Assessment of absorbed power density and temperature rise for nonplanar body model under electromagnetic exposure above 6 GHz

Yinliang Diao1,2,∗, Essam A Rashed3,4 and Akimasa Hirata2,4

1 College of Electronic Engineering, South China Agricultural University, Guangzhou 510642, People’s Republic of China
2 Department of Electrical and Mechanical Engineering, Nagoya Institute of Technology, Nagoya 466-8555, Japan
3 Department of Mathematics, Faculty of Science, Suez Canal University, Ismailia 41522, Egypt
4 Center of Biomedical Physics and Information Technology, Nagoya Institute of Technology, Nagoya 466-8555, Japan

E-mail: diaoyinliang@ieee.org

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Abstract
The averaged absorbed power density (APD) and temperature rise in body models with nonplanar surfaces were computed for electromagnetic exposure above 6 GHz. Different calculation schemes for the averaged APD were investigated. Additionally, a novel compensation method for correcting the heat convection rate on the air/skin interface in voxel human models was proposed and validated. The compensation method can be easily incorporated into bioheat calculations and does not require information regarding the normal direction of the boundary voxels, in contrast to a previously proposed method. The APD and temperature rise were evaluated using models of a two-dimensional cylinder and a three-dimensional partial forearm. The heating factor, which was defined as the ratio of the temperature rise to the APD, was calculated using different APD averaging schemes. Our computational results revealed different frequency and curvature dependences. For body models with curvature radii of >30 mm and at frequencies of >20 GHz, the differences in the heating factors among the APD schemes were small.

1. Introduction
With the advancement of technology, there has been a trend of adopting higher frequencies to achieve larger bandwidths and hence higher data transfer rates. Additionally, fifth-generation mobile communications allow the use of frequency bands above those employed for current wireless communications (typically <6 GHz) (Imai et al 2015, Lee et al 2018). Exposure to electromagnetic (EM) radiation in this frequency range may potentially cause adverse heating effects on biological tissues. Limits for exposure to radio frequency (RF) EM fields in the range of 100 kHz to 300 GHz have been established by the International Commission on Non-Ionizing Radiation Protection (ICNIRP) (ICNIRP 2020) and the IEEE International Committee on Electromagnetic Safety (IEEE ICES) Technical Committee 95 (Bailey et al 2019). Both the guideline and standard specify not-to-be-exceeded internal EM doses, which were determined according to the threshold for established health effects with reduction (safety) factors. ICNIRP calls this limit the basic restriction (BR), and IEEE refers to it as the dosimetric reference limit (DRL).

Above 6 GHz, the heating effect occurs predominantly in the superficial tissue, particularly in the skin (Ziskin et al 2018, Hirata et al 2019). With an increase in the frequency, the EM power absorption in the body and the resultant temperature rise become more superficial. For RF exposure above 6 GHz, the BR is specified in terms of the absorbed power density (APD) in the ICNIRP-2020 guideline, and the incident power density (IPD) corresponding to the APD is used as the reference level (ICNIRP 2020). The equivalent
DRL is defined as the epithelial power density in IEEE C95.1 (Bailey et al 2019). For convenience, the term ‘APD’ is used for both the APD and the epithelial power density herein.

Morimoto et al (2016) evaluated the specific absorption rate (SAR) and temperature rise in human head models up to 30 GHz. Strong correlations between the peak spatial-average SAR (psSAR) and temperature rise were observed at frequencies up to 3–4 GHz, and weaker correlations were observed at higher frequencies. Dosimetric studies for frequencies above 6 GHz have been conducted for both near- and far-field exposure scenarios (Alekseev et al 2008, Kanazaki et al 2010, Wu et al 2015, Sasaki et al 2017, Xu et al 2017, He et al 2018, Colombi et al 2018, Li et al 2019). Theoretical solutions to EM and bioheat problems for a two-dimensional (2D) layered body model were derived, and Monte Carlo analyses were performed for different tissue thicknesses (Sasaki et al 2017, Li et al 2019). Samaras and Kuster (2019) computed the transmittance coefficients for a 2D exposure scenario using the exact solution for a simple layered body model. Nakae et al (2020) revealed that the normal component of the IPD correlates well with the skin-temperature rise, regardless of the incident angle. Several studies have been performed on the correlation between the power-density averaging area and the skin surface temperature (Hashimoto et al 2017, Neufeld et al 2018, Funahashi et al 2018). Funahashi et al (2018) reported that the APD averaged over 4 cm² exhibited a strong correlation with the skin-temperature rise in the frequency range of 30–300 GHz and was reasonable and conservative at frequencies as low as 10 GHz. Kageyama et al (2019) developed an exposure system and measured the temperature rise in forearm skin exposed to a focused beam generated by a lens antenna at 28 GHz. In the foregoing studies, cubic models with planar surfaces were generally adopted for the assessment of millimeter-wave exposure, in contrast to dosimetric studies for frequencies below ∼10 GHz (Hirata et al 2002, Dimbylow et al 2008, Conil et al 2011, Wu et al 2011), where anatomically accurate body models have been widely used. The effects of the curvature of the body surface and the internal tissue composition on the heating factors remain unclear. These effects are difficult to evaluate directly via theoretical solutions, and numerical approaches are preferred. Additionally, current product safety standards have not yet provided detailed calculation schemes for the averaged APD, particularly for human models with curved surfaces used in numerical dosimetry. Open questions include the following: i) Should the averaging area be parallel to the grid axes in the finite-difference time domain (FDTD) method or bent along the curved body surface? ii) What are the integration limits for the curved body surface? For details regarding additional studies, the reader is referred to previous reviews (Foster et al 2016, Hirata et al 2019).

Considering the rationale for setting the limit, the APD should be linked to the maximum temperature rise. There are several major uncertainty sources in the thermal analyses, particularly for models with curved surfaces. The finite-difference (FD) approach has been commonly used in thermal analysis with voxelized human models. Laakso (2009) investigated the effect of the spatial resolution on the SAR and temperature in head models under RF plane-wave exposure. In general, the error in SAR computation results in that of the temperature rise. The error in the temperature computation is generally smaller than that in the SAR computation, because of the heat diffusion. One disadvantage of the FD analysis is that the model surface is discretized into small cubes, resulting in a large surface area. This increases the heat transfer from the model, hence underestimation of the temperature rise, if the measured heat-transfer coefficient is directly used as a boundary condition. A few computational schemes have been proposed for improving the accuracy of the heat flux at the model surface (Samaras et al 2006, Neufeld et al 2007, Laakso 2009). Neufeld et al (2007) proposed a conformal scheme for correcting the heat flux. However, this method requires accurate knowledge of the normal directions for each boundary voxel. A simple correction method was adopted by Laakso (2009); the heat convection rate for the skin voxel $H$ was corrected as $H/\sqrt{n}$, where $n$ represents the number of neighboring air voxels. However, the total surface area of a voxel model was still slightly overestimated (Laakso and Hirata 2011).

The objective of this study was to reliably assess the averaged APD and temperature rise for body models with nonplanar surfaces at frequencies above 6 GHz. Different calculation schemes for the averaged APD were proposed and evaluated. We proposed a local compensation method for the heat convection at the model boundary in bioheat calculations. The heating factors (ratios of the surface-temperature rise to the APD) for the nonplanar body model were evaluated to investigate the discrepancies caused by the different curvature radii and frequencies.

2. Models and numerical methods

2.1. Models

Two models were adopted for evaluation of the APD and temperature rise. The first was a 2D cylindrical multilayer model, as shown in figure 1(a). This model comprised three types of tissues: skin (thickness of 1.4 mm), fat (4 mm), and muscle. Different outer radii of the cylindrical models (i.e. 20, 30, 40, and 50 mm) were considered. The other model was a three-dimensional (3D) partial forearm model extracted from the
Figure 1. Models used for numerical assessments. (a) 2D cylindrical multilayer model; (b) 3D partial forearm model with a height of 40 mm; (c) position of the source at a distance $d$ from the forearm model.

Table 1. Dielectric properties of the tissues of the multilayer model.

| Tissue | Conductivity $\sigma$ (S m$^{-1}$) | Relative permittivity $\epsilon_r$ |
|--------|---------------------------------|---------------------------------|
|        | 6 | 10 | 20 | 30 | 45 | 60 | 6 | 10 | 20 | 30 | 45 | 60 |
| Skin   | 3.89 | 8.01 | 19.2 | 27.1 | 33.4 | 36.4 | 34.9 | 31.3 | 22.0 | 15.5 | 10.4 | 7.98 |
| Fat    | 0.31 | 0.59 | 1.26 | 1.79 | 2.39 | 2.82 | 4.94 | 4.60 | 4.00 | 3.64 | 3.32 | 3.13 |
| Muscle | 5.20 | 10.6 | 24.7 | 35.5 | 46.1 | 52.8 | 48.2 | 42.8 | 31.0 | 23.2 | 16.5 | 12.9 |

Table 2. Dielectric properties of the tissues of the forearm model at 28 GHz.

| Tissue              | Conductivity $\sigma$ (S m$^{-1}$) | Relative permittivity $\epsilon_r$ |
|---------------------|---------------------------------|---------------------------------|
| Skin                | 25.8                            | 16.6                            |
| Fat                 | 1.70                            | 3.70                            |
| Muscle              | 33.6                            | 24.4                            |
| Bone (Cortical)     | 4.94                            | 5.17                            |
| Bone (Cancellous)   | 8.87                            | 7.51                            |
| Blood               | 37.0                            | 23.9                            |
| Tendon              | 23.6                            | 13.9                            |

XCAT phantom (Segars et al 2010), as shown in figure 1(b). The forearm model consisted of seven types of tissues (skin, fat, muscle, blood, tendon, and cortical and cancellous bones). For all the models and frequencies, a spatial resolution of 0.1 mm was adopted.

2.2. EM calculations

For EM calculations, the FDTD method (Taflove and Hagness 2005) was used. A 15-layer convolutional perfectly matched layer (Roden and Gedney 2000) was adopted to truncate the simulation domain. For 2D analysis, both transverse-electric (TE) and transverse-magnetic (TM) plane incident waves were considered as radiation sources. For 3D analysis, a $4 \times 1$ half-wave dipole antenna array was used. The antenna array was located in the front of the forearm, as shown in figure 1(c). Different distances ($d$) between the antenna and the forearm, i.e. 5, 10, 15, 20, 30, and 40 mm, were considered. The four dipole elements were fed by delta-gap voltage sources with the same amplitudes and phases. The local SAR was calculated by $\text{SAR} = \left(\frac{\sigma |E|^2}{\rho}\right)$, where $\sigma$ is the tissue conductivity, $\rho$ is tissue mass density, and $E$ is the root-mean-square value of averaged electric field components, which were defined on the edges of the target voxel.

The simulated frequencies range from 6 to 60 GHz for the 2D analyses. The dielectric properties were obtained from the work of Gabriel et al (1996) and are presented in table 1. For the 3D analysis, the working frequency was set as 28 GHz. The forearm model comprised seven tissues, and their dielectric properties at 28 GHz are presented in table 2. Note that the dielectric properties given by Gabriel et al (1996) are based on measurement data below 20 GHz. Recently study (Sasaki et al 2014) reported the measured dielectric properties of epidermis and dermis up to 110 GHz. The effect of the dielectric properties on the dosimetry quantities at millimeter wave band was discussed in Sasaki et al (2017) using Monte Carlo simulation, it has been shown that a standard deviation of 15% in the dielectric properties has little effect on the transmittance and resultant temperature rise. In this study, as the dermis and epidermis are not distinguished in XCAT phantom, also considering the consistencies at lower frequencies, the dielectric properties reported by Gabriel et al (1996) were adopted.
2.3. Calculation schemes for averaged APD

According to (ICNIRP 2020) and IEEE C96.1, there generally exists two calculation methods of the spatial-average APD. The first is as follows:

\[
S_{ab} = \int_0^{z_{max}} \rho(x,y,z) \cdot \text{SAR}(x,y,z) \, dz / A,
\]

where \(z = 0\) corresponds to the body surface, \(z_{max}\) encloses most of the power deposition in the body, \(A = 4 \, \text{cm}^2\) represents the averaging area, \(\rho(x,y,z)\) and \(\text{SAR}(x,y,z)\) denote the tissue mass density and local SAR at location \((x,y,z)\), respectively.

The other method is equivalent to the specification in IEEE C95.1 (Bailey et al. 2019), where the APD is defined as the EM power flow through the epithelium per unit area directly under the stratum corneum. For a time-harmonic signal waveform, the time-averaged APD averaged over an area of \(A\) can be evaluated using equation (2):

\[
S_{ab} = \frac{1}{2} \int_A \text{Re} [E \times H^*] \cdot ds / A,
\]

where \(E\) and \(H\) denote the complex electric and magnetic field vectors, \(ds\) represents the integral variable vector normal to the body surface, \(\text{Re}\) and \(*\) denote the real part and complex conjugate, respectively. In both the guidelines and standards, the APD should be averaged over a square area of 4 cm² at frequencies between 6 and 300 GHz. This averaging area generally provides good estimations of the surface-temperature rise (Hashimoto et al. 2017, Funahashi et al. 2018).

For practical implementation of the averaged APD in voxel models, the definition of (1) was adopted. Moreover, we developed four different calculation schemes for the spatial-average APD for nonplanar models, as illustrated in figure 2. The bounds of the integration volumes are represented by red polygons. In figures 2(a)–(b), the upper bound \(L_1\) is parallel to the grid axis. In figures 2(c)–(d), \(L_1\) is bent along the skin surface. In figures 2(a)–(c), \(L_2\) and \(L_3\) are parallel to the grid axis, whereas in figures 2(b) and (d), \(L_2\) and \(L_3\) are parallel to the internal electric field gradients at the model surface. The lower bound of the integration volume, i.e. \(L_4\), is defined as the contour where the electric field strength is 1/1000 of the maximum value in the integration volume.

2.4. Bioheat calculation

The widely used bioheat equation (Pennes 1948) was employed for calculation of the temperature rise inside the human body model. Under the assumption of the steady-state condition, the following equation was used for the temperature rise \(\Delta T\):

\[
\nabla \cdot (k \nabla \Delta T) - B \Delta T + Q_v = 0,
\]

where \(k\) [W (m·°C)⁻¹] represents the thermal conductivity; \(B\) [W (m³·°C)⁻¹] is a coefficient related to the blood perfusion rate; and \(Q_v\) [W m⁻³] represents the power-loss density, which is related to the SAR by \(Q_v = \text{SAR} \cdot \rho\), where \(\rho\) [kg m⁻³] represents the tissue mass density. The temperature rise \(\Delta T\) is defined at the center of each voxel. The thermal parameters used in this study were identical to those employed by Hirata et al. (2006).

The Neumann boundary condition in (4) was applied to the boundary of the model:

\[
\frac{k \partial \Delta T}{\partial n} + H \Delta T = 0,
\]
where $H$ [W (m$^2$·C)$^{-1}$] represents the heat convection rate from the skin to the surrounding air. The bioheat equation is linear at the thermally steady state (Hirata and Shiozawa 2003). A lower heat convection rate results in a slightly higher temperature rise. In this study, $H$ was set as 8 W (m$^2$·C)$^{-1}$, which was consistent with that used in several previous studies (Hirata et al 2007, Laakso 2009, Kodera et al 2018), and also within the range of values used by Sasaki et al (2017) considering different air velocities.

### 2.5. Compensation method for heat convection rate at air/skin interface

For the implementation of the Neumann boundary condition, the heat flux through the boundary of the voxel via convection must equal the heat reaching that voxel from its neighboring voxels via conduction (Bernardi et al 1998). The total heat flux through a boundary voxel is given as $H \cdot S$, where $S$ represents the effective area of the air/skin interface. For a voxel belonging to a planar surface parallel to the voxel surfaces, $S = \Delta^2$, whereas in stepped boundaries, $S > \Delta^2$. To reduce the uncertainties in bioheat calculations, an accurate estimation of the surface area is required. Mullikin and Verbeek (1993) proposed a simple method for estimating surface area of 3D binary objects. This method assigns surface-area weights to each boundary voxel. The boundary voxels can be classified into five types: $S_n$, $n = 1, 2, \ldots, 5$, where $n$ represents the number of voxel faces exposed to the background. Figure 3(a) shows a local boundary region containing different types of boundary voxels. The total surface area of the model can be estimated as follows:

$$A = \left( \sum_{n=1}^{5} w_n \cdot N_n \right) \cdot \Delta^2, \quad \text{(5)}$$

where $N_n$ represents the number of voxels of type $S_n$, and $w_n$ represents the weight for voxel type $S_n$. The following optimized $w_n$ values were reported: $w_1 = 0.8940$, $w_2 = 1.3409$, and $w_3 = 1.5879$ (Mullikin and Verbeek 1993). If the spatial resolution is sufficiently high, there is no obvious deviation of the boundary region from a plane. In such regions, only three types of voxels ($S_{1-3}$) exist. Voxels of type $S_{4-5}$ can occur on sharply curved surfaces. For type $S_{4-5}$, the weights $w_4 = 2$, and $w_5 = 8/3$ were presented by Mullikin and Verbeek (1993). Nonetheless, for a sphere model, there are significantly less $S_{1-3}$ voxels than $S_{4-5}$ voxels; therefore, the weights of the $S_{1-3}$ voxels are insignificant. A simple and straightforward compensation method involves adopting a factor $w_n$ for voxel type $S_n$. Consequently, the heat flux through the boundary voxel becomes $Hw_n \Delta^2$. Thus, the total heat flux through the body surface can be accurately estimated.

However, particular attention should be paid to local stepped regions, as shown in figures 3(b)–(d). In these locally planar regions, only single type of boundary voxel exists. With the foregoing compensation method, the local effective surface area would be underestimated as $w_n < \sqrt{n}$. To resolve this issue, we propose a local correction method using a $3 \times 3 \times 3$ moving cube, which is centered at the target boundary voxel (outlined in red in figure 3). Assuming that $N'_n$ represents the total number of voxels of type $S_n$ within the moving cube, if there is only one type of boundary voxel within the cube (i.e. $N'_n = N'_A$, $N'_A$ is the total number of boundary voxels), the heat transfer through the central voxel is compensated as $H\sqrt{n} \Delta^2$. Thus, the heat convection rate in stepped regions, as shown in figures 3(b)–(d), is corrected.

### 3. Validation of EM computation for APD

The validation of the 2D FDTD analyses were confirmed by comparing the EM absorption of an infinite cylinder exposed to TE and TM plane incident waves at 10 and 30 GHz with analytical solutions obtained by

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**Figure 3.** Air/skin interface for voxel models. Black voxels indicate boundary voxels; yellow voxels indicate internal tissues. The boundary voxels are labeled according to the voxel type; the center voxels are outlined in red. The region in (a) contains different types of boundary voxels. The regions in (b) to (d) contain one type of boundary voxel. The effective surface areas for the central voxels are (b) $\Delta^2$, (c) $\sqrt{2}\Delta^2$, and (d) $\sqrt{3}\Delta^2$. 

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**References:**
- Bernardi et al, 1998
- Hirata and Shiozawa, 2003
- Mullikin and Verbeek, 1993
- Laakso, 2009
- Kodera et al, 2018
- Sasaki et al, 2017
- Mullikin and Verbeek, 1993
- Hirata et al, 2007
- Kodera et al, 2018
- Sasaki et al, 2017
- Mullikin and Verbeek, 1993
Table 3. Validation of the 2D FDTD Program with Mie Solution.

|          | 10 GHz |        |        | 30 GHz |        |        |
|----------|--------|--------|--------|--------|--------|--------|
|          |        | FDTD   | Mie    | RD     | FDTD   | Mie    | RD     |
| TE       |        |        |        |        |        |        |        |
| TAP [mW m$^{-2}$] | 28.19  | 28.36  | −0.59% | 28.28  | 28.73  | −1.56% |
| $APD_i$ [W m$^{-2}$] | 0.517  | 0.521  | −0.69% | 0.534  | 0.544  | −1.68% |
| $APD_i$ [W m$^{-2}$] | 0.481  | 0.485  | −0.82% | 0.526  | 0.535  | −1.70% |
| $APD_i$ [W m$^{-2}$] | 0.507  | 0.511  | −0.73% | 0.524  | 0.533  | −1.70% |
| $APD_i$ [W m$^{-2}$] | 0.470  | 0.474  | −0.91% | 0.516  | 0.525  | −1.70% |
| TM       |        |        |        |        |        |        |        |
| TAP [mW m$^{-2}$] | 42.10  | 42.09  | 0.01%  | 44.20  | 44.98  | −0.73% |
| $APD_i$ [W m$^{-2}$] | 0.514  | 0.517  | −0.50% | 0.550  | 0.560  | −0.72% |
| $APD_i$ [W m$^{-2}$] | 0.476  | 0.479  | −0.62% | 0.540  | 0.550  | −0.73% |
| $APD_i$ [W m$^{-2}$] | 0.503  | 0.506  | −0.56% | 0.538  | 0.548  | −1.69% |
| $APD_i$ [W m$^{-2}$] | 0.465  | 0.469  | −0.73% | 0.530  | 0.539  | −1.72% |

Table 4. Validation of the 3D FDTD Program with Mie Solution.

|          | 10 GHz |        |        | 30 GHz |        |        |
|----------|--------|--------|--------|--------|--------|--------|
|          |        | FDTD   | Mie    | RD     | FDTD   | Mie    | RD     |
| TAP [mW] | 2.107  | 2.093  | 0.67%  | 1.949  | 1.958  | −0.47% |
| $APD_i$ [W m$^{-2}$] | 0.559  | 0.561  | −0.33% | 0.551  | 0.556  | −0.95% |
| $APD_i$ [W m$^{-2}$] | 0.477  | 0.479  | −0.35% | 0.532  | 0.537  | −0.91% |
| $APD_i$ [W m$^{-2}$] | 0.537  | 0.539  | −0.34% | 0.529  | 0.534  | −0.93% |
| $APD_i$ [W m$^{-2}$] | 0.458  | 0.460  | −0.37% | 0.511  | 0.516  | −0.91% |

Mie theory (Schäfer et al 2012, Schäfer 2020). The radius of the cylinder was set to 30 mm, and the spatial resolutions of 0.2 and 0.1 mm were adopted at 10 and 30 GHz, respectively. The dielectric properties of dry skin were assigned to the model. The IPD was normalized to 1 W m$^{-2}$.

Table 3 shows the computation results for different exposure scenarios. $TAP$ [mW m$^{-2}$] denotes the total absorbed power in the cross section of the infinite cylinder; $APD_i$, $i = 1, 2, 3, 4$, denotes the APD obtained using calculation scheme $i$ (figure 2). As seen from table 3, the averaged $APD_i$ obtained by FDTD computation agree promisingly well with those obtained by Mie theory. The relative differences (RDs) are always lower than 1% at 10 GHz and are lower than 1.8% at 30 GHz.

The validation of the 3D FDTD analyses were confirmed by considering a sphere composed of skin exposed to plane wave at 10 and 30 GHz. The sphere radius was also set as 30 mm. The IPD was normalized to 1 W m$^{-2}$. As seen from table 4, the RD in $APD_i$ calculated by FDTD and by Mie solution are all below 1%. In Wang et al (2006), comparable RD levels in averaged SAR between FDTD and analytical solutions have also been reported, and a finer spatial resolution generally leads to lower computational uncertainty (Okada and Cole 2012).

In IEEE Std 1528 (2013), the SAR measurement uncertainty was defined directly from the standard deviation of multiple observations from the same measurement setup, with the consideration of a coverage factor of 2 ~ 3 determined based on the probability distribution. In accordance with the FDTD validation procedure introduced in IEC 62704-1 (2017), the maximum deviation of the computed psSAR under the specified exposure scenario should lie within ±10% of the given reference values. In this study, we adopted the similar procedure for the validation of spatial-average APD using Mie solution as a benchmark. As seen from table 3 and 4, the maximum deviation is well below the requirements specified in IEC 62704-1 (2017).

For computations using human voxel models, one major source of the numerical uncertainties is the staircasing error. For low-frequency (LF) exposures, as the stimulations of neuron cell or network are local effects, the BR is specified in terms of maximum local electric field (99th percentile value of the field averaged over a small volume of 8 mm$^3$) (Reilly and Hirata 2016). The local in situ electric field is prone to large numerical artefact, especially for voxels at tissue boundaries with high conductivity contrast (Gomez-Tames et al 2018, Diao et al 2019). For RF exposures under 6 GHz, psSAR averaged over 10 g of tissue and whole-body average SAR are used as BR (ICNIRP 2020). The uncertainties in averaged SARs are less significant due to the rather large averaging volumes (Dimbylow et al 2008). Since no exact solution exists for anatomical models, intercomparison has often been adopted to examine the numerical uncertainties for LF and RF exposure assessments (Beard et al 2006, Dimbylow et al 2008, Hirata et al 2010, Aga et al 2018). In Beard et al (2006), an average standard deviation of approximate 30% in psSAR was reported by comparing the results for a specific anthropomorphic mannequin (SAM) phantom from 14 research groups. As stated in IEC 62704-1 (2017), the calculated psSAR under the specified exposure scenario should be with in ±50% of
4. Validation of compensation method for bioheat calculation

To validate the proposed compensation method for the heat flux at the air/skin interface, we considered a homogeneous sphere model with radius of 60 mm. The heat conduction rate was set as 1.0 W (m·°C)\(^{-1}\), and a 200-W m\(^{-3}\) heat source was uniformly distributed inside the sphere. The heat convection rate between the skin and air was set as 8 W (m\(^2\)·°C\(^{-1}\)). The analytical solution to this problem is \(\Delta T(r) = -\frac{200r^2}{6} + 0.62\) °C, where \(r\) represents the radial distance.

The calculation results of the conformal and proposed compensation methods are presented in table 5 and figure 4. Implementation of the conformal method requires knowledge of the normal directions for each boundary voxel, which were obtained using a 3D Sobel operator in this study. As indicated by table 5, the total surface area was 2.2% larger than the real surface area for the conformal method. This resulted in an underestimation of the temperature rise, with a mean value of 0.4886 °C for the sphere surface, compared with the exact solution of 0.5 °C. The \(\Delta T\) calculated using the proposed method was almost identical to the exact solution, as shown in table 5 and figure 5. Moreover, compared with the conformal method, the proposed method provided a smaller standard deviation of \(\Delta T\) for the boundary voxels.

An ellipsoidal model, as shown in figure 6(a), was also adopted for validation. The three principal semi-axes of the ellipsoid were set as \(a = 30\), \(b = 50\), and \(c = 100\) mm. The thermal parameters were identical to those used for the spherical model. Spatial resolutions of 1, 0.5, and 0.25 mm were employed. The ellipsoid was tilted around the \(y\)-axis in steps of 15°. The calculated maximum temperature rises inside the ellipsoids for different spatial resolutions and tilt angles are presented in figure 6(b). As shown, the maximum temperature rises inside the models and on the surfaces were almost independent of the tilt angle, and the relative standard deviations caused by the tilt angle were smaller than \(\sim 0.3\%\). The discrepancies of the maximum temperature rises inside the models and on the model surfaces from those obtained via the MATLAB Partial Differential Equation Toolbox\textsuperscript{TM} (version R2020a) (Mathworks 2020) using the finite-element method (FEM) were within \(\sim 0.8\%\) and \(\sim 0.5\%\), respectively.

5. APD and temperature rise for nonplanar models

5.1. Comparison of the two definitions of averaged APD
IEEE C95.1 (Bailey et al 2019) defined the epithelial power density as the EM power flow through the epithelium per unit area directly under the body surface, which can be evaluated using equation (2). This
The definition was also mentioned in ICNIRP (2020) as one of the evaluation methods for APD. Theoretically, the APD obtained using equations (1) and (2) should be identical for the exposure scenario of plane-wave incidence on planar model. However, differences can be expected for nonplanar model. In numerical simulations, it is generally difficult to accurately calculate the APD using equation (2) owing to the staircasing error at the boundary voxels. To discuss the difference between the two equations, we adopted Mie theory to calculate the EM absorption in a 3 cm radius infinite cylinder composed of skin.

For calculating APD using equation (1), local SARs were first evaluated in each Cartesian grids inside the cylinder; the averaged APDs were then evaluated using the four schemes described in subsection 2.3; spatial resolutions of 0.2 mm and 0.1 mm were adopted for frequencies <30 GHz, and ≥30 GHz, respectively. For calculating APD using equation (2), electric and magnetic field vectors were calculated on a contour 1 μm under the cylinder surface in polar coordinates with a resolution of $\Delta \phi = 0.25^\circ$. Note that the differences between $APD_1$ and $APD_3$, and between $APD_2$ and $APD_4$, lie only in the effective averaging areas. To avoid redundancy, only the $APD_1$ and $APD_2$ results are presented here.

As seen from table 6, the averaged APD calculated using equation (2), $APD^{(2)}$, lies between the APDs calculated using equation (1) with schemes 1 and 2, i.e. $APD^{(1)}_1$ and $APD^{(1)}_2$, where the superscript indicates the equation used for the evaluation. The calculated $APD^{(2)}$ are closer to $APD^{(1)}_2$ than $APD^{(1)}_1$. At 6 GHz, the largest deviation of $APD^{(1)}_1$ from $APD^{(2)}$ is about 15% and decreases to below 3% at frequencies ≥20 GHz. This is because of the decreasing penetrations depth with frequency, thus the integration volumes for $APD^{(1)}_1$ and $APD^{(2)}_2$ become comparable.
Table 6. Comparison of APD (Percentage Deviation from APD\(^{(2)}\)) Calculated using equations (1) and (2).

| Pol. | Frequency | \(\text{APD}_1^{(1)} \ [\text{W m}\sp{-2}]\) | \(\text{APD}_2^{(1)} \ [\text{W m}\sp{-2}]\) | \(\text{APD}_3^{(2)} \ [\text{W m}\sp{-2}]\) |
|------|-----------|---------------------------------|---------------------------------|---------------------------------|
| TE   | 6 GHz     | 0.573 (15.4%)                   | 0.482 (−2.88%)                  | 0.496                           |
|      | 10 GHz    | 0.521 (6.41%)                   | 0.485 (−0.94%)                  | 0.490                           |
|      | 20 GHz    | 0.517 (2.46%)                   | 0.505 (−0.01%)                  | 0.505                           |
|      | 30 GHz    | 0.544 (2.45%)                   | 0.535 (0.75%)                   | 0.531                           |
|      | 45 GHz    | 0.585 (1.96%)                   | 0.579 (0.88%)                   | 0.574                           |
|      | 60 GHz    | 0.623 (1.96%)                   | 0.618 (1.10%)                   | 0.611                           |
| TM   | 6 GHz     | 0.545 (16.7%)                   | 0.454 (−2.89%)                  | 0.467                           |
|      | 10 GHz    | 0.517 (6.91%)                   | 0.479 (−0.94%)                  | 0.484                           |
|      | 20 GHz    | 0.529 (2.68%)                   | 0.516 (0.08%)                   | 0.516                           |
|      | 30 GHz    | 0.560 (2.69%)                   | 0.550 (0.85%)                   | 0.545                           |
|      | 45 GHz    | 0.602 (2.11%)                   | 0.595 (0.98%)                   | 0.590                           |
|      | 60 GHz    | 0.641 (2.10%)                   | 0.641 (2.10%)                   | 0.628                           |

5.2. Results for 2D multilayer model

The calculated local SAR and temperature rise in the 2D multilayer models for TE and TM incident waves are presented in figures 7(a)–(b), respectively. As shown, the EM power depositions were distributed superficially above 10 GHz. The peak local SARs for the TM cases were lower than those for the TE cases. This is because for the TE cases, the electric fields were parallel to the axis of the infinitely long cylindrical model. The maximum temperature rises were generally slightly larger for the TM cases than for the TE cases. As shown in figure 7(b), the EM wave penetrated into the cylindrical model from different incident angles, resulting in broader SAR and temperature distributions compared with the TE case. This phenomenon was attributed to Brewster’s effect (Li et al 2019).

In figure 8, the orange regions indicate the integration volumes for different APD calculation schemes at various frequencies. For the cylindrical model with a 20-mm radius, increasing the depth to enclose all the power transmitted through L1 is impossible at 6 and 10 GHz. Therefore, the lengths of L2 and L3 are set as 20 mm at 6 and 10 GHz for the 20-mm-radius model. This setting is in accordance with the definition of the
Figure 8. Integration volumes for different APD schemes. The radius of the cylindrical models is 20 mm. Orange shapes indicate different integration volumes.

Figure 9. Heating factors calculated using different APD schemes for 2D cylindrical models with radii of (a) 20, (b) 30, (c) 40, and (d) 50 mm, for TE incident waves.

10-g spatial-average SAR (Hirata et al 2019), where the side length of the equivalent integration cube is approximately 20 mm.

The calculated heating factors for different APD schemes are presented in figures 9 and 10 for TE and TM incident waves, respectively. As shown, the heating factors for the TM cases were slightly larger than those for the TE cases. The largest differences in the heating factor between the TE and TM cases was $\sim 20\%$, for scheme 4 of the 20-mm-radius model at 6 GHz. The differences decreased with the increasing frequency and curvature radius. For frequencies above $\sim 20$ GHz and curvature radii larger than $\sim 30$ mm, the differences in the heating factor between the TE and TM cases were smaller than $\sim 6\%$. 

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Figure 10. Heating factors calculated using different APD schemes for 2D cylindrical models with radii of (a) 20, (b) 30, (c) 40, and (d) 50 mm, for TM incident waves.

The effect of the curvature radius on the heating factor was rather small. For the TE cases, the largest difference in the heating factor between the 20- and 50-mm-radius models was \(\sim 10\%\) (at 6 GHz). For the TM cases, the largest difference was \(\sim 20\%\) (at 10 GHz).

Above \(\sim 20\) GHz, the heating factors for the APD were almost independent of the frequency. Reductions in the heating factors below 10 GHz were observed for most schemes. For convenience, we denote the heating factor evaluated via scheme \(i\) as \(HF_i\). At 6 GHz, \(HF_1\) and \(HF_3\) were \(\sim 20\%\) smaller than those at higher frequencies; while \(HF_2\) and \(HF_4\) were \(\sim 13\%\) and \(\sim 5\%\) smaller than those at higher frequencies for TE and TM incident waves, respectively.

For schemes 1 and 2, the upper bounds of the integration volume were tangent planes to the cylinder surface, whereas for schemes 3 and 4, the upper bounds were bent along the surface. Therefore, the integration volumes for the latter two schemes were smaller than those for schemes 1 and 2, resulting in larger heating factors. The differences between \(HF_1\) and \(HF_3\) and between \(HF_2\) and \(HF_4\) decreased with the increasing curvature radius. This is mainly attributed to the reduction in the deviation of the model surface from a plane. \(HF_3\) and \(HF_4\) decreased steadily with an increase in the radii, whereas \(HF_1\) and \(HF_2\) were relatively stable for all the radii.

In general, among the schemes examined, \(HF_2\) seems to be most stable across all radii and frequencies. For a cylindrical radius of \(\geq 30\) mm, the four schemes provided comparable heating-factor results, with relative standard deviations of \(< \sim 5\%\) for frequencies of \(\geq 20\) GHz and \(< \sim 10\%\) for frequencies of \(\geq 6\) GHz.

The heating factors calculated using 1D FDTD method were also included, as ICNIRP adopted 1D models to derive the exposure restrictions. The results are presented on both figure 9(d) and figure 10(d). The 1D analyses simulate the exposure of a planar stratified model to plane incident wave. For TE cases, all \(HF_i\) are approaching the 1D heating factors with increasing curvature radius; for 50 mm radius, the differences are generally lower than 8\%, and lower than approximate 4\% at frequencies \(\geq 30\) GHz. For TM cases, \(HF_1\) and \(HF_2\) are comparable to the 1D results, \(HF_3\) and \(HF_4\) show a steady decreasing trend toward the 1D heating factors with the increasing radius. As seen, the 1D heating factors may slightly underestimate the maximum temperature rise, compared with \(HF_3\) and \(HF_4\). However, the RDs are much smaller compared to the reduction factors, which account for biological variations, environmental factors, dosimetric uncertainties, etc, adopted in ICNIRP (2020) when deriving the restrictions.
Figure 11. Distributions of the (a) local SAR and (b) temperature rise in the partial forearm model. The antenna–forearm distance is 10 mm. The antenna accepted power is normalized to 20 dBm. The dark red box in (a) indicates the integration volume for APD averaging scheme 1.

Figure 12. Heating factors for a 4 × 1 dipole antenna array with different antenna–forearm distances.

5.3. Results for 3D forearm model
A partial human forearm model was adopted for 3D analysis. The temperature rises were calculated using the heat convection compensation method described in Section II. The calculated local SAR and $\Delta T$ distributions in the partial forearm model are shown in figures 11(a)–(b), respectively.

Figure 12 shows the heating factors for different distances between the antenna and the forearm. The differences in the heating factors among the different averaging schemes were marginal and smaller than those for the 2D cylindrical models. Note that in the 2D cases, a plane wave was used as the radiation source. The largest heating factors were observed for $d = 5$ mm. Fluctuations in the heating factors were observed within the reactive near-field region of the antenna ($d \leq \sim 15$ mm) and were mainly attributed to the oscillating peak power density in this field region (Colombi et al. 2015). The heating factors generally decrease with the increasing antenna-model separation. For $d = 40$ mm, the heating factors were approximately 0.22 °C·m$^2$W$^{-1}$ for schemes 1 and 2 and 0.23 °C·m$^2$W$^{-1}$ for schemes 3 and 4. These values were slightly higher than those for 2D multilayer models with 30-mm radii (the curvature radius of the forearm was approximately 30–40 mm). Such differences were mainly attributed to the field non-uniformity and the thicker fat tissue of the forearm model. As reported by Alekseev et al. (2008), a thicker fat layer tends to produce a slightly larger surface-temperature rise in the skin. For $d > 20$ mm, the heating factors become comparable to those of 1D simulations (figure 9 and figure 10), this is also in line with the findings in previous study (Hashimoto et al. 2017).

6. Discussion and concluding remarks
In bioheat calculations, the boundary conditions significantly influence the calculation results. The discretization of the model surface with cubes results in a larger model surface area; hence, the temperature rise is underestimated. This issue is particularly crucial for the assessment of millimeter-wave exposure, owing to the extremely shallow penetration depth. In the study of Laakso (2009), the heat convection rate for the skin voxel, i.e. $H$, was corrected as $H/\sqrt{n}$, where $n$ represents the number of exposed faces of the voxel. However, the total surface area of the voxel model was slightly overestimated (Laakso and Hirata 2011). This problem can be solved by introducing a compensation factor $f$, which is defined as the ratio of the real body surface area (if known) to the calculated one. The compensated heat transfer from one boundary voxel then
becomes $H\sqrt{\pi}\Delta^2$, $f < 1$. This method is applicable to whole-body exposure scenarios, as the total heat flux through the body surface is corrected. For localized exposure scenarios, however, compensation of the whole-body surface area may still lead to underestimation of the heat flux though the local surface parallel to the grid axes (as shown in figures 3(b)–(d)), because $f < 1$. In this paper, a new conformal compensation method was proposed for reducing the uncertainties in FD bioheat calculations for voxel models with nonplanar surfaces. This method does not require knowledge of the normal directions of the boundary voxels and can be easily incorporated into FD iterative programs with a minimal computational load.

Additionally, we confirmed that the proposed method provides an accurate estimation of the temperature rise. The variation in the maximum temperature rise was observed to be smaller than $\sim$1% for different spatial resolutions and different model rotation angles (shown in figure 6).

Because no clear definition of the calculation scheme for the APD is prescribed in the existing standards, four schemes were proposed, particularly for models with nonplanar surfaces. The relationships between the maximum temperature rise and the averaged APD were determined using different schemes. Compared with previous studies using planar models, comparable heating factors for the APD were observed for curvature radii larger than $\sim$30 mm and frequencies above $\sim$20 GHz. Above $\sim$20 GHz, all four schemes provided stable heating factors, and the heating factors decreased below $\sim$10 GHz. As reported by Hirata et al. (2019), even though there may be optimized physical quantities for estimating the surface-temperature rise below 10 GHz, the APD is a good surrogate, with relative variations smaller than $\sim$20%. In general, if the curvature radius is larger than $\sim$30 mm, the differences in the heating factors for different APD schemes are small, with relative standard deviations smaller than $\sim$5% for frequencies above $\sim$20 GHz and within $\sim$10% for frequencies above 6 GHz. In the 3D cases, the relative standard deviations of the heating factors are within $\sim$3.5% for all the APD schemes. Such differences resulted from APD schemes are not significant, considering that ICNIRP (2020) has already employed reduction factors of 2 and 10, which account for various sources of variabilities, including biological variability in the population (e.g. age, sex), variation in baseline conditions (e.g. tissue temperature), environmental factors, and dosimetric uncertainty, etc, when deriving the restrictions for occupational and general public exposures above 6 GHz.

In previous studies, slightly larger heating factors were observed for TM-like exposure using 2D infinite-slab models (Samaras and Kuster 2019, Li et al. 2019, Nakae et al. 2020). This phenomenon was also observed in the present study for the 2D multilayer cylindrical model, as shown in figures 9 and 10. However, the differences in the heating factors between the TE and TM cases were no larger than $\sim$20%. For the 2D multilayer models, the calculated heating factors were on the order of $\sim$0.02 $^\circ$C W$^{-1}$·m$^{-2}$ at frequencies of $\geq$10 GHz. For the 3D forearm model, when the antenna–forearm distance was shorter, the EM power was more narrowly distributed, and higher heating factors were observed. In such cases, the coupling effects between the antenna and the body are predominant under exposure to a reactive near field (Funahashi et al. 2018). This coupling effect generally depends on the antenna design, frequency, and separation distance. If the antenna mismatch is considered, the effect is mitigated owing to the reduced output power (Colombi et al. 2018). With an increase in the distance, the heating factors become comparable to those for 1D simulations. This is consistent with the results of Hashimoto et al. (2017), who reported that the heating factors for beams with diameters larger than the side length of the averaging area are comparable to those for incident plane waves. In general, the calculated heating factors were in good consistence with $<0.025$ $^\circ$C·m$^2$ W$^{-1}$, which is suggested by ICNIRP-2020 guideline (ICNIRP 2020).

In this study, the effect of the incidence angle was not considered. Owing to the rotational symmetry of the multilayer models, the results of our 2D simulation were independent of the incident angle of the exposed plane wave. Nakae et al. (2020) also found that the heating factors for the APD are insensitive to the incident angle.

In conclusion, the averaged APD and temperature rise in body models with nonplanar surfaces under EM exposure at $\geq$6 GHz were evaluated. Different calculation schemes for the averaged APD were investigated. To reduce the uncertainties in the FD bioheat calculations caused by the staircasing artifacts at the model boundaries, we proposed a new local compensation method for correcting the heat convection rate. This compensation method was validated by analytical solution using a sphere model and by FEM solutions using prolate ellipsoidal models with different tile angles and spatial resolutions. It was demonstrated that the proposed method can significantly improve the thermal calculation at stepped boundaries, regardless of the model resolution and rotation. The APDs and temperature rises for 2D cylindrical models and a 3D forearm model were then evaluated at frequencies ranging from 6 to 60 GHz using the APD schemes and the heat convection compensation method. When the model curvature radii were larger than $\sim$30 mm and the frequency was above $\sim$20 GHz, the calculated heating factors agreed well with those obtained in previous studies using planar models, and the differences in the heating factors among different APD schemes were marginal.
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ORCID iDs

Yinliang Diao  https://orcid.org/0000-0002-6492-4515
Essam A Rashed  https://orcid.org/0000-0001-6571-9807
Akimasa Hirata  https://orcid.org/0000-0001-8336-1140

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