# Understanding the performance of microwave hydration assessment with layered planar models

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***Abstract*—Microwave hydration monitoring has been tested experimentally in volunteer studies with at least two prototype systems. However, there are differences between the systems both in terms of technical design, but also in the use of either the transmission loss or the time-of-flight differences to estimate the hydration status of the volunteer. In this work, a simplified planar model is used to estimate the best case sensitivity of microwave hydration monitoring to changes in the hydration status. Using this model, a change of 0.5 dB in transmission amplitude and a 0.89 % to 0.99 % change in time-of-arrival is calculated for muscle permittivity changes caused by each 1 % of dehydration, which correlates well with the literature. These results suggest that also simplified, planar models are useful for understanding microwave transmission through the arm.**

***Index Terms*—microwave hydration monitoring, bioelectromagnetics, propagation**

## I. Introduction

Microwave medical imaging has been proposed for biomedical applications in breast imaging brain monitoring, bone health, skin or shallow lesion monitoring. Fundamentally, microwave medical imaging exploits contrasts or changes in electrical properties of the tissues for imaging or monitoring. The electrical properties of biological tissues are correlated with the water content of the tissue where higher water content tissues typically have higher electrical properties than lower water content tissues. Microwave measurements of the body have been shown to be comfortable and easy-to-use.

Fluid status assessment and hydration monitoring is important in a variety of clinical and consumer electronics settings from emergency care to daily monitoring, activity tracking and athlete health [1]–[3]. The dependence of the electrical properties on the water content of the tissues is widely reported in experimental and clinical studies and has movitated the use of microwave imaging to infer hydration status directly. Malhydration is complicated and can refer to a variety of different clinical conditions including loss of total body water or changes in the distribution of body water without total loss.

Microwave reflection measurements have also been proposed to estimate the hydration levels of the skin with numerical, experimental and volunteer studies from several prototype systems [4]–[7]. Typically, transmission through the forearm in the 1 GHz to 8 GHz range is used to infer the bulk electrical properties of the arm. As the arm is mostly muscle, the transmitted signals and bulk properties are likely to be representative of the muscle permittivity.

Simplified analytical planar models, anthropomorphic two-dimensional simulations and several volunteer studies from at least two different systems have examined the experimental basis for this technique [4], [6], [8]. However, there are several differences between the volunteer studies in terms of the patient cohorts and hardware, and whether the transmitted signal amplitude, phase or otherwise is analysed. The volunteer study results are also difficult to conclusively analyse: studies with athletes in [6] measured before and after exercise have shown correlations between both the amplitude and phase of the transmitted signals and weight loss (assumed to be due to total body water loss), however, a similar approach with fasting volunteers in [5] showed little to no correlation.

The main aim of this work is to reexamine analytical planar modelling of the arm in light of the recent experimental and volunteer results and identify any confounding factors that may explain the findings. Specifically, this study seeks to validate the analytical planar models through comparison to the volunteer studies in the literature. Using the validated model, the effect of confounding factors such as skin changes, modelling decisions in terms of the numbers and orders of layers are quantified. Together, these will help understand the role of simplified analytical geometries in identifying the fundamental limits of performance of transmission-based microwave hydration assessment.

## II. Background and Methods

The arm predominately consists of muscle, tendons, nerves and blood vessels with two bones in the forearm [9]. Beneath the skin is normally adipose tissue surrounding the interior of the arm. From an electrical perspective, the skin is usually treated as one or two layers: one layer ($\sim 2\,\mathrm{mm}$) representing the various layers of the epidermis and dermis and another layer beneath this modelling the hypodermis and the subcutaneous adipose tissues. At higher frequencies, more granular electrical planar models may be needed, however, at microwave frequencies, these layers are very thin compared to the wavelength. A typical example of a three-layer arm model in shown in Fig. 1 with no subcutaneous fat.

Although simplified, for reflection and time-of-flight measurements, these planar models can model the arm well as the fastest path will typically be the closest to the straight-line path through the arm and reflections from the other scatterers, such as the bones and blood vessels, will be both more

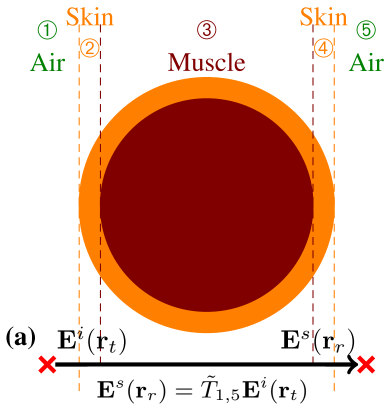


Fig. 1: An example of how a planar model is commonly used to represent the forearm. In many studies with through transmission of the arm, the scattered signals are processed as though they travelled along a straight-line path between the transmitter and receiver.

attenuated and arrive later in time. However, the sensitivity of the planar models to decisions about the number, thicknesses and properties of the layers is not well-established. Across the population, different skin types and structures are expected depending on age, gender, BMI (body mass index) and other factors [9], while the skin can also vary locally for a given individual and may not be symmetrical or may vary from mechanical compression during the measurement.

For a normally incident plane wave travelling in the $-z$ direction impinging on $N$ layers, the generalised transmission coefficient is [10]:

$$\tilde{T}_{\mathrm{IN}} = \prod_{j=1}^{N-1} e^{ik_{jz}d_j} \frac{T_{j,j+1}}{1 - R_{j,j+1}\tilde{R}_{j+1,j+2}e^{2ik_{(j+1)z}d_{j+1}}} \tag{1}$$

where, $R_{j,j+1}$ and $T_{j-1,j}$ are the Fresnel reflection and transmission coefficients between regions $j$ and $j+1$; $\tilde{R}_{j,j+1}$ is the generalized reflection coefficient which includes the reflections from subsequent layers; and $k_{jz}$ and $d_j$ are the wavenumber and thickness of region $j$ respectively.

A comprehensive analysis of layered planar models of the skin is presented in [8] in the context of dosimetry studies. In some studies at higher frequencies, no tissue deeper than the subcutaneous fat is modelled at all (e.g. [11]), however, in most cases, a subcutaneous fat layer on the order of millimeters is considered. For most dosimetry models, only the skin and the immediate layers below are modelled, and the other side of the body is not typically included. For hydration monitoring, the antennas are normally placed in direct contact with the back and the front of the arm with at approximately the middle of the forearm. At this location, the antennas are typically between 50 mm to 70 mm. One and two layer skins are most common for microwaves and that the type of model is very variable across the literature. In this study, a two or three layer skin is used consisting of a "skin" layer representing both the dermis and epidermis, a subcutaneous fat

TABLE I: Five different skin models used with different arm thicknesses are used to model healthy variation in the population. The thicknesses of the skin layers (in mm) is shwon where SC—Strateum Corneum.

| Ref. | SC | Skin | Fat |
|---|---|---|---|
| [12] | 0.02 | 1.46 | 4 |
| [13] | 0.02 | 1.45 | 3 |
| [14] | 0.02 | 2.52 | 5.6 |
| [14] | 0.01 | 0.46 | 1.1 |
| [8] | 0 | 2.7 | 1.7 |

layer directly below the skin and an optional strateum corneum which is typically very thin and not expected to impact the results at these frequencies. By modelling the arm interior as muscle, this results in a five or seven layer symmetrical arm model depending on the skin type, however, there can be asymmetries: for example, the skin on the front and the back of the arm is not the same and can differ in thickness, colour and hair. In the following analysis, arms of different thicknesses are compared with one of five skin models based on the literature as described in Table I.

The electrical properties for each layer type are taken from the data reported in both [15] and [16]. Dehydration is modelled as a proportional change in the electrical properties of the muscle with no correlated changes in the other tissue layers and changes in the other layers are analysed as confounding factors.

## III. Results

The results are presented in several parts:

- firstly, a preliminary analysis of the reflection and transmission coefficients is shown in Section III-A.
- secondly, possible confounding factors to the estimate of the sensitivity of the signals to changes in the underlying permittivites are analysed in Section III-B.
- thirdly, the estimated bulk permittivity in Section III-C.

### A. Transmission loss and Frequency Range

In this section, the total transmission loss through the arm with respect to frequency, dielectric properties, skin model and arm thickness and are compared to experimental results from the literature. Figure 2 shows transmission loss in dB for a "thin" arm of thickness 50 mm and a "thick" arm of thickness 70 mm. Comparable data from [4]–[6] is also shown from experimental measurements. Broadly speaking, the analytical values correspond well to the experimental data, considering that no antenna effects, experimental noise or systematic uncertainty is included in the analytical data.

In [4] with 20 recreationally active volunteers, loss through the arm for one participant was approximately −30 dB at 2.5 GHz whereas the analytical model would suggest −20 dB to −35 dB. A study of fasting volunteers in [17] reports losses around −18 dB at 2 GHz and for similar arm thicknesses, the analytical model predicts −15.9 dB to −27.5 dB for the same frequency. A study of athletes in [6], antennas are in direct

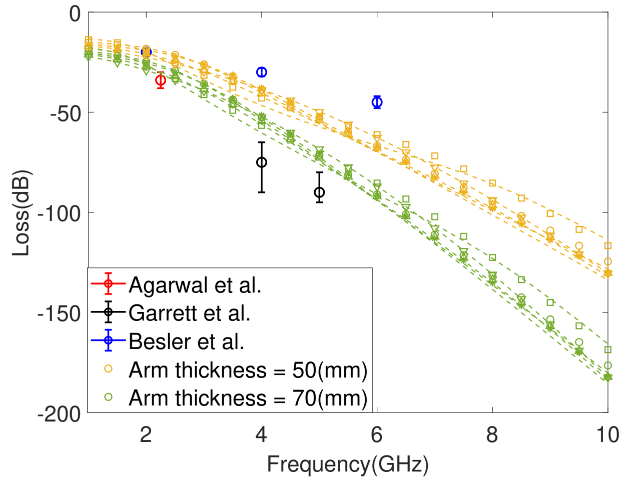


Fig. 2: Relative attenuation for the five skin models, arms of two thicknesses and using the IFAC (dashed) and ITIS (points) averaged electrical properties for the layers. Depending on the arm thickness, a maximum frequency of between 6 GHz to 8 GHz would need measurement equipment with a 100 dB dynamic range.

contact with the skin, with a separation distance ranging from 51.3 mm to 67.2 mm. Although this study found a correlation between the estimated average permittivities and body mass while [17] did not, the loss in the range 4 GHz to 5 GHz is reported between −90 dB to −65 dB, 10 dB to 20 dB below the thicker analytical arm model.

The magnitude of representative reflection and transmission coefficients are shown in Fig. 3 for a 4 % change in permittivity of the muscle. As is expected, the sensitivity of the amplitude of the reflection coefficient is less sensitive to changes in the muscle permittivity, with a 0.2 dB change at 3 GHz compared to a 1.7 dB change in the magnitude of the transmission coefficient at the same frequency.

### B. Confounding factors

As discussed in Section II, many models of the skin, particularly at higher frequencies, include the stratum corneum. However, as this layer is very thin compared to the wavelength at microwave frequencies, it is not expected to have a large influence. Considered a wide range of arm thicknesses and skin models, the maximum difference in amplitude of the transmission or reflection coefficients is 0.004 dB, so the stratum corneum is not included in the following analyses.

To estimate the effect the choice of skin model may have on the estimated change in amplitude of the transmission coefficient due to changes in the muscle permittivity, the change in amplitude of the transmission coefficient for the range of models shown in Table I at 3 GHz was calculated and is shown in Fig. 4. Each plot shows the absolute magnitude of the change due to muscle permittivity changes caused by each 1 % of dehydration for arms varying in thickness between

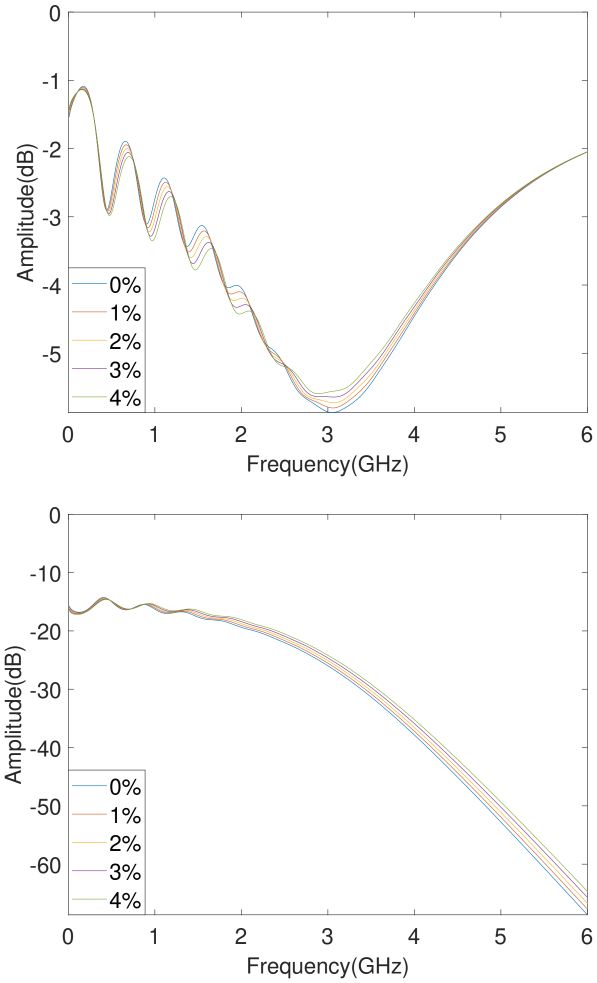


Fig. 3: The amplitude of the reflection (top) and transmission (bottom) coefficients for a representative arm for four differing muscle permittivities. As can be seen, in this ideal case, the magnitude of the reflection coefficient changes by 0.2 dB at 3 GHz and the magnitude of the transmission coefficient by 1.7 dB for a 4 % change in permittivity of the muscle, which is comparatively large.

30 mm to 70 mm. Each individual box plot includes data from 1,368 forearm models.

The change in amplitude in this model is due to two effects: reflections at the boundaries of the layers and, losses in the media. Regardless of skin model, the change in amplitude of the transmission coefficient for muscle permittivity changes caused by each 1 % of dehydration increases with arm thickness, as the signals travel for a much longer path through the comparatively lossy muscle. However, as shown in Fig. 2, the absolute magnitude of the transmission coefficient is also lower for larger arms.

The results from for the 50 mm arm from Fig. 4 are analysed in detail in the remainder of this section. Firstly, the absolute magnitude of the transmission coefficient for each combination of the skin and fat thickness is shown in Fig. 5. Depending on

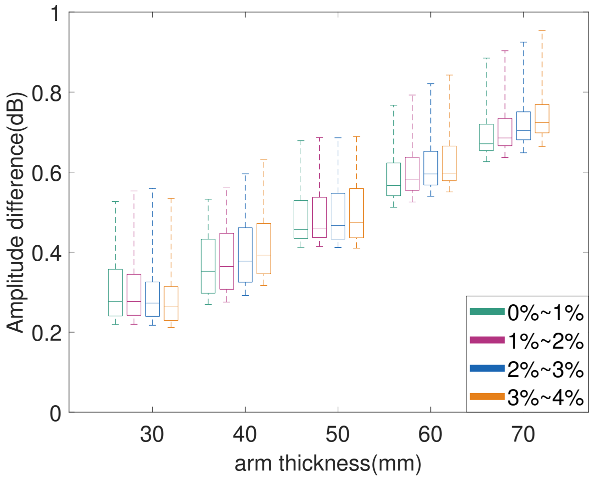


Fig. 4: The magnitude of the change of the transmission coefficient due to muscle permittivity changes caused by each 1 % of dehydration for a range of arm models is shown. The magnitude of the change depends on the thickness of the arm as well as the type of skin model chosen.

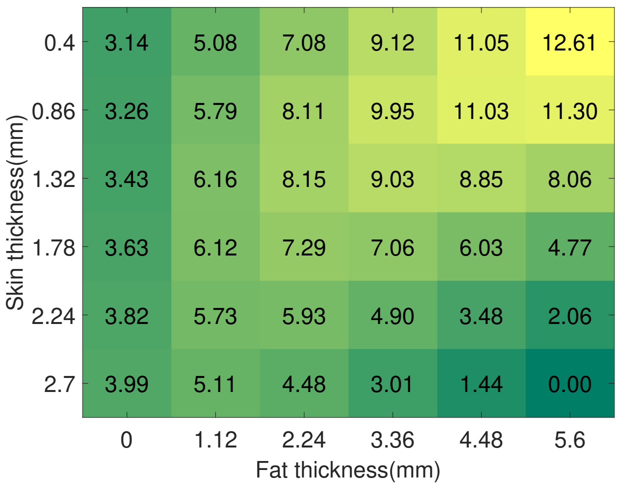


Fig. 5: The magnitude of the transmission coefficient depending on the thickness of the skin and fat layers of the model is shown, normalised to the model with 2.7 mm skin, 5.6 mm fat layer and 33.4 mm of muscle is shown.

the relative thicknesses of the skin and fat layers, there can be 1 dB to 2 dB changes in the amplitude of the transmission coefficient for 1 mm changes in the thicknesses of the underlying layers. These changes suggest that the transmission coefficient amplitude from planar models is somewhat sensitive to the assumptions in the skin model and the exact geometry.

A similar analysis is presented in Fig. 6 except the change in magnitude of the transmission coefficient due to a 1 % dehydration change in muscle permittivity is shown. In this case, the change is much more stable with respect to the modelling assumptions. Most applications of microwave fluid

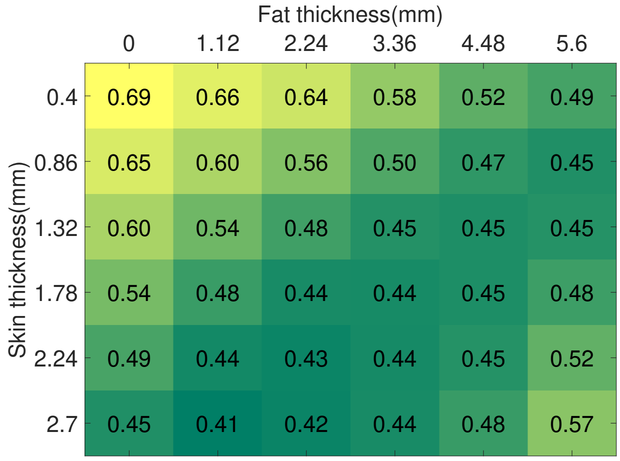


Fig. 6: The change in the magnitude of the transmission coefficient for muscle permittivity changes caused by each 1 % of dehydration depending on the thicknesses of the layers in the skin model is shown. Compared to the absolute magnitude, the predicted change is more stable and 1 mm changes in the skin thickness have less of an impact.

assessment suggest short or long term monitoring rather than instantaneous measurements [18]. In these cases, it is the trend over time that is of interest. As the probes will typically not be worn consistently, slight differences in pressure may compress the skin differently each time. These results suggest that in terms of transmission loss, changes in skin thickness during measurement are unlikely to affect the measurement, although the quality of the contact between the probe and the skin may have an effect which is not included in this model.

Together, these results suggest that the change in the magnitude of the transmission coefficient for a 1 % dehydration change in permittivity of the muscle may be on the order of 0.5 dB depending on the thickness of the arm. The absolute amplitude of transmission coefficient is sensitive to the modelling parameters and is less likely to be reliable.

### *C. Average permittivity calculated from time of arrival*

In this section, the changes in the time-of-arrival of the transmitted signal due to muscle permittivity changes caused by each 1 % of dehydration are examined. The results are compared to the volunteer studies in [5], [6] which use the same processing method.

Firstly, depending on the skin model, for each 1 % change in estimated bulk permittivity the muscle permittivity changed by between 0.99 % to 1.13 %. As the propagation path is mostly through the muscle and there is straight-line propagation in the case of a plane wave impinging on the layers, it is expected that the bulk permittivity tracks the muscle permittivity closely. This also represents a best case measurable change.

The technique from [18] is used to relates the change in muscle permittivity to the change in body mass. Shown in

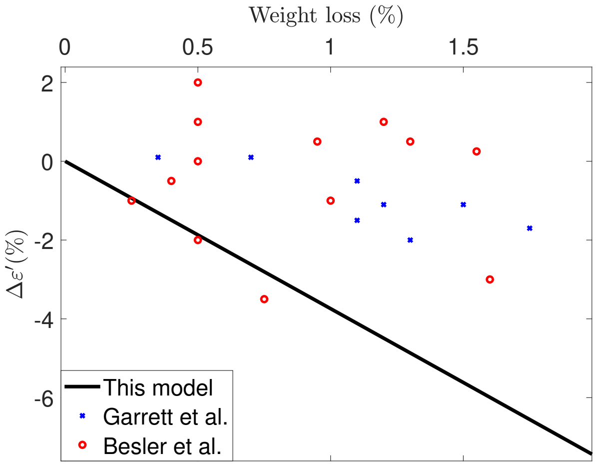


Fig. 7: Average permittivity calculated from time of arrival from a planar model compared to experimental measurements in [5], [6] from the literature using the same technique.

Fig. 7, the bulk permittivity changes by 3.7 % for muscle permittivity changes caused by each 1 % of dehydration in the planar model. Similar, but anthropomorphically correct, computational simulations in [18] suggest a 1.86 % to 2.5 % change and the experimental validations in [5], [6] suggest, at best, a 1.43 % change.

The results of the planar model are likely to be a best case estimate, however, given the geometric simplicity of the model, they estimates correspond well with the simulations in [18] and suggest that these simple models can be useful for understanding the propagation through the arm in some circumstances.

## IV. Conclusion

In this paper, a multi-layered planar model is used to examine the sensitivity of microwave reflection and transmission measurements to fluid status changes. At least two prototypes using this technique have been tested with healthy volunteers, however, differences in the trial design and reported results make it difficult to synthesise.

The transmission loss and estimated permittivity measured with the planar model corresponds well with the experimental and volunteer results in the literature. This suggests that muscle permittivity changes caused by each 1 % of dehydration result in an approximate 0.5 dB change in transmission amplitude and a 0.89 % to 0.99 % change in estimated permittivity.

Many different layered planar models have been proposed for the arm and the skin with different thicknesses and numbers of layers. The calculated changes in transmission due to changes in muscle permittivity are not very sensitive to the modelling parameters, however, the absolute magnitude of the transmission coefficient can change by more than 10 dB depending on the thicknesses of the skin and fat layers.

These results suggest that the planar models are useful for understanding the best cases changes in transmission due to changes in permittivity and that these findings are not very sensitivity to the modelling assumptions. However, the absolute magnitude of the transmission coefficient is more dependent on the exact model used.

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