Waveguide Applicator System for Head and Neck Hyperthermia Treatment

Ondrej Fiser[†], Ilja Merunka* and Jan Vrba*

Abstract – The main purpose of this article is a complex hyperthermia applicator system design for treatment of head and neck region. The applicator system is composed of four waveguides with a stripline horn aperture and circular water bolus. The specific absorption rate (SAR) and temperature distribution from this applicator in various numerical phantom models was investigated. For used targets, the treatment planning based on the optimization process made through the SEMCAD X software is added to show the steering possibilities of SAR and thereby temperature distribution. Using treatment planning software, we proved that the SAR and temperature distribution can be effectively controlled (by amplitude and phase changing) improving the SAR and temperature target coverage approximately by 20 %. For the proposed applicator system analysis and quantitative evaluation of two parameters 25 % iso-SAR and 41°C iso-temperature contours in the treatment area with the respect to sensitive structures in treatment area were defined. To verify our simulation results, the real measurement of reflectivity coefficient as well as the temperature distribution in a homogenous phantom were performed.

Keywords: Head and neck cancer, Hyperthermia system, Hyperthermia treatment, Oncology, Phased array

1. Introduction

Head and neck cancer is the sixth most common cancer type in the European Union [1]. This is due to smoking and excessive alcohol consumption [1]. Other carcinogens causing cancer in head and neck region are human papillomavirus (HPV) and Epstein-Barr virus (EBV) [1], [2]. The frequently occurring head and neck cancers, which are appropriate for microwave hyperthermia treatment, are tumors of the vocal cords, thyroid gland and adjacent lymph nodes [2]. Curability depends on used frequency and depth of the tumor under the skin [3]. The clinical trials (phase III) show that a combination of hyperthermia treatment, radiotherapy [4] and chemotherapy [5] improves the survival chance of patients with head and neck tumors [6, 7] and [8]. Microwave hyperthermia belongs to nontoxic oncological treatment using frequencies 60-120, 434 and 915 MHz [3, 9] and [10]. The goal of the hyperthermia treatment is to achieve a temperature range of 41-45 °C (for 40-90 min or longer) in tumor tissue [3, 11] and [12]. This temperature leads to tumor cell apoptosis or it can cause greater cell sensitivity to ionizing radiation [3].

To improve treatment of malignant tumors in relatively complicated head and neck region, a semicircle applicator

system composed of four equal waveguide applicators was designed. In case of the using multiple coherent applicators utilization, it is needful to plan the treatment through the optimization of specific absorption rate (SAR) distribution [13] for each particular specific patient case. As we know, some circular phase array systems intended for head and neck hyperthermia treatment are currently under development (e.g. Paulides et al. (2007) [6] and Trefna et al. (2010) [14]).

The main purpose of this article is to introduce and show specific advantages of a novel approach of hyperthermia treatment in head and neck region. We numerically and experimentally evaluated the microwave focusing to targets of various sizes. As a suitable working frequency with the respect to the SAR steering ability in head and neck region, the frequency of 434 MHz was chosen.

2. Waveguide Applicator Description

The presented applicator system is composed of four waveguide applicators. The advantage of the waveguide type applicator is the ability to transfer maximum energy without energy losses and without irradiation to the surrounding area [15, 16] and [17]. The mentioned waveguide applicator is designed for local hyperthermia treatment. In order to minimize the dimensions of the whole applicator, the waveguide is filled with distilled water and is terminated by a strip line horn aperture. The reason and purpose of dielectric sidewalls (strip-line

[†] Corresponding Author: Dept. of Electromagnetic Field, Czech Technical University in Prague, Czech Republic. (fiseron2@fel.cvut.cz)

* Dept. of Electromagnetic Field, Czech Technical University in Prague, Czech Republic

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Table 1. Waveguide dimensions (see symbols in the Fig. 1 a)

Symbol	Dimensions [mm]
L_a	50
L_b	50
a	58
b	29
A	100
B	50

(a)

(b)

(c)

Fig. 1. Waveguide applicator with dimension symbols: (a), applicator from the front; (b) and detail of applicator feeding with tuning screws (c)

aperture) is to increase the effective aperture and field homogeneity in front of the applicator. The described applicator is in the Fig. 1(a) side view, (b) front view and (c) detail of applicator feeding with tuning screws). The applicator operates on frequency 434 MHz.

The material used for the waveguide applicator is aluminum while the sidewalls of the stripline horn are made with acrylic glass. The dimensions of the waveguide are designed for excitation of the dominant mode TE_{10} at working frequency (see Table 1). To match the applicator impedance to the coaxial cable and phantom, two impedance transformers at a distance of $\lambda/4$ and $\lambda/8$ (where λ is wavelength in waveguide) from the coaxial waveguide transition probe are placed. Thanks to that we are able to have applicator adaptability for all types of phantoms.

3. Applicators Configuration

We studied the amplitude and phase impact of used waveguide applicators on the electromagnetic field stratification, cf. our previous research (published in [15, 16] and [17]). A higher number of applicators, e.g. 7 (analyzed in [17]), led to the creation of unwanted hotspots close to sensitive tissues (e.g. spinal cord and

Fig. 2. Applicator system setup with water bolus

brain tissue). Therefore we redesigned the applicator system by reducing the number of applicators (from seven to four). This modification led to lower hotspot formation risk but approximately 12 % reduction of target coverage. This new applicator system is composed of four waveguide applicators. The applicators-phantom distance is 77 mm. This distance maintains relatively high sensitivity for field steering by phase and amplitude changing. Applicators array can be rotated around the patient's head and neck for optimal position and ideal coverage of the treated tissue by the electromagnetic power, and subsequently by temperature. The angle among applicators is 13 degrees (see Fig. 2). The penetration depth of all four applicators (in case of same amplitude and phase excitation) is 4.8 cm corresponding to 75% decrease of maximum SAR. For comparison: one single applicator is of 2.4 cm penetration depth only. The applicator setup with water bolus is obvious from Fig. 2.

4. Water Bolus

Between patient and applicators, a circular water bolus is placed. Fig. 2 shows the applicator system with circular water bolus which is filled with distilled water (tempered to constant temperature 21 °C). The water bolus plays two substantial roles. The first one is to cool the patient's skin, through this avoiding hot spot formation. The second one is a coupling media between applicator and phantom. Circular water bolus is of 380 mm in diameter and 200 mm in height. Size and shape of the water bolus is affecting two important system parameters-namely impedance matching and field homogeneity.

5. Numerical Phantoms

For the design and testing of an applicator system, three numerical phantoms were used. Within these phantoms, the ability of electromagnetic field focusing into various treatment targets and consequently temperature distribution was studied. Dielectric and thermal parameters of all phantoms were taken from the IT'IS foundation materials

Table 2. Targets volume summary

Target		Volume [cm ³]
1	Tumor I	12.5
2	Tumor II	16.4
3	Thyroid gland	10.6
4	Tongue	60.6
5	Tumor III	246

Fig. 3. Homogenous phantom cross-section with applicator array: (a) and real modulus of E-field (b)

database [18] while the working frequency is being 434 MHz. In the Table 2, the summarizing information of targets volume is shown. To meet input parameters read Table 3.

5.1 Homogenous phantom

Homogenous phantom was used for the applicator position proposal i.e. the applicator-phantom distance determination, shape of the water bolus with the respect to the best possible field steering sensitivity. The circular homogenous phantom is composed of 2 different parts.

Dimensions of heterogeneous phantom are 160 mm in diameter and 100 mm in height. In phantom an artificial cylindrical tumor (20 mm in diameter) is placed 10 mm under the surface. In the Fig. 3 (a) there is the cross section of the system with homogenous phantom and the Fig. 3 (b) the distribution of E-field in water bolus and phantom with no optimization is shown.

5.2 Human model Ella

We used “human model part” of the SEMCAD X models database [18]. For our purposes, the human model with name Ella from Virtual Family 1.2 was utilized. Around the neck we placed proposed applicator system with water bolus. Using this numerical model we choose three different targets:

- Thyroid gland
- Tongue
- Tumor II (18 mm under skin)

Each target has a specific shape and position (see Fig. 4

Fig. 4. Numerical Human model Ella: (a) targets and (b) applicators position **Fig. 5.** Numerical model of real patient with tumor: (a) and applicators configuration (b)

(a). The detail of the array localization is obvious from Fig. 4 (b).

5.3 Real Patient Model

The function of our proposed array of applicators we verified using the model of a real patient that was adopted from [19]. The volume and size of tumor is so large that it cannot be appropriately covered by a single aperture only [17]. Our proposed applicator system is more convenient to be used in this case. The 3D numerical patient model was reconstructed from CT scans by semi-automatic segmentation toolbox iSeq integrated in SEMCAD X. In the Fig. 5 (a), the resulting numerical model is shown. Fig. 5 (b) shows the placement applicator system around the patient's neck. Against applicator positions in Fig. 4 (b) all applicators are rotated by 40 degrees to show positioning options of heating system.

6. Simulation and SAR Optimization

The effectiveness of the new applicator system is shown through the SAR and temperature distribution calculated by SEMCAD X 14.8 [20]. SAR distribution is useful especially for technical development and comparison of the discussed applicator system. From a clinical point of view it is better to utilize the temperature simulations. The thermal simulation keeps us informed about temperature distribution. The temperature simulation enables us to consider parameters like thermal conduction, blood perfusion and bolus cooling.

Two parameters SAR_{25} and T_{41} are used for the evaluation of gained results. SAR_{25} is related to absorption

energy and T_{41} is related to temperature distribution. SAR_{25} is a volume where the SAR is greater than 25 % of its maximum in the considered target (according to the Eq. 1). T_{41} is the volume of temperature 41-45 °C in the heated target (Eq. 2).

$$SAR_{25}[\%] = \frac{V_{target}(SAR > SAR_{max} \cdot 0.25)}{V_{target}} \cdot 100\% \quad (1)$$

where

SAR_{max} is maximum SAR in the whole model and V_{target} is the volume of treated target.

$$T_{41}[\%] = \frac{V_{target}(T > 41^{\circ}\text{C})}{T_{target}} \cdot 100\% \quad (2)$$

where

T is temperature.

6.1 Treatment planning (SAR optimization)

Hyperthermia treatment planning is necessary activity improving the quality of the hyperthermia treatment (similarly like in radiotherapy). For more complex system composed of more than one applicator (like our applicator system) it is desirable to perform treatment planning with field optimization. The planning system predicts the SAR and temperature distribution in the patient, optimizes the system properties and through this it improves energy delivery to the tumor, reduces healthy tissue hotspots, improves the quality and general success of the hyperthermia treatment. An optimized SAR distribution can be reached by tuning the phase and amplitude of the applicator. For evaluation of our system, the planning software (SAR optimizer) as a part of simulator EM-field

SEMCAD X based on Genetic algorithm was used. The optimization process is based on the following cost function L (3) adopted from [20]:

$$L = \frac{\int_{target} w(x, y, z) \cdot SAR(x, y, z) dV}{\int_{all-target} w(x, y, z) \cdot SAR(x, y, z) dV} \quad (3)$$

where $w(x, y, z)$ is a weighting parameter. The weighting function w determines the priority and sensitivity of target against the whole phantom volume. The algorithm is searching for maximum of used cost function L .

6.2 Thermal simulation

SEMCAD X simulator is also appropriate for thermal simulations [20]. Thermal solver uses results from the optimized SAR distribution as a heat source through Penne's Bioheat equation. In the beginning of heating, all tissue temperatures were set to 37 °C and the temperature of water bolus was set to 25 °C. The value of the heat transfer coefficient between human body and circulated water was set to 65 W/m²/K [21]. Within the results of temperature distribution, the 41°C iso-volume is marked (like lower limit of treatable temperature).

Temperature parameters used in temperature calculations (Specific heat capacity, Thermal conductivity and Heat generation rate) are listed in Table 3. Blood circulation in arteries and veins are also involved in the calculations.

7. Simulation Results

In Fig. 6 and 7, the results from numerical simulations

Table 3. Dielectric and thermal properties of numerical phantoms and parts of waveguide applicator for frequency 434 MHz [18]

Material	Relative permittivity ϵ [-]	Conductivity σ [S/m]	Specific Heat Capacity [J/kg/K]	Thermal Conductivity [W/m/K]	Heat Generation Rate [W/kg]
Artery	61	1.58	3617	0.51	-
Bone	13.1	0.09	1312	0.32	0.15
Water bolus	81	0	-	0.56	-
Fat	11.6	0.08	2348	0.21	0.51
Muscle	56.9	0.79	3421	0.49	0.96
Skin	40.9	0.89	3390	0.37	1.65
Tumor	59.5	0.89	2654	0.27	1.53
Trachea	43.9	0.8	3568	0.48	0.54
Thyroid gland	61.3	0.97	3421	0.49	87.1
Esophagus	64.8	1.23	3500	0.52	2.94
Vertebrae	13.1	0.09	1312	0.32	0.15
Spinal cord	32.2	0.59	3630	0.51	2.48
Larynx	45.1	0.59	3568	0.48	0.54
Intervertebral disc	45.1	0.59	3560	0.49	0.54
Vein	61	1.59	3617	0.52	-
Cerebrospinal fluid	70.6	2.25	4095	0.57	-
Connective tissue	47.1	0.56	2372	0.39	0.58
Teeth	13.1	0.09	1255	0.59	-
Nasal cavity	45.71	0.88	3150	0.34	-
Tongue	5.74	0.78	3568	0.49	1.21

are presented. Using these results we investigate applicator system ability to focus the microwave energy to the treated

Table 4. Phase (P, degrees) and relative amplitude (A,-) setting for each applicator for all targets

Nr. App	Target							
	Tumor I		Tumor II		Thyroid gland		Tumor III	
	A	P	A	P	A	P	A	P
1	0.39	0	0.87	0	1	0	0.41	0
2	1	-56.3	0.93	-38.3	0.83	64.9	1	-69.2
3	0.82	-70.7	1	-0.6	0.88	8.47	0.99	-59.3
4	0.55	-6.5	0.87	-8.7	0.98	42.4	0.64	-67.8

target. This energy is manifested by temperature increasing. In Fig. 6, the phantom SAR distribution is shown in the first and second columns. The third and fourth columns show us the T_{41} iso-volume contour coverage. In all cases the output power of each applicator was set in order not to exceed the 45 °C temperature maximum in the whole phantom. Fig. 6 a) demonstrates tongue, b) tumor II, c) thyroid gland and d) tumor III, all as for the treated area. For easier comparison all SAR values shown in figures are normalized to total input power 1 W. In the Fig. 7 there is a SAR-Volume histogram of targets coverage. This figure is obtained from optimized SAR simulations and is used to determine the value of SAR_{25} volume cover. The dashed

Fig. 6. Optimized SAR distribution in transversal (index 1), longitudinal cross sections (index 2) and temperature iso-surface T_{41} (index 3 and 4); tongue (a), tumor (b), thyroid gland (c) and real tumor (d)

Fig. 7. Cumulative optimized SAR histogram

Table 5. SAR₂₅ and T₄₁ iso-contour target volume coverage

Target	SAR ₂₅ [%]	T ₄₁ [%]	No optimized SAR ₂₅ [%]
Tumor I	85	95	70
Tumor II	80	90	58
Thyroid gland	87	92	75
Tongue	21	51	6
Tumor III	54	80	36

vertical line indicates the value of 25 % residual SAR. The light blue curve corresponds to SAR distribution in the spinal cord (temperature sensitive tissue).

In Table 5, target coverage is shown. The best cover results were achieved through Tumor I (SAR₂₅ 85 % and T₄₁ 95 %), Tumor II (SAR₂₅ 80 % and T₄₁ 90 %) and thyroid gland (SAR₂₅ 87 % and T₄₁ 92 %). On the contrary we did not achieve satisfactory SAR and temperature tongue cover (SAR₂₅ 21 % and T₄₁ 51 %). The applicator system is effective also for the larger tumor III (SAR₂₅ 54 % and T₄₁ 80 %). Because the tumor is located on the surface it is possible to use a higher power and cooler water bolus for good T₄₁ coverage. In Table 4 the applicators settings for each target are listed. For comparison we added to Table 5 SAR₂₅ in the non-optimized case. One can see that the values are significantly lower. That means that the optimization increases the power concentration in the treated area.

8. Measurement (Experiments)

8.1 Agar phantom for verification

For measuring purposes, a cylindrical homogeneous (160 mm in diameter) phantom with an inserted cylindrical tumor (20 mm in diameter) was made. Phantom ingredients and their quantities are listed in the following Table 6. A completed phantom can be seen in Fig. 8 (a).

For determination of complex permittivity of a real agar phantom the method based on reflection coefficient measurement by coaxial probe was used [23]. This method is applicable in a frequency range from 30 MHz to 1 GHz. The results of measurement are listed in Table 7.

Table 6. Ingredients quantity of healthy [3] and tumorous tissue phantom [22]

Substance	Healthy tissue [%]	Tumorous tissue [%]
Deionized water	95	66
Agar	4	3
NaCl	0.35	0.5
Food color	0.65	0.5
Ethanol	-	30

Table 7. Measured dielectric parameters of heterogeneous phantom for 434 MHz

Type of tissue	Relative permittivity [-]	Conductivity [S/m]
Muscle phantom	60.79	0.58
Tumor phantom	62.26	1.01

Table 8. Applicator amplitude and phase settings for measurement

Applicator number	Amplitude [-]	Phase [degrees]	Cable Extension [cm]
1	0.39	0	9.03
2	1	-56.3	1.84
3	0.82	-70.7	0
4	0.55	-6.5	8.20

Fig. 8. Photo of the laboratory prototype, top view (a) and measurement setup (b)

8.2 Laboratory prototype

For the laboratory prototype production we used a special plastic container with required dimensions. The applicators at desired positions were placed at the container wall. The container and applicators are filled by deionized water. The homogenous phantom is placed into the container (see Fig. 8 (a) and (b)). For the correct phase and amplitude settings of applicators, the pre-calculated lengths of RF coaxial cables of type RG58 were used (see Table 8). Relative permittivity of dielectric material of this type of cable is 2.26 for 434 MHz.

The wavelength of coaxial cable determines the phase shift. The phase shift 2π corresponds to λ_{cable} that is $\sqrt{\epsilon_{rc}}$ times lower than the vacuum wavelength (Eq. 4).

Fig. 9. Simulated and measured return loss

$$\lambda_{\text{cable}} = \frac{c}{f \cdot \sqrt{\epsilon_{rc}}} = \frac{299792458}{434 \cdot 10^6 \cdot \sqrt{2.26}} = 0.4598m \quad (4)$$

where

c is speed of light in vacuum, f is working frequency, ϵ_{rc} is relative permittivity of coaxial cable. Calculated wavelength in coaxial cable (λ_{cable}) is 45.98 cm.

8.3 Experiment 1, reflection coefficient measurement

In experiment 1 we evaluated impedance matching of each applicator separately and subsequently of the whole system. The analysis was made by vector analyzer of the Agilent E5062A type in the frequency band from 300 to 600 MHz. In the Fig. 9 there are results from this measurement (blue dashed curve) compared to results from numerical simulation (green curve). These curves represent return loss of whole system when all applicators were turned on. The main cause of differences in resonances in curve of reflection coefficient are caused by deviation in phantom dielectric parameters. The matching of the whole system is under -10 dB for 434 MHz and this is satisfactory.

8.4 Experiment 2, temperature distribution in homogenous phantom

In this experiment the distribution of absorbed power through temperature in a homogeneous phantom was measured. The main goal of power measurement is to verify energy focusing and heating ability of whole applicator system. SAR and temperature are related to the following relation (Eq. 5) adopted from [3]:

$$SAR = \frac{\sigma}{2\rho} |E|^2 = c \cdot \frac{\Delta T}{\Delta t} \quad (5)$$

where SAR is the specific absorption rate [W/kg], σ is tissue conductivity [S/m] at working frequency, ρ is tissue density [kg/m³], E is vector of electrical intensity field [V/m], c is specific heat capacity [J/kg/K], T is temperature

Fig. 10. Measurement setup with two power dividers

Fig. 11. Numerical simulation of temperature distribution in homogenous phantom (cross section)

[°C] and t is time [s]. The Eq. (5) is valid only for experiment of short duration, because in case of experiments with longer duration the thermal conductivity must be involved (e.g. Penne equation).

In the Fig. 10 the measurement setup with a microwave generator and applicator system is shown. The temperature of the phantom was 22 degrees and the temperature of water bolus (deionized water) was 21 degrees. Waveguide applicators were powered by the generator “UHF-POWER-GENERATOR PG 70.150.2” (it has two separated synchronized power output channels) at frequency 434 MHz and of total power of 150 W for 8 minutes. Applicators 1 and 4 were each excited by 25 W power (output B), applicators 2 and 3 each with power of 50 W (output A). In the Fig. 11 the longitudinal cross section of numerical temperature simulation through homogenous phantom is shown. The simulation settings correspond to conditions during measurement. Phantom thermograms (Fig. 12) show temperature distribution in homogenous phantoms. Energy focus is in the place where the tumor is located (black frame). Thermograms of the phantom were taken by IR camera type Flir P25.

8.5 Results verification

For verification of results obtained from numerical simulation, we use temperature distribution in a homogenous phantom (simulation results Fig. 11). Utilization of parameter SAR₂₅ is inaccurate in this case, because the time of exposure was longer (8 minutes). For this occasion the parameter T₄₁ was chosen. In clinical use there is an effort to increase the temperature of tumorous tissue from 37 °C to maximum 45 °C (the temperature difference is

Fig. 12. Thermograms after power deposition, front view (a) and cross section (b)

8 °C). In our capabilities it was not possible to have a thermally tempered phantom at 37 °C but only at 22 °C. To keep the same conditions, our maximum temperature in the phantom was 30 °C. A temperature of 26 °C corresponds to 41°C used in simulations. In numerical simulation, 95 % of the whole tumor volume is covered by temperature 26 °C in this case. In the thermogram (Fig. 12 (b)) we calculated the temperature as 26 °C for the volume cover 88 %. The measurement deviation from the numerical simulation is 7 %.

Generally speaking, temperature increased by 7-8 degrees inside the tumor phantom in 8 minutes. This temperature increase is more than sufficient for high quality hyperthermia. Outside the tumor phantom there was a gentle increase in temperature (not greater than 3 degrees).

9. Conclusion

In this article we designed, evaluated and analyzed (numerically and experimentally) a new applicator system intended for focused head and neck hyperthermia therapy. First, according to our previous research, we proposed four waveguide applicator system working on frequency 434 MHz. For evaluation of predicted SAR and temperature distribution in a phantom two volume parameters (SAR_{25} and T_{41}) were defined and used. Performance of the new applicator system can be also improved by the treatment planning software based on the Genetic algorithm through the distribution of SAR and temperature in the target region. The system is designed with the respect to higher sensitivity of E-field pattern to phases and amplitudes of each applicator feeding. Our simulations proved that the SAR_{25} is approximately by 20 % higher in a treatment target due to optimization. We also observed lower risk of hotspots formation. These benefits permit energy focusing and selectively the tumor heating with the respect to healthy and sensitive tissues.

To compare our system with other methods, we made a quantitative comparison with the system of Paulides et al.. In mentioned article [6] the circular system on a numerical 3D human model was tested. The analysis was performed numerically (using 3D model with lymphonode and small

tumor) and also by measurement. The targets are relatively similar with our Tumor II (especially in volume). Optimized results in [6] show that the lymphonode and tumor SAR_{25} coverage is approximately 85 %. With our presented system we reached SAR_{25} coverage for Tumor II of 80 %. From this comparison it is clear that our system has similar results as like other systems.

The originality of our approach is based on the fact that our applicator system uses a limited number of applicators only. This solution, of course, reduces complexity of the whole system, cost of components and it is easier to utilize it in therapy (with lower risk of unwanted hotspots).

A disadvantage of this system, which we recognize, is fixed attachment to patient's body. This is, of course, reducing the patient's comfort. Next problem is higher energy absorption on the body surface. The main solution helping us to improve temperature distribution and decreasing absorption on patient surface depends on the water bolus temperature. By the water bolus temperature decreasing we were able to supply greater energy to the target without hotspot creating on the body surface. Respecting the penetration depth and the reflection coefficient the ideal water bolus temperature is between 20 – 30 °C.

For experimental results verification the laboratory prototype was constructed. The reflection coefficient of four waveguide applicators and temperature distribution was measured.

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Ondrej Fiser received his B.S. and M.S. degree from Czech Technical University in Prague, 2011 and 2013 respectively. At present, he is a Ph.D student at Department of Electromagnetic Field (CTU in Prague). His main research interests include micro-wave hyperthermia and microwave imaging.

Ilja Merunka received the B.S. and M.S. degree in biomedical engineering from the Czech Technical University in Prague. He is working towards Ph.D. degree in radioelectronics at Department of Electromagnetic Field (CTU in Prague). His research interests include microwave hyperthermia, microwave

tomography and numerical simulations.

Jan Vrba received the M.Sc. and Ph.D. degrees from the CTU in Prague (1972 resp. 1976) and since 1993 he is a Full Professor of Radioelectronics at the Dept. of Electromagnetic Field, CTU in Prague. In the period 1972 till 1981 his research was oriented on parametric amplifiers and microstrip resonators. Since 1981 his research activities are focused on medical and industrial applications of microwave technology. He was a member of the team which developed the first system for microwave hyperthermia in cancer treatment in the former Czechoslovakia. He is a member of IEEE, ESHO (European Society for Hyperthermic Oncology) and Electromagnetic Academy. He was awarded by ESHO-Pyrexar Award in 2015.