A NUMERICAL STUDY OF MAGNETIC NANOPARTICLES HYPERTERMIA WITH ALTERNATING MAGNETIC FIELD UNDER INFLUENCE OF CONVECTION HEAT TRANSFER

MOSTAFA ZAKARIAPOUR1,*, MOHAMMAD HOSSEIN HAMEDI1, NASSER FATOURAE2

1Department of Mechanical Engineering, K.N.T. University of Technology, Pardis street, Tehran, Iran
2Department of Biomedical Engineering, Amir Kabir University of Technology, Hafez street, Tehran, Iran
*Corresponding Author: mzakariapour@yahoo.com

Abstract

In this paper, a numerical study has been conducted to understand the heating effects of the magnetic nanoparticles in the tumor hyperthermia in order to reach a desirable temperature in the tumor. The developed numerical method has been utilized to obtain the temperature distribution and magnetic induction value using the bioheat and Maxwell equations inside a cylindrical geometry including the tumor and healthy tissue while the perfusion and metabolism rates have been considered. Results show that among all the parameters effected on temperature rise, the diameter of the nanoparticles (ranging from 5.5,5.6 nm) has the maximum effect, the strength of the applied alternating current (AC) magnetic field (ranging from 50, 62.75 mT) has the minimum effect, and the volume fraction (ranging from 0.0004,0.0006,0.0008) and the frequency of the applied AC magnetic field (ranging from 300,400,500 kHz) result in increasing the temperature relatively. The temperature rise for a temperature-dependent metabolism is larger than a temperature-independent metabolism. Among the materials investigated in this study, FePt has the most pronounced effect.

Keywords: Tumor, Hyperthermia, Bioheat, Magnetic nanoparticles, Induction.

1. Introduction

Hyperthermia is the procedure of temperature increase of cancerous tissue to 42-46°C for therapeutic reason. It has been shown in many studies; high temperature can cause direct damage to cancerous cells or sensitize them to other cancer treat-
Hyperthermia is a thermal therapy of tumor, through elevating the target tissue temperature in the human body, which therefore has fewer side effects than that of the traditional chemotherapy or radiotherapy. It has been well established that sustained temperature above 42°C will cause necrosis of living cells [2-5]. Hyperthermia is usually used with other forms of cancer therapy, such as radiation therapy and chemotherapy. For instance, it has been demonstrated that cytotoxicity of many chemotherapeutic agents is maximized at temperatures between 40.5°C and 43.0°C [6]. Hyperthermia also enhances the radio-sensitivity in hypoxic, low-pH areas of cancerous tissues [7]. Hyperthermia is a viable cancer treatment for localized malignant tumors and its success has been reported for head and neck cancer, breast cancer, urogenital tract cancer, melanoma and sarcoma [8]. The effectiveness of hyperthermia treatment is related to the temperature achieved during the treatment. An ideal hyperthermia should destroy the tumor cells, without damaging the surrounding normal tissue. Hyperthermia can be performed by laser beam, microwave, ultrasound or magnetic nanoparticle delivery to the tumor regions [9-13].
In the recent years, magnetic fluid hyperthermia (MFH) has been used due to the advantages of cancer hyperthermia therapy. In MFH, a nanofluid containing the magnetic nanoparticles (MNPs) is injected directly into the tumor or is injected to the tumor vasculature. An alternating magnetic field, is then applied to the target region, and then MNPs generate heat, according to Néel relaxation and Brownian rotation. The heat generated increases the tumor temperature. The tumor tissue is then destroyed by raising its temperature to about 42.5°C whereas healthy cells will be safe at lower temperatures [14-16].

The temperature rise due to magnetic nanoparticles therapy in the tissue, strongly depends on the properties of the magnetic material used, the frequency and the strength of the applied magnetic field, the blood perfusion in the tissue, the duration of application of magnetic field and the volume fraction of magnetic nanoparticles [17-20].

Hergt [21-23] studied about various magnetic nanoparticles with respect to optimization of the specific loss power (SLP) for application in tumor hyperthermia.

Rosensweig [24] developed dissipation relationships based on the rotational relaxation of single domain magnetic particles dispersed in a liquid matrix. Kim et al. [25] made theoretical calculations of heat generation as a function of Manganese Iron Oxid (MnFe₂O₄) nanoparticle diameter and compared these specific absorption rate (SAR) values with experimental data. Their work shows only the effect of nanoparticles in heat generation. Bele et al. [26] investigated the effect of high AC magnetic fields on MNPs for magnetic hyperthermia and radiation applications with just two nanoparticles and in a single computational region. Dhar et al. [27] made an analytical study of temperature control in hyperthermia by microwave to attain a desirable temperature at any point during a fixed time, constant perfusion rate and by controlling optimally time dependent heating power. Lin et al.[12] developed a hybrid numerical scheme for solving the transient bioheat equation in spherical coordinates with constant perfusion rate and in one dimensional state. The increase in temperature of biological tissues is estimated for the heating effect of Iron-Platinum (FePt) MNPs. Yong et al. [28] explored the three-dimensional (3-D) electromagnetic (EM) field and transient temperature field induced by two external plate electrodes in the human body containing a tumor during hyperthermia with micro/nano magnetic particles which tissue and perfusion properties are variable with location. Narasimhan et al. [29] studied transient simulations of heat transfer in human eye undergoing laser surgery in single computational region and with constant perfusion rate in 2-D coordinates. Kettering et al. [30] showed the possibility of temperature increase due to MNPs accumulation in tumor. They found that a larger heating effect occurs after exposing to an alternating magnetic field.

Maenosono and Saita [11] investigated the theoretical assessment of chemically disordered fcc-phase (face-centered cubic) FePt MNPs as heating elements for magnetic hyperthermia by combining the heat generation model and the bioheat transfer equation. To show the heating capability of these MNPs, heating capability of these MNPs compared with other MNPs such as magnetite. Consequently, fcc FePt MNPs were found to have a superior heating capability as compared to other MNPs such as magnetite. Bagaria and Johnson [17] considered the tissue model as a two-finite concentric spherical region together with the blood perfusion effect.
Salloum et al. [18, 19] performed an experimental study in a tissue-equivalent agarose gel and evaluated magnetic nanofluid transport and heat distribution in the gel. The SAR distribution showed that the nanoparticles distribution in the gel is not uniform so that the concentration of the nanoparticles close to the injection site is higher than others. Golneshan and Lahonian [31] studied the effect of MNPs dispersion on temperature distribution in a tumor and surrounding healthy tissue, during MFH. In this work, the Pennes bioheat equation (BHE) in a spherical tissue with Neumann curved boundary condition has been solved. The effects of blood perfusion, metabolism heat generation as well as MNPs heat dissipation in an alternating magnetic field as the source term, have been considered. To solve the Pennes BHE, Lattice Boltzmann Method (LBM) has been used and results were compared with analytical ones. Attar et al. [32] studied the dispersion of nanoparticles inside the tumor to investigate the tissues temperature profiles. The problem is solved for polar coordinate in a cylindrical tumor. Also the heating effect of magnetic fluid in a porcine liver tissue was experimentally examined. Numerical transient solutions were found to be in good agreement with experimental data. They found that the injected nanoparticles do not usually distribute uniformly throughout the entire tumor.

Singh et al. [33] investigated hyperthermia with laser induced heating of a tumor in a 2-D axisymmetric tissue embedded with gold-silica nanoshells in the tumor. Effects of power density, laser exposure time, beam radius, blood vessel diameter and volume fractions of nanoshells on temperature spread in the tissue were analysed.

In view of all-above mentioned paragraphs, it is evident that heating the colloidal magnetic fluid (ferrofluid) due to time-varying magnetic induction and real boundary condition in tumor and normal tissue has not been considerably studied. In this study we investigate the effect of physical characteristics of nanoparticles, characteristics of applied magnetic field as well as characteristics of tissue on heat transformation of human body in hyperthermia. It should be mentioned that here evaporation rate from surface of tissue in contact with surrounding space has also been considered.

2. Mathematical Formulation and Boundary Conditions

In order to find the temperature distribution during hyperthermia, it is essential to solve the energy equation within the tumor domain with real boundary conditions. Figure 1 shows the geometry and dimensions of the problem. It assumes that our model simulates skin cancer starting from skin surface and ending to a defined depth. The Pennes bioheat transfer equation [34] for a tumor and tissue can be written as follows respectively

\[
\frac{\partial \rho_1 c_1 T}{\partial t} = \nabla \cdot \left( k_c \nabla T \right) + \rho_b c_b w_b (T_a - T) + Q_{\text{metabolism}} + P
\]

\[
\frac{\partial \rho_2 c_2 T}{\partial t} = \nabla \cdot \left( k_c \nabla T \right) + \rho_b c_b w_b (T_a - T) + Q_{\text{metabolism}}
\]

where, \( T \) is the temperature, \( t \) is the time, \( w_b, \rho_b, c_b \) and \( T_a \) are the perfusion, the density, the specific heat and the temperature of the blood, \( Q_{\text{metabolism}} \) and \( P \) are the
metabolic heat generation of the tissue and the distributed volumetric heat source due to spatial heating of MNPs.

Considering the evaporation term in the skin, heat equation can be written as follows [35]

\[ \rho_c c_1 \frac{\partial T}{\partial t} = \nabla \cdot (k \nabla T) + \rho_b c_b w_b (T_a - T) + Q_{\text{metabolism}} + P \cdot Q_E \]  

(3)

in which \( Q_E \) is referred to the amount of evaporation which means is the amount of water consumption in the skin per second' is defined as follows. [35]

\[ Q_E = -a \frac{dW}{dt} \]  

(4)

where \( a \), is the latent heat of water, which is equal to 2260 kJ/kg and \( W \) is the density of water in tissue and is, only temperature dependent. Eq. (3) will be obtained as follows

\[ (\rho c') \frac{\partial T}{\partial t} = \nabla \cdot (k \nabla T) + \rho_b c_b w_b (T_a - T) + Q_{\text{metabolism}} + P \]  

(5)

\[ c' = c - \frac{a}{\rho} \frac{\partial W}{\partial T} = c - \frac{a}{\rho} W' \]  

(6)

\[ W(T) = 778 \times (1 - \exp(-\frac{T - 106}{3.42})) \]  

(7)

In 2-D cylindrical coordinates we have

\[ (\rho c') \frac{\partial T}{\partial t} = k \frac{\partial}{\partial r} \left( r \frac{\partial T}{\partial r} \right) + k \frac{\partial^2 T}{\partial z^2} + \rho_b c_b w_b (T_a - T) + Q_{\text{metabolism}} + P \]  

(8)

From Fig. 1, the boundary conditions are as follows. L, the height of the healthy tissue, R, the radius of the healthy tissue, h, the height of the tumor tissue and a, the radius of the tumor tissue

at \( z = 0 \rightarrow h(T - T_w) = -k \frac{dT}{dz} \)  

(9)

at \( z = L \rightarrow \frac{dT}{dz} = 0 \)  

(10)

at \( r = R \rightarrow T = 37^\circ C \)  

(11)

The initial boundary condition for temperature is also as follows

at \( t = 0 \rightarrow T = 37^\circ C \)  

(12)

in which, the specific heat (\( c_1 \)) and density (\( \rho_1 \)) (for the tumor in Eq. (1)) is consisted of the tissue (with index 2) and the nanoparticles (with index \( M \)) and with volume fraction of \( \varphi \) and given by [29]

\[ \rho_1 = \varphi \rho_M + \rho_2 (1 - \varphi) \]

\[ c_1 = \varphi c_M + c_2 (1 - \varphi) \]  

(13)
The power dissipation density for a mono-dispersion with constant susceptibility is expressed as follows [11]

\[ P = \pi \mu_0 X_0 H_0^2 f \frac{2\pi f t}{1 + (2\pi f t)^2} \]  

(14)

in which \( \mu_0 = 4\pi \times 10^{-7} \text{Tm/A} \) is the permeability of free space, \( X_0 = (\mu_0 \phi^2 M_d^2 V_M)/kT \) the susceptibility (here is assumed magnetic field independent and \( M_d \) is the domain magnetization of a suspended particle), \( H_0 \) is the strength of applied AC magnetic field and \( f \) is the cyclic frequency of applied AC magnetic field. Because the Brownian and Néel processes take place, the effective relaxation time, \( \tau \), is defined as follows [11]

\[ \tau = \frac{\tau_N \tau_B}{\tau_N + \tau_B} \]  

(15)

where \( \tau_N \) is the Néel relaxation time and \( \tau_B \) is the Brownian relaxation time. The first mechanism existing in this phenomenon is the Brownian mechanism of relaxation, the magnetic moment is locked to the crystal axis and when the magnetic moment aligns with the field, the particle rotates as well. A second mechanism exists (Néel relaxation) in which the magnetic moment rotates within the crystal. To achieve high heating rates, the Néel relaxation must not be dominated. The Brownian time constant is given by the following relationship [11]

\[ \tau_B = \frac{3\eta V_M}{kT} \]  

(16)

where \( \eta \) is the viscosity of the carrier liquid, \( V_M = \pi (D+2)^3/6 \) is the hydrodynamic volume of the particle, \( D \) is the nanoparticle maximum diameter, \( k_B \) is the Boltzmann constant, and \( T \) is the absolute temperature. The Néel relaxation time, denoted \( \tau_N \), is given by the following expression due to Brown [11]

\[ \tau_N = \tau_0 \exp\left(\frac{K V_M}{kT}\right) \]  

(17)

where \( K \) is the anisotropy constant, \( V_M = \pi (D)^3/6 \) is the volume of the particle, \( T \) is the absolute temperature, and \( \tau_0 \) is the time constant, (\( \tau_0 = 10^{-7} \text{s} \)).

By uncoupling Maxwell equations, here we consider the magnetic induction formula as follows [36]

\[ \mu_0 \sigma \frac{\partial B}{\partial t} = \nabla^2 B \]  

(18)

where \( B \) is the magnetic induction field, \( \mu_0 \) is the magnetic permeability and \( \sigma \) is the electrical conductivity of the medium. Magnetic induction field is related to the strength of applied AC magnetic field as follows[36]

\[ B = \mu_0 (H + M_d ) \]  

(19)

and we have

at \( r=a \rightarrow B = 37 mT \)  

(20)
3. Numerical Method

To solve Eqs. (3) and (18), a finite difference method (FDM) is used in which the second order forward difference is used for coordinate-dependent terms and an implicit approach is also used to discretize the time-dependent terms. To do this task, Eqs. (3) and (18) are solved simultaneously, i.e., firstly Eq. (18) is solved and the value of magnetic induction in all over the computational domain and all times is obtained. Then, the Eq. (19) is solved using the obtained magnetic inductions from Eq. (18) and considering the value of the thermal power calculated by Eq. (14). Finally, temperature at different times and locations is obtained by solving Eq. (3).

4. Grid Independency

As it was explained in section 2, in this study the described problem is two dimensional and governing equations are discretized along the r and z directions shown in Fig. 1. In order to make sure that the results are independent from the grid resolution, Eq. (3) is solved on four computational grids i.e. 40×40, 50×50, 60×60 and 70×70. The results for temperature along the z direction and relevant to the mentioned grids are shown in Fig. 2. As the figure indicates, because of nearness of the results of 60×60 and 70×70 grids, choosing the 70×70 grid as the independent grid is completely reasonable. The time step is Δt=0.5s.

5. Verification of the code

In this section, for verifying the results, this paper is compared with a numerical research which performed in hyperthermia with MFH therapy [32]. To evaluate
this, the Pennes bioheat equation in cylindrical coordinate is solved. In cylindrical coordinate, two eccentric cylinders are assumed as the computational domain for cancerous tissue with and without nanoparticles. These two cylinders belong to a healthy tissue. The inner cylinder is assumed as the tumor containing nanoparticles while the remaining area of the outer cylinder, is assumed to be the tumor without nanoparticles (Fig. 3).

![Grid independence in z direction.](image)

**Fig. 2. Grid independence in z direction.**

To verify the numerical method performance used in this paper, with Attar et al. [32], temperature profile for the homogenous distribution of nanoparticles in tumor with $r_c=0$ and $d_n=2.76$ cm, due to Fig. 3, in $t = 750$ s with finite difference numerical method is shown, as see in Fig. 4. In Fig. 4 the numerical result of finite difference study in this paper, is compared with the result of Attar et al. [32] that was performed by finite element method.

![Model of cancerous and healthy tissues in cylindrical coordinate.](image)

**Fig. 3. Model of cancerous and healthy tissues in cylindrical coordinate [32].**

As it is mentioned in this paper, one of the best distribution of particles is obtained by consecutive injection [32]. The most important advantage of this
A numerical study of magnetic nanoparticles hyperthermia with alternating injection method is the moving of the location of maximum temperature point to a place far from the center that it produces temperature increase in a larger area of the tumor. Another advantage is obtaining more homogeneous temperature profile. By consecutive injection of specific amount of nanoparticles at θ=0, 90, 180, 270, temperature distribution is better than homogeneous distribution.

![Graph showing hyperthermic temperature difference for the homogenous distribution of nanoparticles with P = 0.0025 (W/cm³).](image)

The temperature distributions of the tumor for consecutive injection of the particles at angles of 0, 90, 180 and 270 are shown in Fig. 5. In Fig. 5(a) the Attar et al.[32] results which have been obtained by finite element method are shown. The results of the two methods are compared with the Attar et al. results. Figure 5(b) portrays the results obtained by COMSOL Multiphysics which is well-known element-based software. The results of the finite difference method used in the developed code are shown in Fig. 5(c). As it was said earlier, this study is carried out using finite difference method and this method has been used for the validation purpose as well. As the mentioned figures indicate, there is a good agreement among these results. The values of the different parameters used for the validation are also listed in Table 1.

### Table 1. Parameters for validation with Attar et al. [32].

<table>
<thead>
<tr>
<th>MNP</th>
<th>R (cm)</th>
<th>D (nm)</th>
<th>f (kHz)</th>
<th>w_b (1/s)</th>
<th>H (kA/m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Magnetite</td>
<td>5</td>
<td>19</td>
<td>450</td>
<td>0.001</td>
<td>10</td>
</tr>
</tbody>
</table>

### 6. Results and Discussion

A numerical simulation is proposed to solve the bioheat transfer and magnetic induction equations in a two zone tissue in cylindrical geometry with blood perfusion and metabolism. Bioheat equation is used to predict the temperature rise in term of characteristics of the magnetic nanoparticles, applied magnetic field and the tissue. Physical properties of magnetic nanoparticles and physical characteristics used for tissue and blood are given in Tables 2 and 3, respectively. Figure 6 shows the induction effect of applied AC magnetic field on the temperature field versus time (Fig. 6(a)) and axial position (Fig. 6(b)).
Fig. 5. Hyperthermic temperature difference for four consecutive injections with \( P = 0.005 \text{ (W cm}^3) \), (a) Attar et al. simulation [32], (b) Simulated result for proposed work by COMSOL Multiphysics, (c) Simulated result by FDM.

As the figure shows, by increasing the strength of applied AC magnetic field, the temperature increases. The reason for temperature increase with increasing the induction amount is that the field intensity increases with increasing the magnetic induction around the tumor and as Eq. (14) indicates the field intensity augmentation has the direct effect on the heat generated by the particles.

Figure 7 shows the frequency effect of applied AC magnetic field on the temperature field versus time (Fig. 7(a)) and axial position (Fig. 7(b)). As the figure shows, by increasing the frequency of applied AC magnetic field, the temperature increases. Frequency increase is not always reasonable and this variable is allowed to change as much as it does not have negative effect on the body.
Table 2. Physical properties of magnetic solids [11].

<table>
<thead>
<tr>
<th>Magnetic solid</th>
<th>Chemical formula</th>
<th>$M_d$ (kAm$^{-1}$)</th>
<th>$K$ (kJm$^{-3}$)</th>
<th>$c$ (J kg$^{-1}$K$^{-1}$)</th>
<th>$\rho$ (kgm$^{-3}$)</th>
<th>$D_{max}$ (nm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Maghemite</td>
<td>Fe$_2$O$_3$</td>
<td>414</td>
<td>4.7</td>
<td>746</td>
<td>4600</td>
<td>23.5</td>
</tr>
<tr>
<td>Magnetite</td>
<td>FeO Fe$_2$O$_3$</td>
<td>446</td>
<td>9</td>
<td>670</td>
<td>5180</td>
<td>19</td>
</tr>
<tr>
<td>Cobalt ferrite</td>
<td>FeCo</td>
<td>1790</td>
<td>1.5</td>
<td>172</td>
<td>8140</td>
<td>340</td>
</tr>
<tr>
<td>Platinum ferrite</td>
<td>Li$_0$ FePt</td>
<td>1140</td>
<td>206</td>
<td>327</td>
<td>15200</td>
<td>9</td>
</tr>
</tbody>
</table>

Table 3. Physical characteristics used for tissue and blood [11].

<table>
<thead>
<tr>
<th>$\rho_t$ (kg/m$^3$)</th>
<th>$c_{p,t}$ (J kg$^{-1}$K$^{-1}$)</th>
<th>$\rho_b$ (kglm$^{-3}$)</th>
<th>$c_{p,b}$ (J/kg.K)</th>
<th>$k_t$ (W/m.K)</th>
<th>$T_a$ (°C)</th>
<th>$w_b$ (1/s)</th>
<th>$\sigma$ (S/m)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1060</td>
<td>3600</td>
<td>1000</td>
<td>4180</td>
<td>0.5</td>
<td>37</td>
<td>0.0064</td>
<td>0.37</td>
</tr>
</tbody>
</table>

Fig. 6. Effect of the strength of applied AC magnetic field on the temperature field, (a) Versus time in $z=8$mm, (b) Versus position in $t=300s$.

Figure 8 shows the effect of nanoparticles maximum diameter on the temperature field versus time (Fig. 8(a)) and axial position (Fig. 8(b)). As the figure shows, by increasing the nanoparticles maximum diameter, the temperature increases. Nanoparticle diameter is the most important parameter in hyperthermia. Both Magnetic permeability and thermal power increases with increasing the particle diameter.

Figure 9 shows the effect of nanoparticles volume fraction on the temperature field versus time (Fig. 9(a)) and axial position (Fig. 9(b)). As the figure shows, by increasing the nanoparticles volume fraction, the temperature increases. Figure 10 shows the effect of tissue metabolism and perfusion rate on the temperature field versus time and axial position. As the figure shows, temperature dependent modeling of metabolism and perfusion rates lead to lower temperature values and the effect of temperature dependent perfusion rate is higher than temperature dependent metabolism rate.

Journal of Engineering Science and Technology February 2017, Vol. 12(2)
Fig. 7. Effect of the frequency of applied AC magnetic field on the temperature field, (a) Versus time in z=8mm, (b) Versus position in t=300s.

The time constant for various cases are illustrated in Table 4. Also Fig. 11 shows the temperature field along radial direction in a defined depth and surface of tumor. In Fig. 11(b) it is shown that the temperature in surface of the tumor in the first times of hyperthermia treatment, is lower than 37°C in some areas. That is why there is convection heat transfer on tumor surface. The peak temperature in all cases appears at ≈ 8 mm from the top surface under this conditions.

Fig. 8. Effect of the nanoparticles maximum diameter on the temperature field, (a) Versus time in z=8mm, (b) Versus position in t=300s.

Fig. 9. Effect of the nanoparticles volume fraction on the temperature field, (a) Versus time in z=8mm, (b) Versus position in t=300 s.
Figure 12 indicates that with increasing convection heat transfer coefficient on the top of skin, temperature field in axial direction decreases and therefore the effect of MFH decreases. As it is expected, by increasing convection heat transfer coefficient, temperature difference between surface and the depth of the tumor, increases. So, MFH is better done in a medium with natural convection. Figure 13 shows the effect of convection heat transfer coefficient on temperature field versus time.

In Table 5, effects of convection and evaporation on tumor temperature (at maximum temperature position) are considered. As it is expected using Eq. (11) for density of water in tissue, the effect of evaporation of tissue on the temperature field is negligible. That is why we can neglect it. But convection has an important role in the temperature field. As it can be seen in Table 5, by changing convection heat transfer coefficient, the peak temperature position changes.
Fig. 12. Convection heat transfer coefficient effect on the temperature field in axial position in $t=300$ s.

Fig. 13. Convection heat transfer coefficient effect on the temperature field in versus time.

Table 4. Time constant for various condition.

<table>
<thead>
<tr>
<th>Condition</th>
<th>Time Constant</th>
<th>Material</th>
</tr>
</thead>
<tbody>
<tr>
<td>$f = 300$ kHz, $B_1 = 37$ mT, $\varphi = 0.0006$</td>
<td>$D = 6$ nm, $B_1 = 37$ mT, $f = 300$ kHz</td>
<td>$f = 300$ kHz</td>
</tr>
<tr>
<td>$D = 5$ nm</td>
<td>$\tau = 80$ s, $\varphi = 0.0004$</td>
<td>$\tau = 83.2$ s</td>
</tr>
<tr>
<td>$D = 5.5$ nm</td>
<td>$\tau = 81$ s, $\varphi = 0.0006$</td>
<td>$\tau = 83.5$ s, $\varphi = 0.0008$</td>
</tr>
<tr>
<td>$D = 6$ nm</td>
<td>$\tau = 83.5$ s, $\varphi = 0.0008$</td>
<td>$\tau = 83.7$ s, $\varphi = 0.0008$</td>
</tr>
<tr>
<td>$f = 400$ kHz, $B_1 = 37$ mT, $\varphi = 0.0006$</td>
<td>$D = 6$ nm, $B_1 = 37$ mT, $f = 300$ kHz</td>
<td>$f = 300$ kHz</td>
</tr>
<tr>
<td>$f = 500$ kHz, $B_1 = 37$ mT, $\varphi = 0.0006$</td>
<td>$D = 6$ nm, $B_1 = 37$ mT, $f = 300$ kHz</td>
<td>$f = 300$ kHz</td>
</tr>
<tr>
<td>$D = 6$ nm, $f = 300$ kHz, $\varphi = 0.0006$</td>
<td>$B_1 = 50$ mT</td>
<td>$\tau = 83.7$ s, $\varphi = 0.0006$</td>
</tr>
<tr>
<td>$D = 6$ nm, $f = 300$ kHz, $\varphi = 0.0006$</td>
<td>$B_1 = 75$ mT, $\varphi = 0.0006$</td>
<td></td>
</tr>
</tbody>
</table>

Journal of Engineering Science and Technology | February 2017, Vol. 12(2)
Table 5. Effect of convection and evaporation (compared with no convection and no evaporation condition).

<table>
<thead>
<tr>
<th>Condition</th>
<th>(Z_{T_{\text{max}}} (\text{mm}))</th>
<th>(\Delta T (\degree \text{C}))</th>
</tr>
</thead>
<tbody>
<tr>
<td>(h=0 \text{ w/m}^2\cdot \text{k} ) &amp; (0) &amp; (0)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>(h=5 \text{ w/m}^2\cdot \text{k} ) &amp; (6) &amp; (0.462)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>(h=10 \text{ w/m}^2\cdot \text{k} ) &amp; (7.4) &amp; (0.802)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>(h=20 \text{ w/m}^2\cdot \text{k} ) &amp; (8) &amp; (1.452)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>(h=40 \text{ w/m}^2\cdot \text{k} ) &amp; (9.7) &amp; (2.457)</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

\(^\text{1Temperature difference between } T_{\text{max}} \text{ in conditions with no convection and evaporation condition.}\)

7. Conclusions

A numerical study has been conducted to understand the heating effects of the magnetic nanoparticles in the tumor hyperthermia in order to reach a desirable temperature in a specific location of the tumor inside the healthy tissue. The developed numerical method has been utilized to obtain the temperature distribution and magnetic induction value using the bioheat and Maxwell equations inside a cylindrical geometry including the tumor and healthy tissue while the perfusion and metabolism rates have also been considered. Results show that the strength of the applied AC magnetic field has the minimum effect, the volume fraction and the frequency of the applied AC magnetic field result in increasing the temperature relatively and the diameter of nanoparticles has the maximum effect on the temperature increase. Among the materials investigated in previous studies, FePt has the most pronounced effect. Also, the temperature increase for the location-dependent perfusion rate, is less than that of for the location-independent one.

Similarly, the temperature rise for a temperature-dependent metabolism rate is larger than that found for a temperature-independent metabolism rate. At small times the maximum occurs at the tumor boundary, while in steady state it occurs at the tumor center. As it can be observed, convection heat transfer on surface of the tumor has cooling effect on the surface. That is why, maximum temperature in the tumor occurs in a defined depth. By examining four types of nanoparticles in Table 2, results show, FePt is the best nanoparticle in hyperthermia for rising temperature. Also, as it is shown in results, it is better to perform MFH with natural convection condition, so the temperature of the tumor in MFH therapy decreases less.

Table 4 shows the time constant at different conditions. As the table shows, among traditional particles used in hyperthermia, the time needed for FePt to reach the steady state is less than other particles; therefore this particle has more thermal capability. Additionally, both hyperthermia performance and the rate of reaching the steady state increase with increasing the frequency. However the frequency increase should be in an accept range.
References


