

Thermal Lesion of Renal Tumour as a Function of Catheter Material Property

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Abstract—The extent of lesion achieved during microwave ablation is dependent on some factors which include the time period of application, its intensity, antenna geometry, and relative permittivity of the tissue. Several studies have been conducted on microwave ablation for the treatment of tumours and have focused on different antenna geometries, its intensity, and time of application. This work seeks to find a correlation between the relative permittivity of the catheter and the temperature distribution which determines necrosis of the tissue by using Tefzel ETFE, Teflon FEP, PFA Teflon type A, PFA Teflon type B, Teflon AF, and PTFE Teflon type B while modelling using COMSOL Multiphysics. The extent of the thermal lesion was observed to be dependent on the relative permittivity of the catheter material, with Tefzel ETFE giving the best performance and Teflon AF providing the least.

Index Terms—Microwave Ablation; Necrosis; Relative Permittivity.

I. INTRODUCTION

Microwave ablation and radiofrequency ablation, the two most commonly used heat ablation which are minimally invasive procedure are been increasingly used for the treatment of tumours and cancerous cells based on the principles of hyperthermia using image guidance systems for antenna insertion. Necrosis is induced by hyperthermia as a result of protein coagulation and denaturation. Normothermia, the normal human body temperature falls within the range of 36.5-37.5°C [1], and body temperatures a few degree Celsius above this can be lethal. The inability of tumours to cause an increased blood flow in response to heat application as does a normal cell makes them act as heat sinks and thus more vulnerable to damage is exploited for its treatment [2].

Different pieces of literatures have given different values of temperature as lethal to human cells; however, there seems to be a pointer to a value above 42°C [2]-[5].

The application of heat for treatment especially for the case of the dreaded cancer cells and for pain relief although of recent gaining relevance have been in existence for a very long time and this can be attested to from the quotes of Hippocrates “Those who cannot be cured by medicine can be cured by surgery. Those who cannot be cured by surgery can be cured by fire (heat). Those who cannot be cured by fire, they are indeed incurable” Hippocrates (479 – 377 B. C.)

As good and effective hyperthermia is in the treatment of tumours, care must be taking in its administration as studies have shown that mild thermo-tolerance can set in if cells are heated to a temperature of 40°C for about three hours [2]. This could, however, also be put to an advantage by making the surrounding cells of a tumour site to develop thermo-tolerance, and are made to survive the subsequent application of lethal temperatures to the tumour site.

Microwave ablation is usually favoured over radiofrequency ablation as it is believed to be less dependent on electrical conductivities of tissues and provides consistent higher intra-tumour temperatures making it more effective for the treatment of larger sized tumours as a result of larger ablation zones [3],[6]-[12].

Microwave ablation involves the use of electromagnetic radiations having a wavelength range of between one millimetre and one meter delivered through special purpose antennas and are capable of causing rapid temperature increase adequate enough to achieve necrosis [3],[13]-[16]. The theory of the concept of microwave ablation is available in many pieces of literatures, and basically this entails the use of electromagnetic field to cause water molecules in tissues to oscillate and the subsequent absorption of the electromagnetic energy and its conversion to heat [3],[10],[14], [17]-[18]. It has evolved as a modality for the treatment of large soft tissues in short time periods.

Renal microwave ablation using image guidance systems are offering a viable alternative to partial nephrectomy just like other microwave application modes, especially where small tumours are involved, [10] as it allows for faster recovery period, offers a lifeline to patients who are not amenable to surgery due to one or more reasons, and at a reduced cost [3],[11],[14],[19]-[23].

The extent of lesion achieved during microwave ablation is a function of its propagation, and is determined by the time of application, the intensity, antenna geometry, and the relative permittivity of the tissue [3],[15],[17],[23]-[25] and are now the standard option for tumour size less than 7cm [26]. However, with larger sizes of tumours, multiple antennas can be employed to achieve greater ablation zone [27]-[28].

Different antenna geometries have been proposed and studied by many scholars, they are thin co-axial based and have been categorised basically as monopole, dipole, and slot [9],[13],[15],[23],[29]-[30]. Undertaking studies on monopole, single slot, double slot and sleeved antennas, it was observed that the sleeved antenna type showed the best performance in terms of ablation sphere shape index and temperature distribution [13]. The antenna geometry is always done in a way as to fulfil some basic requirements of being minimally invasive and highly efficient [31].

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The typical frequencies used for microwave ablation as adopted by the Federal Communications Commission are 915 MHz and the 2.45 GHz [3],[12]-[15],[17],[22]-[23],[28], these are so chosen to avoid interference with other waves emitting and absorbing devices/equipment.

The efficacy of microwave ablation antennas is determined by the specific absorption ratio (SAR) which is the electromagnetic power deposited per unit mass of the tissue [9], and the ablation zone ideally is expected to be spherical [3],[31]. The catheter material and its dimensions also contribute to the antenna performance.

To ensure better prediction and production of spherical ablation zones, a new technology “the Thermosphere” have been developed and it relies on three modes of energy control: wavelength, thermal and field controls [18],[25].

Studies carried out on microwave ablation heating pattern suggest that there are changes in SAR at elevated temperatures (above 100°C) as it was experimentally observed, however, this was not the case for simulation studies mainly due to the fact of the non-proper capturing of the temperature dependence nature of the dielectric properties of tissues at elevated temperatures [12].

This study is focussed on the extent of thermal lessen achieved with the use of different materials as the catheter in microwave ablation.

II. MATERIALS AND METHODS

The numerical modelling approach developed for cancer microwave therapy by COMSOL [32] was adapted for the coaxial slot antenna geometry.

The simulation of the kidney ablation zone achieved by the antenna was carried out using the radiofrequency and heat transfer modules in COMSOL Multiphysics.

The transverse electromagnetic field characterise the coaxial cable electromagnetic field propagation and are governed by the following equations in a two-dimensional axially symmetric cylindrical coordinates;

$$E(r) = e_r \frac{C}{r} e^{j(\omega t - kz)} \quad (1)$$

$$H(r) = e_\phi \frac{C}{Z} e^{j(\omega t - kz)} \quad (2)$$

$$\text{And } C = \sqrt{\frac{P_a Z}{e_z \pi \ln\left(\frac{r_{out}}{r_{in}}\right)}} \quad (3)$$

where ω is the angular frequency, r_{in} and r_{out} are the inner and outer radii of the dielectric, Z is the wave impedance in the dielectric of the coaxial cable, P_a is the input power, r , ϕ , and z are the cylindrical coordinates centred on the axis of the coaxial cable.

The propagation constant “ k ” also referred to as the wave number = $\frac{\omega}{V}$, and noting that $\omega = 2\pi f$, and $V = f\lambda$, hence “ k ” can be expressed in terms of the medium wavelength as;

$$k = \frac{2\pi}{\lambda} \quad (4)$$

Since the magnetic field is purely azimuthally in direction and the electric field has a finite axial component, the antenna can be modelled using the axial symmetric transverse magnetic formulation with the source modelled using a low-reflecting boundary condition.

$$n * \sqrt{\epsilon E} - \sqrt{\mu H_\phi} = -2\sqrt{\mu H_{\phi 0}} \quad (5)$$

The time-dependent heat transfer equation is governed by the bio-heat equation;

$$\rho C_p \frac{\partial T}{\partial t} + \nabla \cdot (-k \nabla T) = \rho_{bd} C_{bd} \omega_{bd} (T_{bd} - T) + Q_m + Q_e \quad (6)$$

where Q_e is the external heat source, Q_m is the heat from metabolism, T_{bd} is the blood temperature, ω_{bd} is the rate of blood perfusion, C_{bd} is specific heat capacity of blood, ρ_{bd} is the density of blood, and k is the thermal conductivity of kidney.

The heat from metabolism can be neglected, making (6) to become;

$$\rho C_p \frac{\partial T}{\partial t} + \nabla \cdot (-k \nabla T) = \rho_{bd} C_{bd} \omega_{bd} (T_{bd} - T) + Q_e \quad (7)$$

The tissue damage integral during the process was computed using the Arrhenius equation as expressed in (8) and the fraction of necrotic tissue is expressed in (9) ;

$$\frac{d\alpha}{dt} = F \exp\left(-\frac{dE}{RT}\right) \quad (8)$$

$$\theta_d = 1 - \exp(-\alpha) \quad (9)$$

where α is the degree of tissue injury, F is the frequency factor, dE is the irreversible damage activation energy, and θ_d is the fraction of necrotic tissue.

The antenna specific absorption rate (SAR) was computed using;

$$SAR = \frac{\sigma}{\rho} |E|^2 \quad (10)$$

σ is the tissue conductivity and ρ is the tissue density.

The impact of the relative permittivity of the antenna dielectric material on its performance was studied by using available alternative materials to polytetrafluoroethylene (PTFE). The relative permittivity of the materials chosen is stable for a temperature range of (-55 – 125) °C, and hence the simulation was carried out such that the maximum temperature attained was less than 125°C. The properties of the materials used in the simulation are given in Table I. Antenna frequency of 2.45GHz was used and the blood temperature was taken to be 37°C. The other parameters used in the simulation are as shown in Table II.

The simulation does not model the interior of the metallic conductors and the tangential component of the electric field is set to zero to model the metallic parts.

TABLE I: MATERIAL PROPERTIES

Materials	Relative permittivity	Conductivity (S/m)
Kidney	52.7	2.43
Inner dielectric	2.03	

Catheter	
Tefzel ETFE	2.311
Teflon FEP	2.067
PFA Teflon type A	2.049
PFA Teflon type B	2.024
Teflon AF	1.871
PTFE Teflon type B	2.037

TABLE II: BLOOD PROPERTIES

Property	Value
Density	1045 (kg/m ³)
Specific heat	3639 (J/(kg.K))
Perfusion rate	0.0036 (1/s)
Temperature	37 (°C)

The mesh parameters used for the simulation is as shown in Table III.

TABLE III: MESH PARAMETERS

Maximum element size	1.5e-4 m
Minimum element size	2.4e-5 m
Maximum element growth rate	1.3
Curvature factor	0.3

III. RESULTS

Fig. 1, 2, 3, and 4 are the output plots for the temperature distribution, necrotic tissue fraction in surface plot, necrotic tissue fraction at four points; 4 mm, 6 mm, 8 mm and 10mm, and the specific absorption rate at point 2mm from the antenna axis respectively.

The Tefzel ETFE with the highest relative permittivity value was observed to have the highest peak temperature and as depicted in Fig. 1, its temperature distribution as shown by the oval shape for the lethal value is the largest. Teflon AF with the least value of relative permittivity showed the least giving an indication of its dependency on the relative permittivity.

The necrotic tissue fraction achieved during the process was also observed to be best with the use of Tefzel ETFE as is evidenced in Fig. 2 and Fig. 3. The necrotic fraction with the use of Tefzel ETFE was '1' at a radial distance 8 mm from the conductor at a time of twenty-five (25) minutes and 0.6 at a radial distance of 10 mm at time 't' = thirty (30) minutes which from Fig. 3 is the best recorded for any of the catheter materials used in this study. Teflon AF showed to be the least effective of all the catheter materials used.

The specific absorption ratio (SAR) along a line parallel to the antenna and at a distance 2 mm from the antenna axis was also observed to be highest for Tefzel ETFE and least for Teflon AF as evidenced in Fig. 4.

The major objective of microwave ablation therapy is to destroy cancerous cells and tumours, and the extent of its destruction is connoted by the necrotic fraction which is a function of the electromagnetic power deposited per unit mass of tissue.

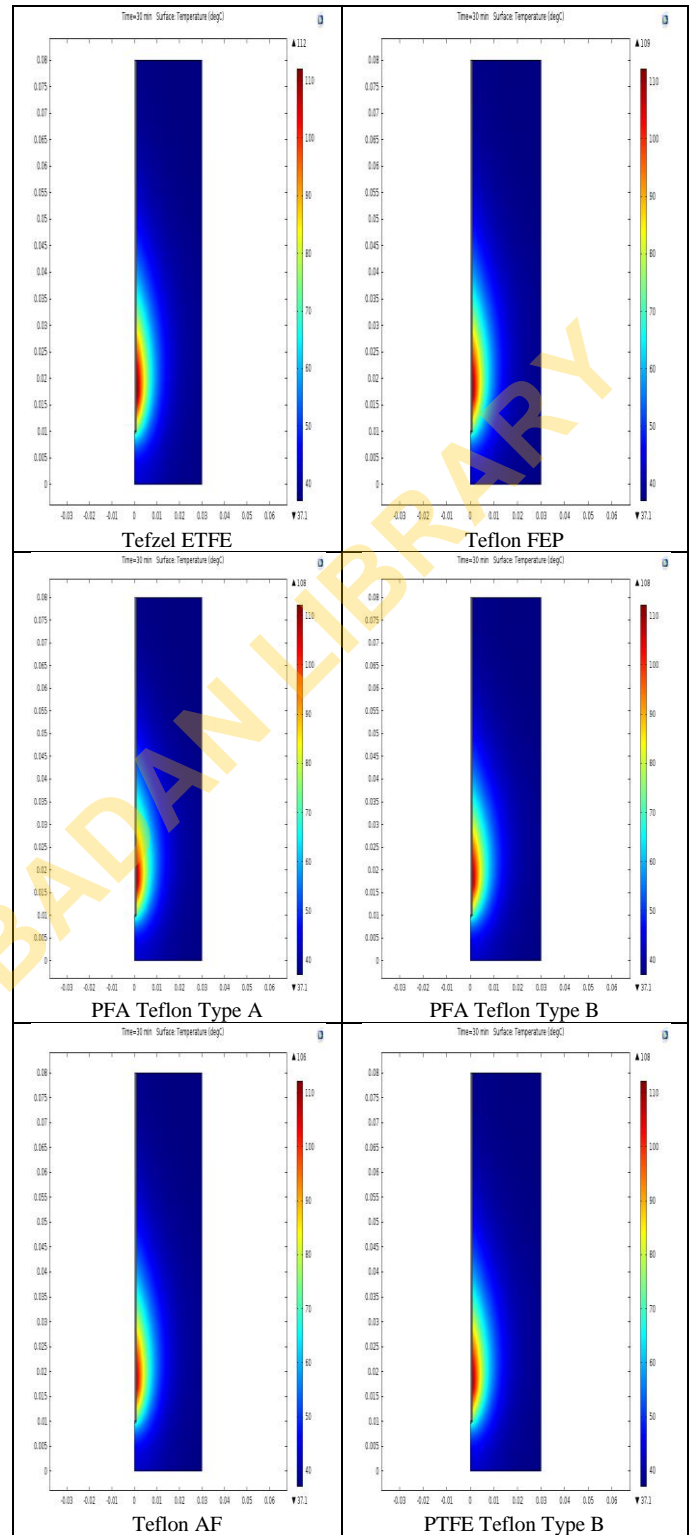


Fig. 1. Temperature distribution in Kidney tissue

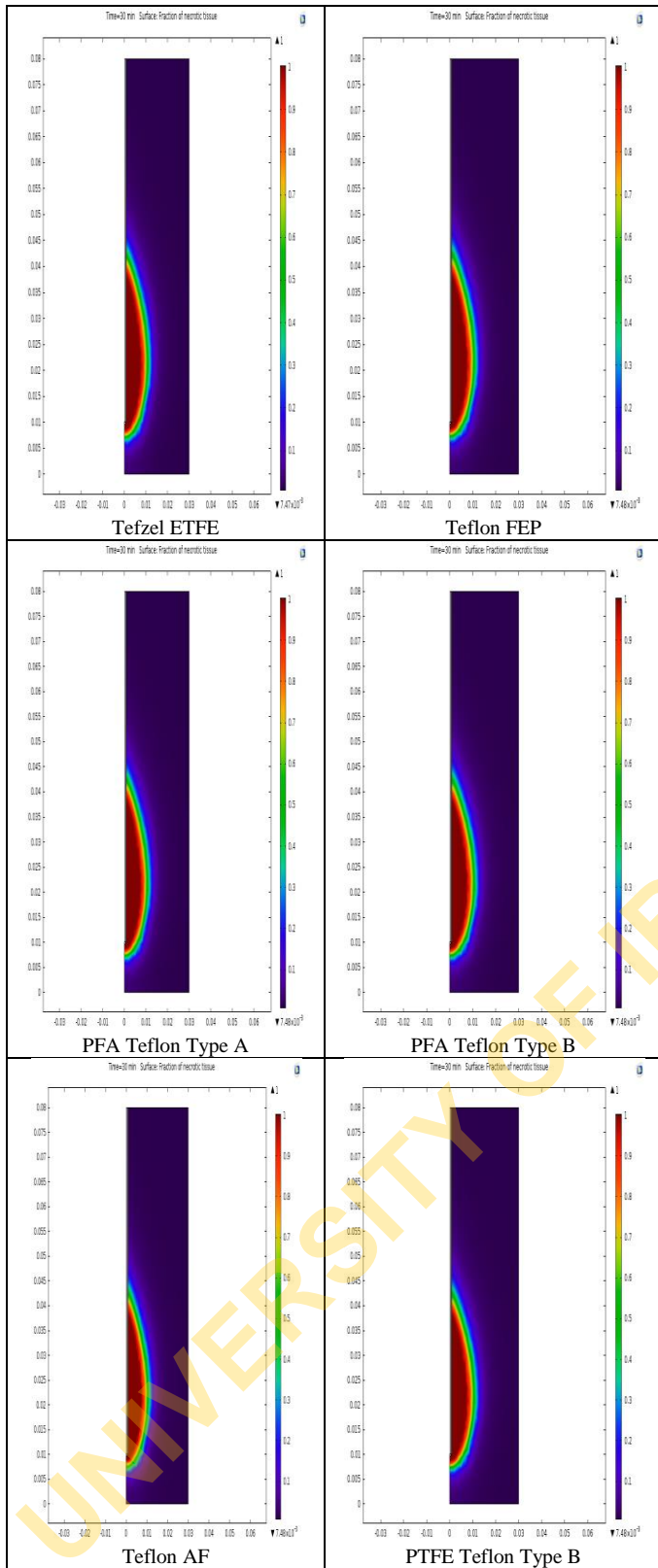


Fig. 2. Necrotic tissue fraction

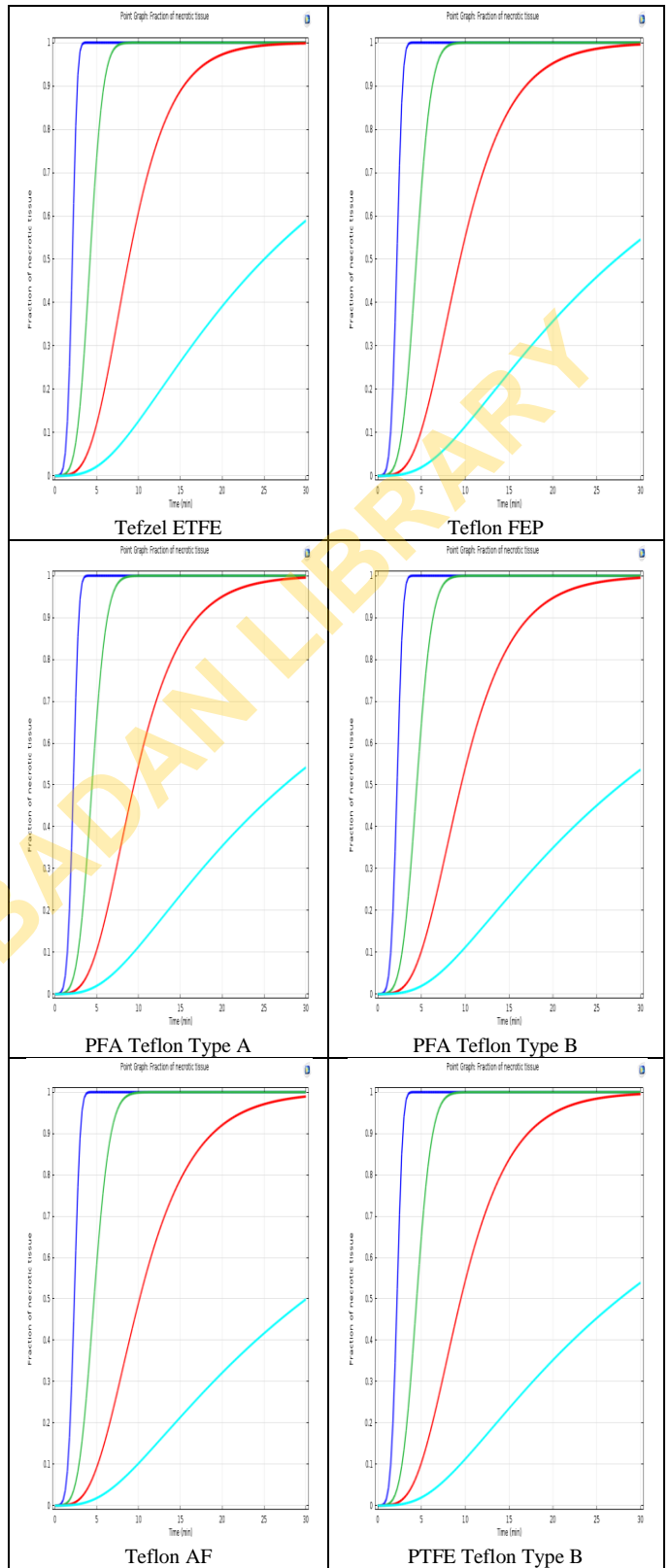


Fig. 3. Tissue necrotic fractions at points 4, 6, 8, and 10 mm

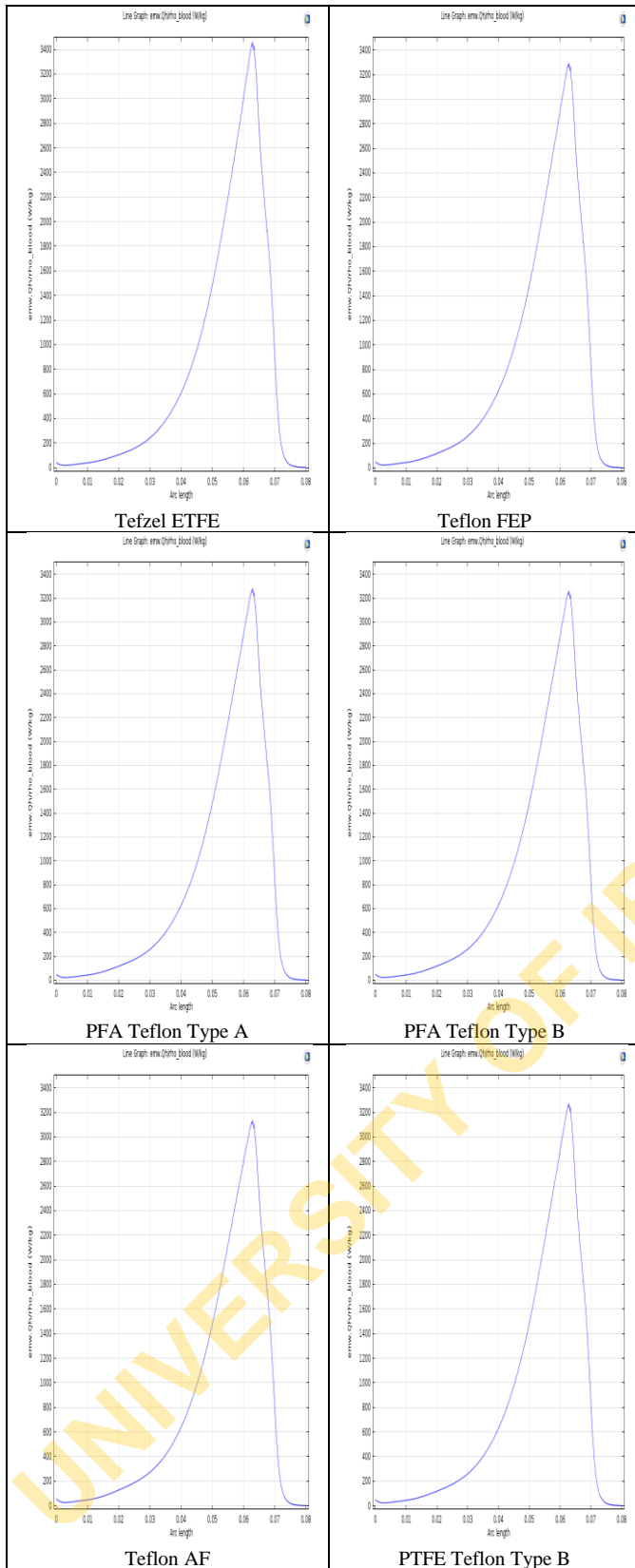


Fig. 4. Specific absorption rate at point 2 mm from the conductor axis

The rate of absorption of input power by tissue to a great level contributes to the determination of the extent of thermal lesion. The extent of absorption is shown in Table IV for all the catheter materials used in this study. The Tefzel ETFE catheter just as for the other considered parameters absorbed the most of the input power, and the rate of this absorption was found to be proportional to the

relative permittivity of the catheter materials with Teflon AF having the least absorption.

TABLE IV: INPUT POWER ABSORPTION RATE

Catheter Material	Tissue Absorption (W)
Tefzel ETFE	9.5555
Teflon FEP	9.4762
PFA Teflon type A	9.4678
PFA Teflon type B	9.4555
Teflon AF	9.3609
PTFE Teflon type B	9.4620

IV. CONCLUSION

The use of Tefzel ETFE as the catheter material has resulted in the best performance of the therapy despite the fact of subjecting all the studies to the same conditions. This is attributable to his superior relative permittivity values of all the materials tested giving an attestation to the dependence of the extent of thermal lesion achieved which is a function of the tissue temperature on relative permittivity of the catheter material.

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