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TLM modelling of Laser thermal treatment of benign prostate hyperplasia

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Abstract — A two–dimensional Transmission Line Matrix (TLM) model for the study of living tissue when exposed to laser applications is represented and used to investigate quantitative damage size due to thermal coagulation in thermal treatment of benign prostate hyperplasia (BPH). Results show a quasi linear dependency of perfusion with temperature until the beginning of coagulation then perfusion falls rapidly to vanish completely when all tissue under investigation has coagulated. Increasing perfusion rate (ω) leads to larger values of t_{100}, time required to reach a primary zone temperature of about 100°C, and reflects the increased sinking created by perfusion. The TLM numerical model predicts the coagulation damage contours and thus has a clinical interest in therapy and as an aid for clinicians since damage cannot be easily measured within patients. TLM method can be used in the study of a wide variety of heat transfer clinical treatments.

Keywords — **TLM, Modelling, Thermal coagulation, Prostate Hyperplasia, Perfusion, Clinical therapy.**

I. INTRODUCTION

Benign Prostate Hyperplasia (BPH), also known as Benign Enlargement of Prostate (BEP) is a pervasive condition of enlargement of the male prostate gland which leads to several urinary difficulties. As a treatment of (BPH), interstitial laser coagulation (ILC) appears to be among the choices of an invasive therapy. It is similar in efficacy and safety to standard transurethral resection of the prostate (TURP), according to findings of last few years randomized trials [1],[2]. Among these clinical trials, our simulations are based on those of the Indigo Medical Inc [3]. A standard cystoscope is introduced into the urethra delivering a fibrotic probe into the prostate. A low-power laser light (15-20W), with a wavelength of 800-850 nm, is delivered through the probe to heat and destroy a controlled amount of prostate, which is evacuated by the body over time. In this study, we are interested in thermal burning of tissues due to laser irradiation. Absorption of the laser light goes into raising the tissue temperature according to its specific heat, approximately that of water. Heat is then carried away from the laser deposition region by thermal conduction, which is modeled by the Penne's bioheat equation. This one is a differential equation and needs to be solved numerically rather than simplified analytical approaches which may be inadequate. Different techniques have been adopted to solve this equation including the transmission line matrix (TLM) technique [4-6], which shows higher stability criteria than classical ones [7-9].In this paper, a two-dimensional TLM model is used to analyse temperature distribution under blood perfusion and to predict lesion size on prostate tissue using respectively Penne's bioheat equation and Arrhenius model.

II. GOVERNING EQUATIONS OF THE PROBLEM

Thermal therapeutic lasers are used to produce three basic thermal effects depending on time exposure and beam energy: Hyperthermia, Coagulation and Volatilization. Other energy sources beside lasers can also be used to produce these tissue effects, but lasers seem to possess certain basic advantages [10-14]. In comparison with monopolar or bipolar diathermy and heater probes, lasers can deliver more power, and presents good accuracy with better control of damage size. In comparison with microwave and ultrasound therapy, lasers are again more precise and can be used with more compact and accurate delivery devices.

To avoid simulation of light distribution in target tissue, we consider the conversion of laser light to heat within a source of heat called "primary" source [3],[15] modelling the first zone of tissues centred around and in contact with laser fibre. From clinical trials of the Indigo Medical Inc. [3], laser power was controlled to maintain source temperature as close to 100 °C as possible. In this study, we consider primary zone surrounding the laser emissive section probe as isothermal nodes with an initial temperature of 37°C which begins to increase linearly until 100°C, thereafter it is maintained constant till the end of simulation as shown on Fig.1. One has to notice that boundaries are sufficiently far from laser deposition and heated region, that they have very little influence on inner region of interest.

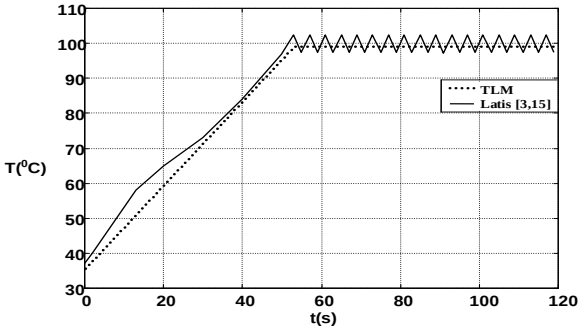

Fig.1: Temperature evolution of primary zone.

Heat transfer through tissues tends to enlarge the volume of "primary" zone, creating a "secondary" heated zone bigger than the first one. It is this "secondary" zone that is considered in this study. Heat transfer is essentially produced by conduction since the influence of blood circulation even for large vessels (transport by convection) is negligible for most existing bioheat models [16-19]. It has been proved that standard Penne's bioheat model is valid for a region far from large blood vessels, which is the case in our study since prostate is not a highly perfused organ with many large blood vessels as liver and many others organs. Penne's bioheat equation

[17- 21] represents a continuum model in which blood flow is treated as a temperature-dependent heat source term that introduces some biological effects into the standard equation of thermal diffusion:

$$\rho c \frac{\partial T(x,y,t)}{\partial t} = k \nabla^2 T(x,y,t) + Q + \omega \rho_b c_b \left(T_a - T(x,y,t)\right) \quad (1)$$

where T is temperature, k, ρ, and c are thermal conductivity, density and specific heat capacity of prostate tissue, respectively. T_a is arterial blood temperature and index b indicates blood properties.

The metabolic heat generation rate Q is assumed homogeneous through the tissue and takes the form of energy per unit volume. The perfusion rate in tissue ω depends on several factors such as pH, temperature, hormonal activity, blood pressure and activity of other organs. Penne's model [19] assumes that blood perfusion is homogeneous and isotropic and that thermal equilibrium occurs at the micro blood vessels (capillaries). In this case, blood enters the capillaries with arterial blood temperature T_a, where heat exchange occurs to rise this value to that of the surrounding tissue temperature.

In a general sense, thermal damage is used to express any kind of irreversible changes, therapeutic or not, in biologic tissues due to high temperature exposure. As thermal ablation treatments are used more frequently in clinical treatments, to remove pathologic tissue or destroy its function, models incorporating tissue damage will be required for predicting treatment efficiency. A quantitative description of tissue thermal damage has been suggested by Henriques & Moritz [25] and is given by the Arrhenius integral formulation where an exponential function of temperature is integrated over time:

$$\Omega(t) = \ln\left(\frac{C_{on}(x,y,0)}{C_{on}(x,y,t)}\right) = \int_0^t A e^{\left[\frac{-E}{R\,T(x,y,t)}\right]} dt \quad (2)$$

Ω is a dimensionless damage parameter, C_{on} is remaining concentration of native state of tissue at time t, R is universal gas constant ($J.mol^{-1}.K^{-1}$) and T is absolute temperature history of tissues (K). A and E are Arrhenius parameters, called respectively, pre-exponential factor or frequency factor and activation energy required to transform tissue from normal to damaged state, A is expressed in s^{-1} and E is expressed in $J.mol^{-1}$. These two parameters have been measured for various tissues [22].

The increase in tissue temperature is also expressed by an internal energy excess [3], which depends on tissue properties:

$$E_{int} = \rho \cdot \left(c \cdot \Delta T + fd \cdot E_d\right) \quad (3)$$

Where E_d is the energy per unit mass required for coagulation and f_d is the fraction of deactivated or damaged tissues given by:

$$fd = \frac{C_{on}(0) - C_{on}(t)}{C_{on}(0)} = 1 - e^{-\Omega} \quad (4)$$

Since coagulation is a result of tissue vasculature damage, the perfusion coefficient ω is dependent on the tissue damage integral (Equ.2). However, experimental measurements treatments [12],[20], [23] have demonstrated that blood perfusion in response to elevated temperature is a complex function of temperature and time. In this study, perfusion is modelled as an increasing function with temperature due to vessels dilation, before decreasing exponentially with tissue damage until its non existence. The following approximation of function ω can be assumed [3],[15]:

$$\omega = 0.32 \times (1 + 0.059T) \quad before\ damage \quad (5)$$

$$\omega = \omega_0 e^{-\Omega} \quad during\ damage \quad (6)$$

III. TLM NUMERICAL MODEL

Because of complex device geometries and nonlinear tissue properties, analytical techniques are seldom used for solving partial differential equations (PDEs). Instead, numerical techniques are frequently employed with the advent of computers to solve most relevant PDEs. Transmission Line Matrix (TLM) is based on Huygens principle and could be used for modeling any phenomena which obeys this principle. It is a time and space discrete method that solves field problems using their electrical circuit equivalent. It assembles in space a lattice of discrete points connected by transmission lines and defines the transmission matrix between lattice points, so that successive calculations can be performed. Transmission-lines are considered as distributed models of capacitors C, inductors L and resistors R. The user therefore has a good grasp of properties and model behavior, the nature and significance of errors and the manner in which material properties may be introduced. The physical variable is modeled as a sequence of voltage pulses (V) travelling through this network of transmission lines to become incident simultaneously on all parts of all nodes. These incident pulses are scattered instantaneously into reflected pulses which, during the time step Δ*t*, travel along transmission lines to become incidents upon neighboring nodes.

A network of transmission lines obeys Maxwell's curl equation for electromagnetic propagation:

$$LC \frac{\partial^2 V(x,y,t)}{\partial t^2} + RC \frac{\partial V(x,y,t)}{\partial t} + RG\, V(x,y,t) = \nabla^2 V(x,y,t) \quad (7)$$

When taking small time steps, the inductance (L) term can be neglected and an analogy between bioheat equation (1) and Maxwell's equation (7) is possible. Hence, biological tissue is modelled as a lumped RC network of interconnected transmission lines. Then analogies are made where temperature T becomes analogous to potentials V with:

$$RC \cdot \frac{\partial V}{\partial t} \equiv \frac{\rho c}{k} \frac{\partial T}{\partial t} \quad , \quad RGV \equiv \frac{\omega \rho_b c_b}{k} T \quad (8)$$

The resistor R and capacitor C are electric models of thermal tissue properties. From dimensional analysis, these electric parameters are:

$$R = \frac{1}{2k\Delta x}, C = \rho c\ \Delta x^3 \quad (9)$$

Blood perfusion ω is considered as a heat sink opposed to high temperatures until coagulation. Some researchers [24] consider its effect as an increase in thermal conductivity of perfused tissue and proposed a so called "effective conductivity" k_{eff}. In this work, perfusion is modelled as a controlled current source I instead of a conductance G as shown in equation (7) to allow more stables results [7]. This current source depends at every moment on the node temperature. $I_p \equiv \frac{\omega \rho_b c_b}{k}(T - T_a)$ (10)

The metabolic heat generation rate Q is modelled as a constant current source:

$$I_m \equiv Q \cdot \Delta x^3 \quad (11)$$

The typical form of a TLM node in 2D for bioheat transfer is illustrated in Fig.2. Resistors R remain clustered around the nodes, but capacitance C is modelled in terms of transmission lines of impedance Z which connect each node to its neighbours and carry voltage pulses V between nodes in a finite time Δt.

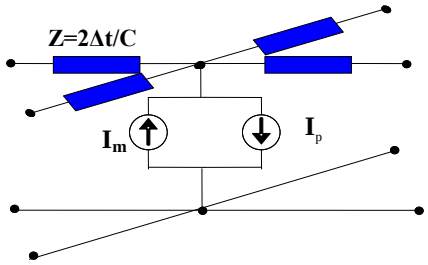

Fig.2: Typical form of 2D TLM node for bioheat transfer.

Our TLM two-dimensional model is based on coupling two processes (Fig.3):

1) Thermal diffusion that takes into account blood perfusion and its variation with temperature.
2) Coagulation which results from increasing temperatures and its effect on blood perfusion.

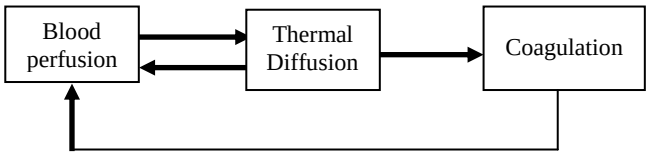

Fig.3: Coupled processes in bioheat TLM modelling.

IV. RESULTS AND DISCUSSION

A 2D numerical analysis is used to predict temperature distribution and lesion size on perfused tissue as prostate when exposed to laser irradiation. Table 1 gives values of thermo-physical parameters [3], [9],[15] for thermal conductivity (k) and heat capacity (c) of soft tissues. They are taken to be standard values for water, which is typical for prostate and other soft tissues. Specific blood perfusion rate ω_0 is not available for human prostate, but this is a typical value for dog prostate and other soft tissues [3],[15]. Standard value of damage energy E_d is taken from differential scanning calorimetric measurements for several tissues.

Table1: Thermo-physical properties used in simulations.

parameter	Standard value	Units
k(tissue)	0.609	W/(m.K)
c(tissue)	4200	J/(kg.K)
c_b	4190	J/(kg.K)
ρ(tissue)	1050	(kg/m^3)
ρ_b	1050	(kg/m^3)
ω_0	6.66×10^{-6}	m^3/(kg.s)
E_d	5×10^5	J/kg
Q	0.748	W/mm^3
A	7.6×10^{66}	s^{-1}
E	4.30×10^5	J/mol

2-D temperature distribution and damage integral after 60s are shown in Fig. 4.a and 4.b, respectively. We can approximate threshold temperature T_{th} for tissue damage, by assuming that damage occurs when Ω=1, which corresponds to a coagulation of about 63% of tissue's volume [25]. This threshold temperature varies with thermal properties of tissue. From TLM modelling a value of $T_{th} \approx 60$ °C is found for prostate.

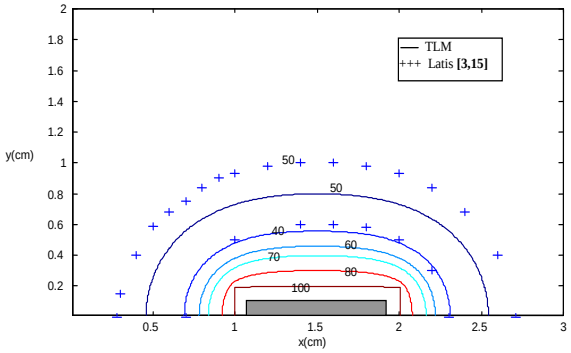

Fig. 4a: Temperature isotherms at 60s.

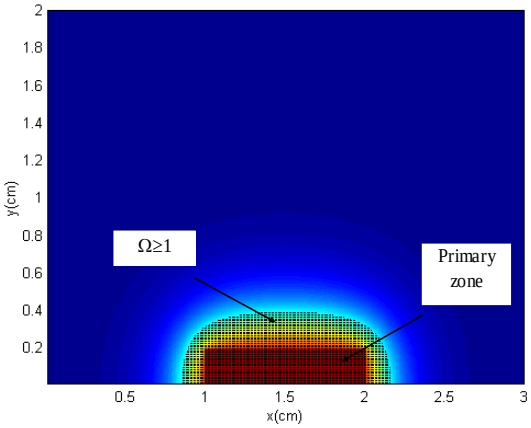

Fig.4b: Coagulated zone at 60s.

Isotherms marked (+) in Fig.4.a are those provided in other works [3],[15] that have included transport of light in tissue. Fig.4.b shows damaged zone indicated by $\Omega\geq1$. Its volume grows with time due both to thermal conduction from the primary zone and to the time accumulation of damage (equ.2). This zone approximates the volume of tissue, which is expected to be coagulated during laser irradiation.

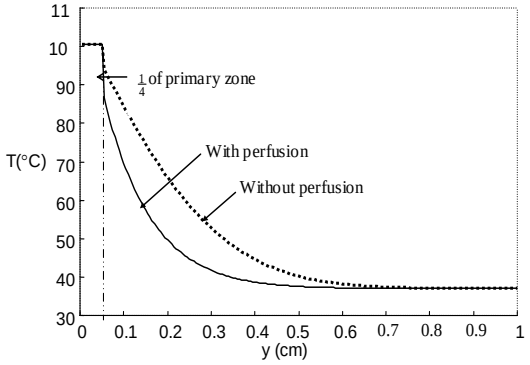

Fig.5: Blood effect on temperature profiles.

Blood perfusion ω affects temperature profiles as shown in Fig. 5. Temperature plot including blood perfusion is lower than that of the model without. Vasodilatation at slightly elevated temperature increases blood flow that behaves as a heat sink [12] and tends to cool tissues. Fig. 6 shows the effect of blood perfusion on time t_{100} required to reach a primary zone temperature of about 100°C. Increasing perfusion rate (ω) leads to larger values of t_{100} which reflects the increased cooling created by blood perfusion. Such dependency is quasi-linear.

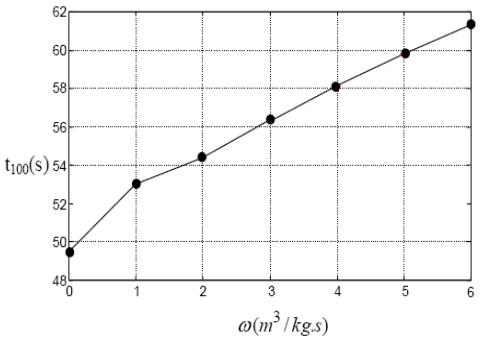

Fig.6: Effect of blood perfusion on exposition time required to approach 100 ^{0}C in the primary

Three test nodes a, b, c (Fig.7) are selected to follow temperature behaviour, internal energy and coagulation rate with time, as shown in Fig.8. It's observed that coagulation (Ω = 1) occurs approximately at the same level of internal energy (1.7 ± 0.3 J/mm^3) and temperature (73 ± 3 °C) for all nodes. The difference is in the time required for coagulation which depends on node position from primary heat source. Furthermore, coagulation at node (b) happens more quickly than at node (c) because it is located on the emission axis of laser radiation where heat flow is stronger. Such result demonstrates the importance of primary heat source geometry and its relationship to the coagulated region.

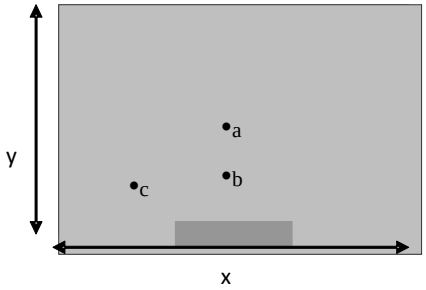

Fig.7: Position of controlled nodes in fig.8.

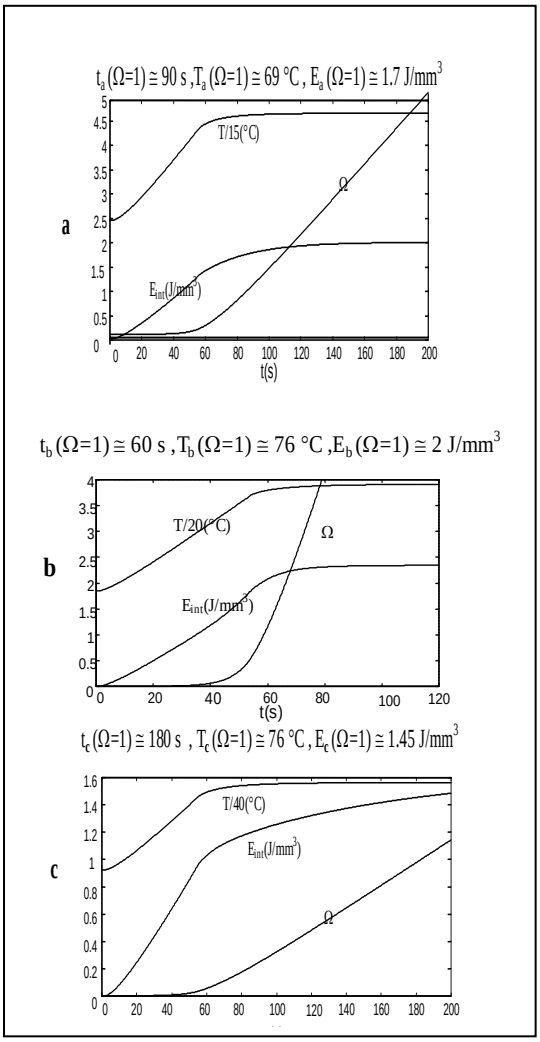

Fig.8: Temperature T, internal energy E_{int} and coagulation rate Ω versus time t; at nodes **a**, **b, c**.

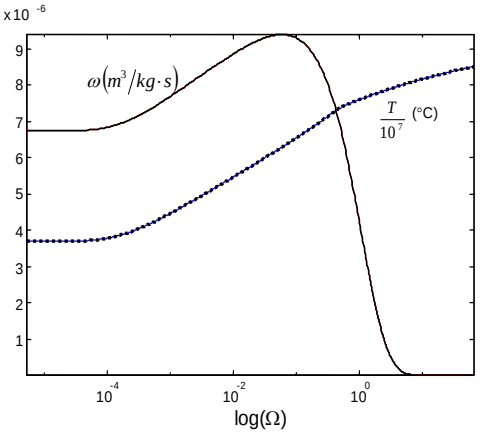

Fig.9: influence of tissue damage on blood perfusion.

To show the effect of tissue coagulation on blood perfusion ω, this one is measured at node (b) during 200s with a temperature display for easy analysis (Fig.9). We note an increase quasi linear of perfusion with temperature until the beginning of coagulation then perfusion falls rapidly to vanish completely when all tissue under investigation has coagulated.

V. CONCLUSION

Transmission line matrix (TLM) method as a potential solver for thermal transfer problems in living biological tissues, such as clinical treatment of benign prostate hyperplasia, has been demonstrated. The accuracy of the present TLM solution is validated through experimental data from other workers [3, 4, 15]. Thermal diffusion, blood perfusion and tissue coagulation are coupled phenomena that TLM model efficiently without loss of the bio-physics underhand. Results show a quasi linear dependency of perfusion with temperature until the beginning of coagulation then perfusion falls rapidly to vanish completely when all tissue under investigation has coagulated. Increasing perfusion rate (ω) leads to larger values of t_{100} and reflects the increased sinking created by perfusion. The TLM numerical model predicts the coagulation damage contours and thus has a clinical interest in therapy and as an aid for clinicians because this damage cannot be easily measured in patients.

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