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A review on micropower generation from biological system

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ABSTRACT

Supply of power for an implantable medical device through a thermoelectric generator shows potential way for a biological system. The direct utilization of the temperature difference existing whole biological body is the unique feature. This paper attempts to review the energy generation of an implanted thermoelectric generator for various physiological or environmental thermal surroundings. Also several technical approaches included to enhance the energy generation from an implanted thermoelectric generator, to shows its long-term potential energy supplier for such medical practices.

Introduction

Implantable medical devices (IMD) are extremely important today as a part of modern advanced medical technologies. Enhancing or replacing certain functions of a specific tissue or organ in the human body is their distinct clinical role. In fact, such devices have already been beneficial not only to the patient but also to healthy people who wish to adopt them to expand their biological limits in the near future such as cardiac pacemakers, cardiac defibrillators, cochlear implants and so on. The energy supply is always a major concern compared to other parameters. There is often a mismatch between the patient's lifetime and the service life of an IMD; it is caused mainly by the batteries since other parts seldom have a problem of abrasion. In the 1980s, over 95% of IMDs were powered by lithium batteries whose service life had originally been rated as 10 years[Holmes, 2001]. Unfortunately, it turned out later that many patients every year had to undergo surgery to check on the performance of the IMD or replace it with new one. Therefore, extending the service life of batteries in implanted devices has become a rather important issue.

Many efforts have been intensely made towards this goal but the situation remains largely unchanged. For instance, attention has always been paid to bio-fuel cells, on the assumption that one can use glucose as fuel in the body for an IMD[Davis and Higson, 2006]. Among the various renewable power sources, the thermoelectric micro-generator is a promising candidate because of its excellent features in terms of reliable performance, long duration and the possibility of getting energy directly from the body heat.

Recently, driven by the need to find a green and low cost method of energy generation, exploitation of the thermoelectric generator (TEG) has become a pretty hot topic throughout the world [Xi, Luo and Fraisse, 2007]. This paper is reviewed the energy generation behaviour of an implanted TEG which employs the undeniable temperature differences between the body core and the skin surface.

Study modelling:

Maximum temperature gradient across human body:

Figure shows the typical implantable TED, which is similar to an IMD. The TEG was embedded into the human body to directly utilize the heat there, i.e. the temperature difference between the body core and the skin surface. For the sake of convenience in calculation, the human body is simplified as a three-layer model consisting of muscle, fat and skin layer, respectively. The blue block structure embedded in the muscle tissue denotes the TEG.

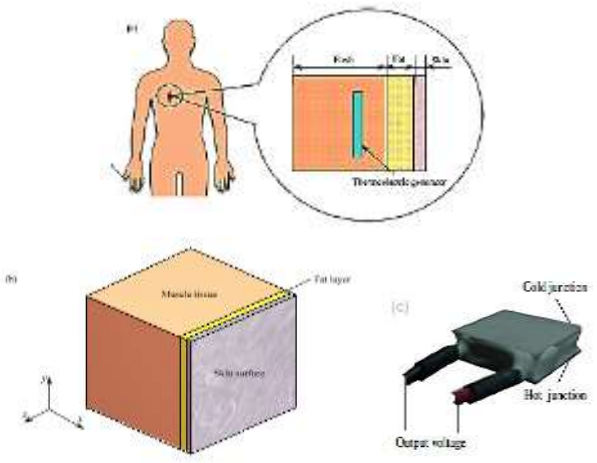

Figure 1. (*a*) Simplified three-layer human body tissues with an embedded TEG and (*b*) sketch of the 3D cubic calculation domain of size 0.08*m* × 0.08*m* × 0.08*m*. (*c*) Sketch of electricity generation experiment of a TEG (Yang Yang, Xiao-Juan Wei and Jing Liu, 2007)

The electrical voltage gained between the two junctions of the TEG could be generally characterized by the following equation,

$$V_o = n\Delta T(\alpha_1 - \alpha_2) \quad (1)$$

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Equation (1) presents the basic equation for correlating the output voltage from a TEG (without load) and the temperature difference across it. Clearly, the voltage depends on both n, α_1, α_2 and ΔT . n, α_1 and α_2 are the intrinsic properties of a TEG device. Therefore, evaluating the magnitude of ΔT will provide important information on the electrical energy generation capability. This will be done by a group of different bio heat transfer analyses on the biological body either with or without a TEG embedded. Considering a generalized one-dimensional (1D) heat transfer case in steady state, the temperature difference across the two junctions of a TEG can be approximately described by

$$\Delta T = \frac{q\delta}{k_t} \qquad (2)$$

It is obvious that the temperature difference ΔT which determines the output voltage depends only on the heat flux from the tissue since δ and k_t are the own properties of a TEG.

In the following, a one dimensional bio heat transfer model without a TEG embedded will first be adopted for simplicity to evaluate which part of the tissues would have the maximum temperature gradient. To characterize heat transfer in the living tissues, the well-known Penne's bio heat equation was used, which describes the influence of blood flow on the temperature distribution in the tissue in terms of volumetrically distributed heat sinks or sources, i.e.

$$\rho c \frac{\partial T(x,t)}{\partial t} = \mathrm{k} \frac{\partial^2 T(x,t)}{\partial x^2} + Q_b + Q_m \qquad (3)$$

The steady-state temperature field for the basal state of biological bodies can then be obtained by solving the following equations:

$$\begin{cases} k\dfrac{d^2T_0(x)}{dx^2} + \omega_b\rho_b c_b\left[T_a - T_0(x)\right] + Q_m = 0 \\ T_0(x) = T_c, x = 0, \\ -k\dfrac{dT_0(x)}{dx} = h_0\left[T_f - T_0(x)\right], x = L, \end{cases} \qquad (4)$$

The analytical solution to equation (4) has been obtained in from which the heat flux at any tissue position x could be written as

$$q(x) = -k\frac{dT(x)}{dx} = -k\left\{\begin{array}{l}\left[A(T_c - T_a - \dfrac{Q_m}{\omega_b\rho_b c_b})sh(\sqrt{A}x)\right. \\ \left.+\left[\dfrac{h_{0\sqrt{A}}}{k}T_f - T_c\right]ch(\sqrt{A}x)\right] \\ \times\left[\sqrt{Ach(\sqrt{AL})} + \dfrac{h_0}{k}sh(\sqrt{AL})\right]^{-1}\end{array}\right\} \qquad (5)$$

The derivative of the heat flux can then be expressed as

$$q'(x) = -k\frac{dT(x)}{dx} = -k\left\{\begin{array}{l}\left[A\left(T_c - T_a - \dfrac{Q_m}{\omega_b\rho_b c_b}\right)ch(\sqrt{A}x) + \right. \\ \left[\dfrac{h_0 A}{k}(T_f - T_c)\right]sh(\sqrt{A}x)]*[\sqrt{A}ch(\sqrt{AL}) \\ \left.+\dfrac{h_0}{k}sh(\sqrt{AL})\right]^{-1}\end{array}\right\} \qquad (6)$$

In equation (6), T_c and T_a are often treated as equal to 310*K*, and T_f is generally less than T_c . It can thus be easily found that $q'x$ is permanently larger than zero. In other words, the heat flux $q(x)$ is always a monotonic increasing quantity with the dimension x , which means the heat flux $q(x)$ would reach its maximum value at the position farthest from the body core.

Based on the above theoretical prediction, a conclusion can be drawn as follows. When implanting a TEG, the best place should be as close as possible to the superficial skin, where a maximum temperature difference between the two junctions of the TEG could be established. This would guarantee a good output of the TEG.

Thermal model:

Theoretical model:

Although a brief evaluation of the heat transfer in the human body has already been given as above, it represents overall only a highly simplified 1D case. A better understanding of the actual conditions requires a three-dimensional (3D) numerical simulation on the heat transfer problems of the human body embedded either with or without a TEG. For this purpose, a rectangular geometry is selected for the analysis. The calculation domain is prescribed in a 0.08*m* × 0.08*m* × 0.08*m* cube, where x denotes the tissue depth from the body core while y and z are coordinates along the surface (figure 1(*b*)). The heat transfer in a TEG can be described by a thermal diffusion equation:

$$\rho_t c_t \frac{\partial T_t(X,t)}{\partial t} = \nabla k_t \nabla\left[T_t(X,t)\right] \qquad (7)$$

Here, the self Joule heating of the TEG and heat transfer by the Peltier effect during working was not specifically considered for simplicity.

The 3D Penne's equation for characterizing bio heat transfer of the surrounding tissues reads as

$$\rho c \frac{\partial T(X,t)}{\partial t} = \nabla k(X)\nabla[T(X,t)] + Q_b + Q_m \qquad (8)$$

Where $\omega_b(x)$ is the space-dependent blood perfusion In the present model, blood perfusion and metabolic heat generation exist only in the muscle tissue.

The boundary conditions for a practical TEG implantation pattern as shown in figure 1(*b*) are prescribed as follows:

$$-k\frac{\partial T}{\partial y} = 0 \quad \text{at y=0,} \quad -k\frac{\partial T}{\partial y} = 0 \quad \text{at y=L,} \qquad (9)$$

$$-k\frac{\partial T}{\partial z} = 0 \quad \text{at z=0,} \quad -k\frac{\partial T}{\partial z} = 0 \quad \text{at z=L,} \qquad (10)$$

$$T = T_c \text{ at x=0,} \quad -k\frac{\partial T}{\partial x} = h_0(T_f - T) \quad \text{at x=L} \qquad (11)$$

The reason for adopting the adiabatic conditions on the boundaries along the y and z directions is based on the consideration that at the positions far from the centre of the domain, the temperature field is almost unaffected by the central domain or external heating and cooling. At the six interfaces between biological tissues and the rectangular TEG, a continuum equation for both temperature and heat flux is adopted, i.e.

$$T = T_t, \qquad k\frac{\partial T}{\partial n} = k_t\frac{\partial T_t}{\partial n} \qquad (12)$$

Numerical Simulation:

Validation of numerical scheme:

For the implanting case, the size of the TEG was set as a $20mm \times 20mm \times 3mm$ thin cube as shown in fig 2 and the skin surface was also subjected to the same convective boundary condition. In order to test the temperature difference formed across the TEG with different implantation positions, several distances between the hot junction of the TEG and the body core along dimensions x are respectively, prescribed as $0.02m$, $0.04m$, $0.06m$ and $0.08m$. The results in figure 2 indicate that no matter where the implanted TEG is located, there always exists a temperature difference between its two junctions. The constant temperature difference is $285K(T_f = 298K, T_c = 310K)$

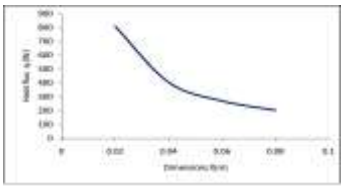

Figure 2. shows the relation between heat flux and distance from TEG to skin surface.

Effect of physiological states on TEG electricity output:

In daily life, when a person takes on various physiological states, there exists a difference between the physiological parameters. For example, the blood perfusion and metabolic heat generation, which have a significant impact on the temperature distribution in biological bodies, are widely different when a person is under different physiological activities. There exists a correlation between blood perfusion rate and metabolic heat generation $w_q = \frac{\omega_b}{Q_m} = 1 \times 10^{-3} kg\, J^{-1}$, where ω_b is the blood perfusion and Q_m the metabolic heat generation. w_q is the ratio of ω_b to Q_m .Metabolic heat rate for men who involved in the following activities: seated relaxed, car driving, shaving, volley ball playing, swimming. Metabolic heat rate for women who involved in the following activities: seated relaxed, shopping, teaching, washing dishes standing, washing & ironing.

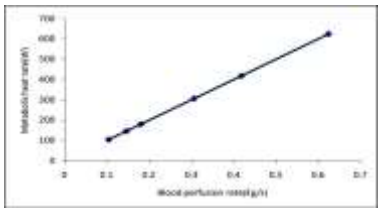

Figure 3. shows the relation between blood perfusion and metabolic heat generation rate for men .

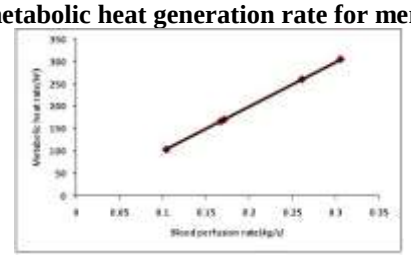

Figure 4. shows the relation between blood perfusion and metabolic heat generation rate for women.

It suggests that the larger the blood perfusion and metabolic heat generation rate, the higher the temperature difference across the TEG that can be formed.

Conclusion:

TEG are becoming more and more attractive power sources which merge the stability, reliability and economy into one assembly to fulfil a wide variety of practical needs.. In this paper, thermal modelling of TEG embedded in the human body to drive an IMD was presented.

Parametric study to be conducted in the further work to analysis the effects of various typical physiological factors, the device configuration and the environmental conditions. These results are expected to be a valuable reference for designing an implantable TEG which may actually be used in future clinics.

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