resulting from the movement of persons or objects within the radio frequency field can be defined as a field disturbance sensor (FDS).”

Due to this definition, even an UWB radar can be classified as a FDS if the moving target is sufficiently close to the receiving antenna.

The field disturbance effect is well known even for microwave systems. In the document 15.503 (subpart g), the FCC sets out a few regulations for unlicensed UWB transmission systems. The document gives definitions of basic UWB parameters and UWB complex systems as medical imaging system, which is referred as: “a field disturbance sensor that is designed to detect the location or movement of objects within the body of a person or animal”.

The working principle of a FDS is a direct coupling between the receiving antenna and a dielectric object which behaves as a load like a transformer secondary winding. The effect of this direct coupling is a variation of the antenna parameters as the impedance or gain. If the coupled object moves (or it changes its dielectric properties) then a modulation of the receiving antenna parameters occurs and the front-end electronics will acquire a signal which is directly correlated with the geometry variation of the target.

Before going on into this discussion, we will introduce the concept of near field coupling in a more precise way.

### 4.1 Field energy theory

In this paragraph, we want to introduce a short analysis of EM energy transfer in order to better understand the
concepts of near and far field around a simple dipole antenna excited with a transient time domain signal like a Gaussian pulse.

The fundamental law of EM energy transfer and conservation follows directly from Maxwell’s equations and it is referred as the Poynting’s theorem (12)

$$\nabla \cdot S + \frac{\partial}{\partial t}(u_E + u_H) + E \cdot J = 0$$

(12)

where $S = E \times H$ is the Poynting vector product representing the EM energy flow trough a particular point. This relation establishes that the divergence of $S$, the time rate of change of local EM energy density $u_E + u_H$ and the ohmic losses $E \cdot J$ must balance out.

If an object is placed at a given distance from the emitting antenna, it adsorsb and reflects EM energy modifying the Poynting’s theorem balance calculated for a free space configuration. That happens because the EM waves do not owe their formation solely to processes at the origin, but arise out of the conditions of the whole surrounding space which is the true seat of the EM energy.

**Figure 9** Field impedance of the time-harmonic dipole fields as a function of the range units of $k \cdot R$, where $R$ is the radial distance and $k = 2\pi / \lambda$ is the wave number (see online version for colours)

Poynting’s theorem can describe the energy balance for whatever spatial configuration; however, the effect of placing an obstacle into an EM field can be very different depending on the distance from the excited antenna. An object placed in the far-field may interact with the antenna only thanks to EM wave reflection coming back to the antenna as return echoes. This is the classical far field approach on which the radar equation (http://earth.esa.int/applications/data_util/SARDocs/spaceborne/Radar_Course/Radar_Course_III/radar_equation.htm) is based.

If the same object is placed into the near-field zone then the interaction with the antenna is not only due to wave reflection, as the object can adsorb a greater amount of energy (ohmic losses) which would be otherwise radiated or reabsorbed by the antenna. What is more, an object in the antenna vicinity will completely modify the EM field configuration making a hard job to formulate the problem in analytical terms.

To better understand the difference between near and far filed we can consider the field generated by an ideal elemental dipole antenna (and its dual elemental magnetic loop antenna), described by the Schelkunoff’s equations (Schelkunoff’s and Friis, 1952).

The curves of Figure 9 show the variation of the field impedance as a function of the range units from the antenna source (electric dipole in red and magnetic loop in blue). At short distances the impedance is purely reactive. Capacitive for the dipole and inductive for the loop, while at long distances both curves asymptotically tend to the intrinsic impedance of free space $Z_s = 376.7 \Omega$.

From the literature the most commonly quoted near-field/far-field boundary is the distance where the $1 / r$ and $1 / r^2$ terms of the electric field are equal, which corresponds to (13).

$$R_L = \frac{1}{k} = \frac{\lambda}{2\pi}$$

(13)

Note that (13) defines the boundary in wavelengths, implying that the boundary moves in space with the frequency of the antenna’s emission. Therefore, in the case of a non-sinusoidal emission, this limit should be weighted over the entire spectral content, or the centre frequency can be assumed as approximation.

Unfortunately, the boundary definition is not this straightforward and a large set of definitions may be found in the literature (Capps, 2001). The limit of (13) is also known as the radian sphere radius which delimits the region where reactive energy is stored and reabsorbed from the region where reactive energy transforms to radiation. Still from the energy point of view $R_L = \lambda / 2\pi$ matches the McLean’s lower boundary (McLean, 1996) as well. McLean’s equation states the fundamental limits on the radiation $Q$ factor of electrically small antennas. In other words, $R_L$ is the sphere radius for the shortest broadband antenna with $Q$ factor lower than 0.5.

### 4.2 NFD effect

What we would like to investigate now is what happens when the UWB medical radar antenna is placed close to the human thorax.

What we are expecting in the near-field operating condition is an overall increase of the radar sensibility due to the direct coupling between the antenna and the target. The multilayer adsorption model is still valid and because the amount of incident energy increases with the target proximity, the thorax will adsorb more energy than in the far-field condition, modifying the Poynting’s theorem balance.

From the antenna point of view, we expect performances degradation in proximity of the human body, especially for small planar UWB antennas with essentially omni-directional radiation patterns. That is because the human body reflects and adsorsbs the radiations from the antennas so that the radiation pattern becomes significantly directional in the horizontal planes while the average gain greatly decreases (Chen et al., 2006).

The performances of a small planar UWB antenna have been investigated using XFdtd Remcom simulator where a complex numerical model of the human body is included in the software package. The body model developed by Remcom is comprised of 24 biological tissues characterised by a frequency-dependent dielectric constant.
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and conductivity based on the curve fitting defined by Gabriel et al. (1996a, 1996b, 1996c).

FDTD simulations have been set up with a mesh cell of $1.0 \times 1.0 \times 1.0$ mm and a time step of 1.926 ps, allowing a correct analysis till the frequency of 29.98 GHz. The simulated volume of space was $540 \times 195 \times 110$ cells.

As an example for this theoretical analysis we used a planar monopole antenna numeric model (Figure 10) excited with a first derivative Gaussian pulse with centre frequency $f_c = 2.25$ GHz ($\lambda_c = 13.3$ cm) and amplitude 1 V.

Figure 10 Planar monopole antenna design (a) top view (b) bottom view (c) 3D transparency view (see online version for colours)

According to the previous paragraph, we are in near-field condition at distances from the thorax less than $R_\lambda = 2.1$ cm.

In that case, the previous far-field argumentations are no more exhaustive to explain how the radar retrieves a heart rate correlated signal.

Figure 11 EM field nearby the planar monopole antenna of Figure 10 close to the numerical human model (a) visible meshes (b) EM field only (see online version for colours)

Figure 11 shows a horizontal slice on the transverse plane where we can see the EM field around the antenna and inside the inner tissue of the human model.

The performances of the UWB antenna in terms of gain, input impedance and reflection coefficient have been evaluated in free space condition first and then at various distances from the human model thorax (Figure 11). The simulations results agree with Chen et al. (2006) where similar tests have been done using a numerical head model.

Figure 12(a) shows a strong degradation of the gain along the direction perpendicular to the thorax while positioning the antenna closer and closer to the skin. At the same time, the antenna input impedance and the reflection coefficient changes following the curves of Figures 12(b) and 12(c).

Figure 12(a) evidences a loss of about 20 dBi simply due to the presence of the human chest model at 10 mm from the antenna. At distances of 11 mm and 9 mm the gain curve shifts of about ±3 dBi with respect to the 10 mm reference. This means that a chest displacement of ±1 mm would generate a direct modulation of the antenna gain along the direction of the movement, which explains why the respiration is well visible using a trivial NFD UWB system.

Figure 12 (a) Antenna gain along the Y axis direction (b) antenna input impedance (c) reflection coefficient S11 (see online version for colours)

About the heart rate, we can say that even in near-field conditions the effect of the periodic variations of inner tissues EM properties (due to peripheral blood perfusion) will be still dominant over the deep tissue return echoes, as
it happens in far-field operating conditions. Therefore, the skin blood perfusion together with the elastic skin deformations induced by the heart mechanical activity are a reasonable explanation for the working principle of UWB medical radars into near-field operating conditions.

4.3 Summary

In this paragraph, we have studied the UWB medical radars as simple FDSs. We have discussed about the near field definition and its boundary around an emitting antenna, which is not constant as depending on the antenna design and the frequency content of the excitation signal.

The hypothesis to verify was that of a higher sensibility of our UWB system when placed in the vicinity of the human body, because of the direct coupling between the antenna and the body parts into the near field range.

Simulations show that a typical UWB planar monopole antenna is affected by strong performance degradations when placed at short distance from a numeric model of the human body. The antenna gain strongly depends on the distance between the antenna and the chest skin. Therefore, even a little movement of the chest (±1mm) can induce a heart-correlated signal on the antenna front end. That mechanism can simply explain the classic measurement of the variation of EM properties.

The influence of deep tissues is still negligible because of the strong attenuation predicted by the multilayer model previously discussed.

In near-field working conditions the energy around the antenna is directly coupled with HT, therefore even a little movement or variation of EM properties can induce a signal which will be highly correlated to that movement or that variation of EM properties.

5 Conclusions

This paper has investigated the mains interactions between EM waves and HT in the attempt to find an acceptable description of the UWB radars working principles. Almost all kind of interactions has been analysed and compared.

First of all, we have to distinguish between near and far field operation. Then, we remark that most of the energy is reflected on the skin and deep echoes have a negligible effect because of the huge attenuation at frequencies higher than 1 GHz.

Both in far and near field operation, the perfusion of the skin seems to have an important role in the heart rate detection with UWB radars. The near field case is more complicated to study and it could open further detection possibilities thanks to the direct coupling between the antenna and the HT.

It appears evident that at the moment no hypothesis can be eventually accepted for explaining UWB medical radar operation.

References


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