MODELING RADIOFREQUENCY HEATING OF EMBOLIZATION COILS FOR THE TREATMENT OF CEREBRAL ANEURYSMS

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INTRODUCTION

Cerebral aneurysm is a weakness in the wall of a cerebral artery which causes a localized dilation or ballooning of the blood vessel. Untreated aneurysms in the brain may rupture and cause subarachnoid hemorrhage (SAH) and in some cases, stroke.

Surgical and endovascular treatments can be applied. The most popular endovascular treatment is coil embolization which was first introduced by G. Guglielmi in 1991. This method, which is currently used to treat approximately 80% of cerebral aneurysms, consists in inserting a catheter in the aorta and then guiding it to the desired region of the cerebral vasculature to insert small platinum coils inside the aneurysm. These coils induce clotting and occlude the area. A possible problem is that the occluded aneurysm can be recanalized after some months because of unknown reasons [1]. Though the exact phenomenon of recanalization remains unclear, one of the hypotheses is that recanalization originates from the endothelium and that endothelial denudation can prevent recanalization. Some methods of denudation have been previously investigated, such as mechanical abrasion with the aneurismal neck-bridge device (ANBD) and cryoablation. Both methods showed undesired results [2],[3]. The new method which is investigated in this paper is radiofrequency ablation (RFA), which was suggested by J. Raymond [4].

Based on preliminary *in vivo* studies, it is believed that RFA can be effective in endothelial denudation and in improving the results of coil embolization. The main objective of our project is to investigate the effects of radiofrequency current applied to an endovascular platinum coil on the temperature distribution of brain tissue so as to optimize the energy delivery process. To achieve this goal, inductive and resistive characteristics of the embolization coils, as well as the effects of the length and shape of the catheter on the temperature distribution was investigated using a modeling approach. *In vitro* experiments were also performed to validate the computer simulations.

MATERIAL AND METHODS

Computer simulations

The finite element method (FEM) was used to compute the current distribution in the brain tissue surrounding an endovascular coil or a current applicator. COMSOL® Multiphysics 3.5a (a finite element analysis and solver software for engineering applications) was employed to construct different finite element models. The embolization coils were modeled with different geometries: cylinders and helices, with variable diameter and length. In addition, a cylindrical current applicator and an applicator in the middle of a platinum coil were also investigated. Figure 1 illustrates an example of a FEM volume conductor model.

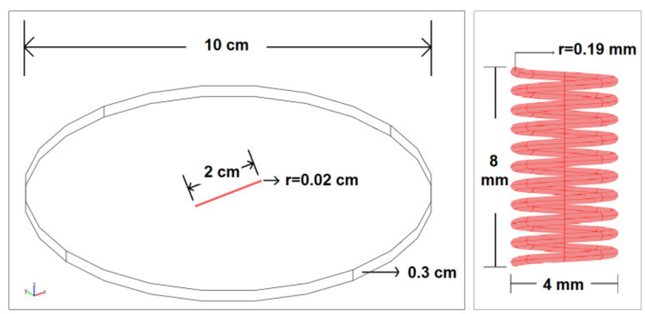

Figure 1: Left: FEM volume conductor model of the Petri dish used for the *in vitro* validation studies; right: a helicoidal coil model.

In order to compute the temperature distribution, the power density produced by the RF current must be found first. The electrical potential *V* must satisfy the following equation:

$$\nabla.(\sigma\nabla V) = 0 \qquad (1)$$

where σ (S m^{-1}) is the electrical conductivity of the medium (tissue phantom and metal). As for the electrical boundary conditions, the potential is constant at one end of the coil or applicator, and zero at the periphery of the cylindrical volume conductor (return electrode), the normal derivative of the potential is zero at the top and bottom of the model; the normal current density and the potential are continuous at the boundary between tissue and metal.

The power density P_d (W m^{-3}) dissipated by the current is calculated by:

$$P_d = \sigma(\nabla V)^2 \quad (2)$$

The temperature is governed by the bio-heat equation which is a parabolic partial differential equation that was first proposed by Pennes to describe heat diffusion in living tissue. This equation is defined as:

$$\rho C \, {\partial T}/{\partial t} + \nabla.(-k\nabla T) = \rho_b C_b \omega_b (T_b - T) + P_d \quad (3)$$

where ρ (kg m^{-3}) is tissue density, C (J kg^{-1} ℃$^{-1}$) is the specific heat of the tissue, T (℃) is the local temperature at any given time, t (s) is time, k (W m^{-1} ℃$^{-1}$) is the thermal conductivity of the tissue, ρ_b (kg m^{-3}) is the density of blood, C_b (J kg^{-1} ℃$^{-1}$) is the specific heat of blood, ω_b (kg m^{-3} s^{-1}) is the blood perfusion rate, T_b(℃) is the arterial blood temperature.

The parameters of the model, including either the platinum coils or the steel applicators, are presented in Table 1.

In vitro experiments

Previous studies [4] showed inhomogeneous temperature distributions along the platinum coils when current was directly injected at one end of the coils, and it was assumed that this was related to possible inductive effects of the coil. To assess this hypothesis, the electrical impedance of different embolization coils was measured at frequencies ranging from 50 to 5000 kHz to characterize the inductive and resistive effects before the construction of the FEM models.

In addition, in order to validate the FEM models, a tissue phantom was constructed. In this phantom, radiofrequency current is applied to one end of a coil immersed in a conductive gel, and the temperature distribution around the coil is monitored with temperature sensitive film and a temperature probe. The maximum temperature at each end of the coil, the voltage and current in the coil were measured as well.

RESULTS

The electrical impedance measurements of different types of coils are given in Table 2. The results show that the coil impedance remains almost constant despite the increase in frequency. It can also be seen that the resistances of all the coils are quite high. This outcome indicates that the drop of temperature that was previously observed along the length of the coil is not due to inductive effects, but is due to the drop of potential produced by the high resistance of the coil. This also suggests that radiofrequency current should not be applied directly to the coil, but to a low resistance applicator placed inside the endovascular coil so as to have a more homogenous temperature distribution.

To investigate this effect, FEM models were created and the parameters were modified to satisfy the characteristics of either the platinum coils or the steel applicators. The models were also modified to satisfy different physical and geometrical properties.

Table 2: Electrical impedance measurements of different types of coils

Name of the Company	Length (cm)	f (kHz)	Z (ohm)	R (ohm)
Boston Scientific	6	50	275.9	199.8
		500	276.9	
		5000	258.2	
HELIPAQ Microcoil	10	50	733.7	545
		500	735.0	
		5000	651.6	
Axium HELIX	8	50	629.4	376
		500	631.2	
		5000	544.9	

The same voltage was applied to one end of an endovascular coil or an applicator (the voltage of the RF source used in the *in vitro* experiments: 17.9 V). The effects of length and width of the coil were also investigated. Figure 2 shows the simulated temperature distribution of selected models obtained after 2 minutes. As can be seen, the platinum coil structure leads to a non uniform distribution with a decreasing temperature along the coil, whereas for the steel applicator, the power is symmetrically focused at the two ends of the coil.

Table 1: Values of electrical and thermal constants of different regions

	σ	ρ	C	k	ρ_b	C_b	ω_b	T_b
Unit	**S m^{-1}**	**kg m^{-3}**	**J kg^{-1} ℃$^{-1}$**	**W m^{-1} ℃$^{-1}$**	**kg m^{-3}**	**J kg^{-1} ℃$^{-1}$**	**kg m^{-3} s^{-1}**	**℃**
Tissue phantom	0.67	4200	1000	0.63	0	0	0	22
Endovascular coil	167	2800	1000	73	0	0	0	22
Steel applicator	4.03e6	7850	475	44.5	0	0	0	22

Figure 2 (g-i) presents the effects of a steel applicator placed in the center of a platinum coil: the temperature distributions are not modified by the presence of the helicoidal coil and are quite similar to those produced by the steel applicator alone.

Table 3 presents the maximum temperature computed after 2 minutes for various sizes of steel applicators, either without (Line 1) or with an enclosing coil (Line 2). It can be found that for short and thin applicators, the maximum temperature is the highest. Also, there is only a slight decrease in the maximum temperature when the steel applicator is placed inside a coil These simulations prove the efficacy of applying RF to a steel applicator instead of a platinum coil itself.

Some *in vitro* experiments were performed to validate the results of the computer models. A steel applicator (with a 0.23 mm section radius) was placed in the middle of an endovascular coil immersed in a tissue phantom while an RF voltage with a 500 kHz frequency was applied. This experiment was performed for applicators with different lengths. The results are shown in Table 4, and selected images of temperature distribution which are demonstrated with the help of temperature sensitive film are shown in Figure 3(a-c).

Table 3: Maximum temperature observed for steel applicators with a various radius *r* and length *L*. Line 1: single applicator, Line 2: applicator placed in the middle of a platinum coil

	r= 0.001 mm					***r= 0.0015 mm***					***r= 0.002 mm***				
	L (mm)					**L(mm)**					**L(mm)**				
	3	**6**	**10**	**15**	**20**	**3**	**6**	**10**	**15**	**20**	**3**	**6**	**10**	**15**	**20**
T (°C)	63.7	50.9	44.2	41.1	39.8	60.3	46.7	39.7	37.6	36.6	53.9	42.1	38.0	35.9	34.6
T (°C)	57.1	48.9	42.7	40.5	39.6	59.2	44.9	39.5	37.4	36.5	55.35	42.6	38.0	35.7	34.9

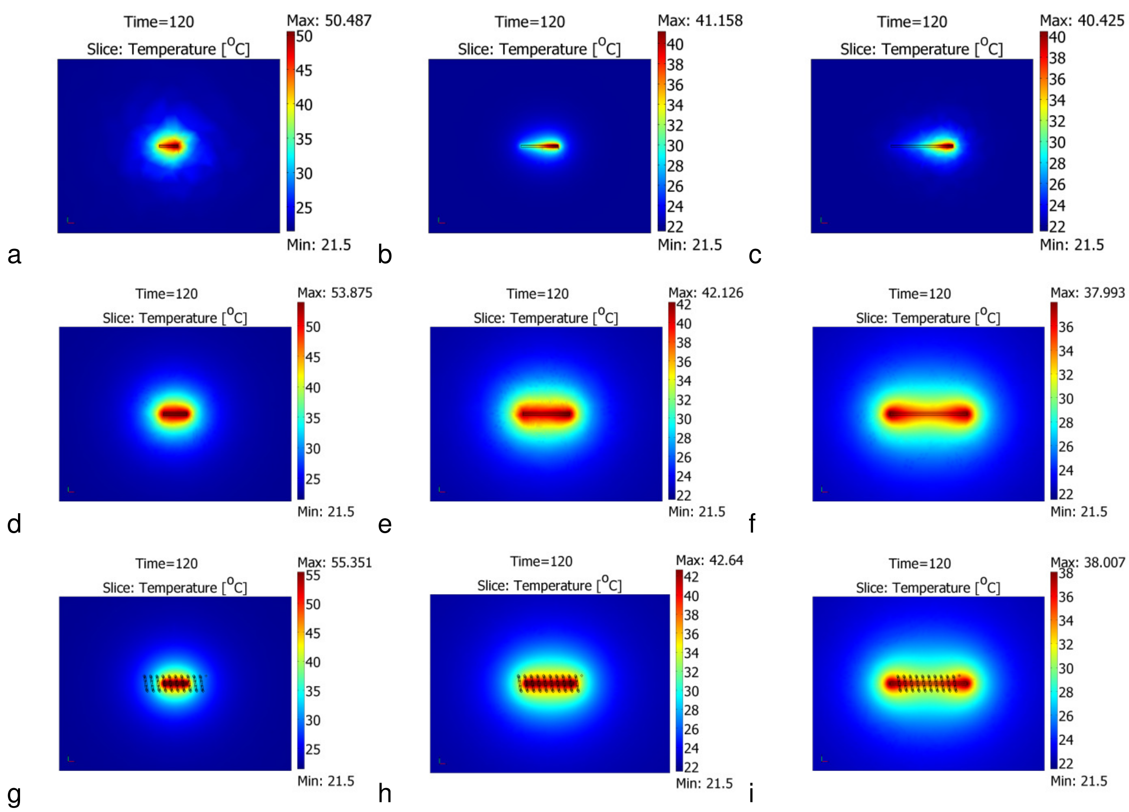

Figure 2: (a-c): models of endovascular coil with length= 3 mm, 6 mm, 10 mm respectively, with radius =200 µm; (d-f): models of steel applicator with length= 3 mm, 6 mm, 10 mm respectively, with radius =200 µm; (g-i): models of steel applicator with length= 3 mm, 6 mm, 10 mm respectively, placed in the middle of a platinum helix, the applicator has a radius of 200 µm whereas the coil has a radius of: 1000 µm.

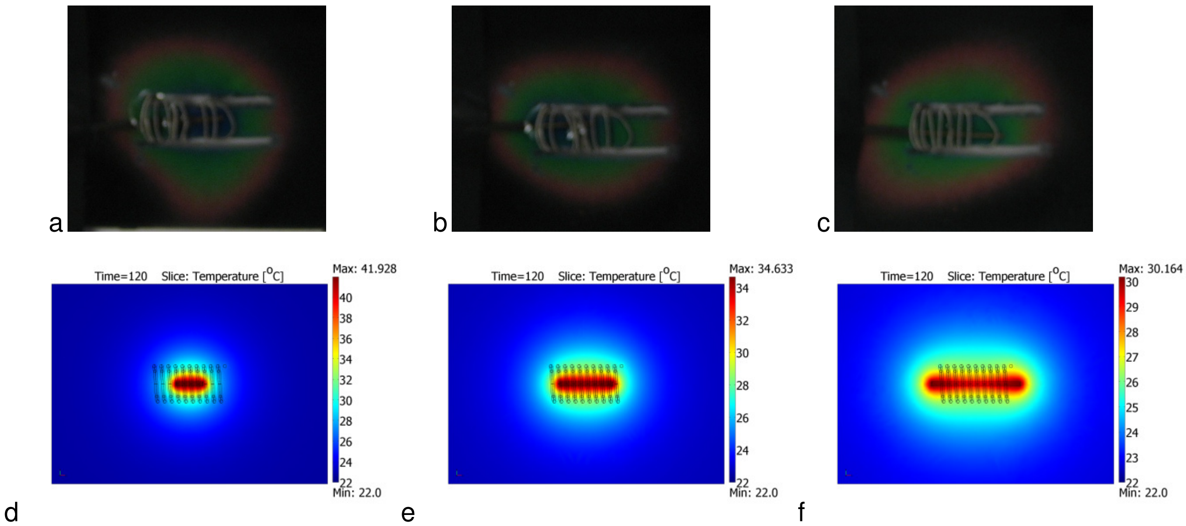

Figure 3: (a-c) Experimental results obtained by applying RF to 3 applicators with length= 3 mm, 6 mm, 10 mm respectively, placed in the middle of an endovascular coil immersed in a tissue phantom; (d-f) simulated temperature distributions computed at 2 minutes for 3 applicators with the same length placed in the middle of an helicoidal coil.

Table 4: Experimental and simulation results for different length of applicator in the middle of a platinum endovascular coil

Length (mm)	t (min)	I (A)	P (W)	Z (ohm)	T max (℃)	T max in simulation
3	2	0.07	0.86	198.2	25	41.928
	5	0.07	0.88	201.2	27	
	10	0.07	0.86	198.2	27.9	
6	2	0.08	0.93	156.6	24.8	34.633
	5	0.08	0.93	160.3	25.2	
	10	0.08	0.92	158.9	26.1	
10	2	0.09	0.98	115.9	24.6	30.164
	5	0.09	0.97	117.4	25.1	
	10	0.09	1.00	118.0	25.9	
15	2	0.09	0.99	114.4	24.5	29.095
	5	0.09	0.99	109.5	25.1	
	10	0.1	1.00	105.8	25.6	
20	2	0.1	1.01	109.2	23.6	28.777
	5	0.09	1.00	115.5	23.9	
	10	0.1	0.98	107.7	24.3	

Models with the same size and characteristics as the in vitro experiment were created in COMSOL; selected results are shown in Figure 3(d-f) and the maximum temperature values are shown in Table 4. As it can be seen the spatial distribution in computer simulations matches with the experimental results but the maximum temperature is lower in the experiments, which can be related to the size of the temperature probe (diameter: 3 mm).

CONCLUSION

Based on this study, we conclude that radiofrequency current should not be injected directly into the platinum coils, but that it should be injected into a steel applicator placed in the center of the endovascular coils. Also, an applicator with a length of 6 to 10 mm is appropriate to generate a uniform temperature distribution. Finally, animal studies should be performed to further investigate this promising approach.

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