EMI Modeling for Cardiac Pacemaker in Human Body Communication

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***Abstract*— In this paper, we have proposed a two-step approach to model the electromagnetic (EM) interference voltage at implanted cardiac pacemaker. In the first step, we calculate the input voltage of the analogue sensing circuit of pacemaker using an EM field analysis tool by considering the pacemaker as a receiving antenna. In the second step, we employ a nonlinear operational amplifier model with Volterra series representation to predict the output voltage of the sensing circuit which consists of an amplifier and a low-pass filter. Comparison between the predicted result and the measured result in literature has shown the validity of the approach. Application of the approach to a user certification situation has demonstrated that the output voltage of the pacemaker sensing circuit is much smaller than the sensing threshold under usual signal intensity of on-body communications.**

Key words: Electromagnetic interference, human body communication, implanted cardiac pacemaker, nonlinear operational amplifier.

I. Introduction

As electronic devices become smaller, lower in power requirements, and less expensive, one can adorn his/her body with various personal information and communication appliances [1]. In such cases the human body acts as a transmission medium. This is known as human body communication and is attracting much attention in user certification, entertainment as well as applications in wellness and medicine [2]. For example, the human body communication allows high security and convenience by transmitting a password signal through the human body to a user certification system, because the signal radiation towards the outside of the human body can be minimized. Also, it may provide new possibility of high-quality service from hospitals by linking various biotal sensors to establish an on-body area network of personal health information.

It is considered that the signal transmission in the human body is based on an approximated surface wave or creeping wave mechanism. The dielectric properties of human tissue determine that such a transmission should be expected in a frequency below dozens of MHz. On the other hand, however, the electromagnetic (EM) interaction of on-body communication signals with the human body is significant in this frequency range because of the easier penetration of EM fields into the human body. This especially implies a potential for EM interference (EMI) to an implanted medical device such as a cardiac pacemaker. The cardiac pacemaker consists of a shielded housing with electronic circuits inside and an electrode. It is connected to the heart by an electrode to read the electrocardiogram (ECG) and to simulate the heart beat by voltage pulses if necessary. External EM fields can couple into the pacemaker to cause an interference voltage at the input of the internal sensing circuit. The induced interference voltage at the input of the sensing circuit of pacemaker will be amplified and low-pass filtered. When the output voltage of the amplifier and low-pass filter exceeds a threshold, the pulse voltage to simulate the heart beat may be triggered and a malfunction of cardiac pacemaker occurs. In the cellular phone frequency band, we have already shown that a pacemaker acts as a receiving antenna with respect to the external EM fields [3]. It should be reasonable to extend this consideration to the human body communication frequencies.

In this paper, we propose a two-step approach to model the induced EMI voltage at the cardiac pacemaker. In the first step, we calculate the input voltage of the pacemaker circuit using a full-wave EM field simulator by considering the pacemaker as a receiving antenna. In the second step, we employ a Volterra kernel description to predict the output voltage of the amplifier and low-pass filter circuit in the pacemaker for evaluating the EMI effect.

II. EM Field Modeling

Fig. 1 shows a basic configuration for an implanted cardiac pacemaker. The cardiac pacemaker consists of a shielded housing with electronic circuits inside and an electrode as well as the lead wire. By considering the internal impedance seen from the connector to the internal circuit as a load, and the metal portions consisting of the pacemaker housing and the lead wire of the electrode as two elements of a receiving antenna, the resultant equivalent circuit for the pacemaker can be shown in Fig. 2. Here, Z_R is the radiation impedance of the pacemaker, V_M is the open voltage induced between the pacemaker housing and the lead wire due to the EM fields from the external communication devices, Z_I is the internal impedance of the pacemaker seen from the connector, and V_I

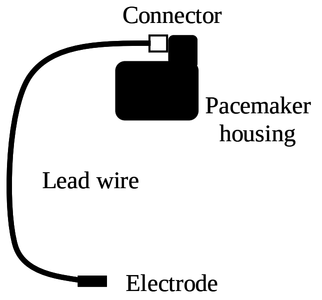

Fig.1 A basic configuration of cardiac pacemaker.

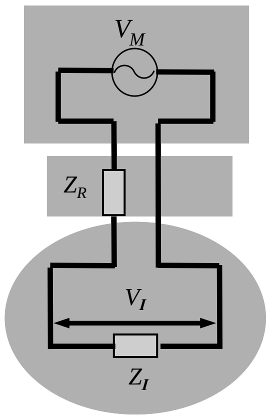

Fig. 2 Equivalent circuit for EMI of pacemaker.

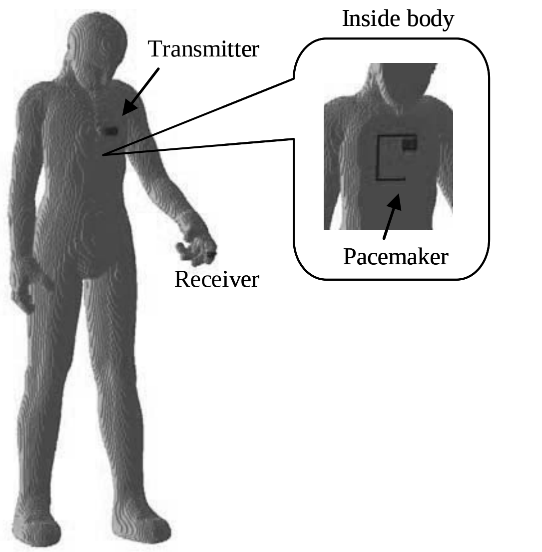

Fig. 3 A model for a pacemaker user with an on-body transmitter.

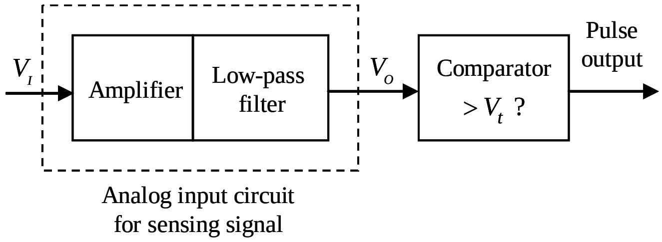

Fig. 4 Block diagram of the internal pacemaker circuit.

is the voltage induced through the connector onto the internal circuit, which is referred here as the input interference voltage to the internal sensing circuit of pacemaker, respectively.

The open voltage V_M at the connector can be obtained by simulating the pacemaker as a receiving antenna with an open load at the connector using a full-wave EM field simulation tool such as the finite difference time domain (FDTD) method. Fig. 3 shows an FDTD model for a pacemaker user with an on-body communication device near the chest. This situation imagines a user certification case. The human body model is a homogeneous one with the dielectric properties of muscle and a spatial solution of 5 mm. The on-body transmitter is an electrode structure consisting of two metal plates on the human body surface. The pacemaker is modeled as a metal housing and a metal lead line of electrode. Using the FDTD method, the open voltage V_M between the metal housing and the lead wire, i.e., at the connector, can be obtained when the transmitter electrode is excited. Then the input interference voltage V_I of the sensing circuit can be obtained as

$$V_I = \frac{Z_I}{Z_R + Z_I} V_M \qquad (1)$$

III. Electric Circuit Modeling

Fig. 4 shows the block diagram of the analogue sensing circuit of pacemaker [4]. The input signal V_I is amplified and low-pass filtered. The resulting output voltage V_o is compared with a sensing threshold V_t. When the voltage exceeds the sensing threshold, it will switch the pulse output, and then yield a malfunction of pacemaker.

The amplifier and low-pass filter can be considered as an operational amplifier (opamp) combined with external elements. Fig. 5 shows a representative negative feedback opamp configuration. Although the on-body communication signal is generally at higher frequencies compared to the heart-beat-simulating pulse signal, due to the demodulating properties of the opamp, high frequencies can be down converted and are able to pass through the low-pass filter. A nonlinear model is therefore necessary for the opamp with low-pass filter character. The Volterra kernel is known as a powerful tool to describe a nonlinear system. Assuming the circuit configuration in Fig. 5 as a second order nonlinear system, we can obtain the interference voltage induced by a narrow-band on-body communication signal at carrier frequency f_c as [5]

$$\begin{aligned} v_o(t) &= V_I H_2(f_c)\cos(2\pi f_c t + \angle H_2(f_c)) \\ &\quad + 0.5 V_I^2 \operatorname{Re}\{H_{22}(f_c, -f_c)\} \end{aligned} \qquad (2)$$

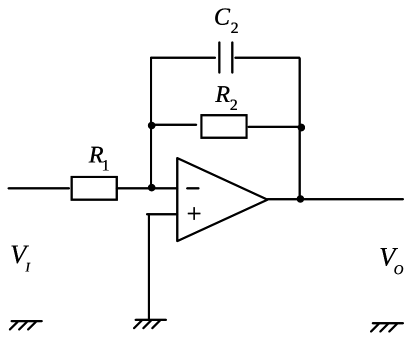

Fig. 5 Negative feedback operational amplifier configuration.

where $H_2(f_c)$ and $H_{22}(f_c,-f_c)$ are the frequency domain Volterra kernels, that depend on the opamp parameters and the external components. Since the term at f_c is beyond the opamp circuit bandwidth, only the second term in Eq. (2) is effective. This indicates that the EMI effect is actually an offset in the output voltage. For a MOS transistor structure of the circuit in Fig. 5, after a complicated mathematical derivation process, we have the interference output voltage V_o as

$$V_o = 0.5V_I^2\ \mathrm{Re}\{H_{22}(f_c,-f_c)\}$$
$$= V_I^2 A_0 \frac{g_m}{4I_0} \mathrm{Re}\left\{ \frac{j2\pi f_c C_T}{j2\pi f_c(2C_{gs}+C_T)+2g_m} \times \left[\frac{Z_2(f_c)}{R_1+Z_2(f_c)}\right]^2 \right\} \quad (3)$$

where A_0 is the amplification factor, I_0 is the bias current at the transistors, g_m is the transfer conductance of the opamp, C_{gs} is the gate-to-source capacitance of each transistor, C_T is the sum of the parasitic capacitances related to the ground and the power supply, and $Z_2(f) = R_2/[1+j2\pi f_c R_2 C_2]$, respectively.

According to [4], the low-pass filter of the pacemaker circuit has a cut-off frequency of 1 kHz. Assuming a 10 dB gain of the negative feedback opamp circuit, we determined the component parameters for the circuit in Fig. 5, i.e., R_1=1 kΩ, R_2 = 3kΩ, and C_2 =53.05 nF. Furthermore, with typical values of A_0, I_0, g_m, C_{gs} and C_T, we can obtain the output voltage V_O for a known interference input voltage at the pacemaker sensing circuit.

IV. Validation

According to [6], the mean value for the sensing threshold voltage V_t of a pacemaker is around 2.2 mV. Based on this sensing threshold level, we determined the measured output voltage V_O from the input voltage V_I, which was almost 1 V from 10 to 100 MHz, as given in [4]. The white circles in Fig.

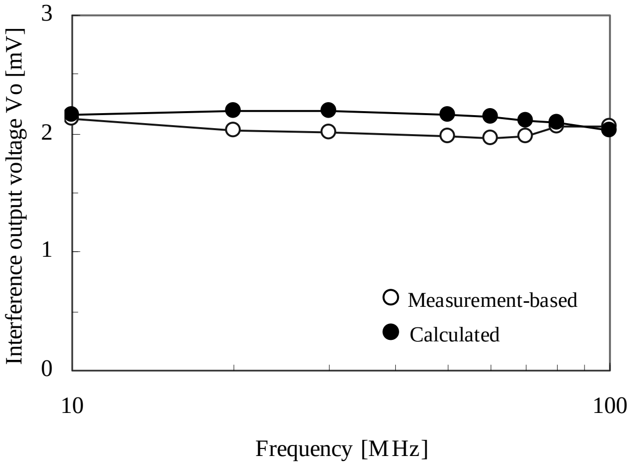

Fig. 6 Comparison of measurement-based and calculated output voltages *Vo*.

6 show the measurement-based output voltage as a function of frequency between 10 MHz and 100 MHz. It can be found that the output voltage is almost flat within this frequency band and has a constant level of about 2.2 mV.

On the other hand, based on the proposed nonlinear opamp model shown in the previous chapter, we calculated the output voltage V_O using Eq. (3) for the same input voltage V_I as given in [4]. The results are also plotted in Fig. 6 with black circles. The predicted interference output voltage is found to agree well with the measurement-based one, which assured the validity of the nonlinear circuit model. The circuit parameters used in the model are listed in Table I.

Table I Opamp circuit parameters

Amplification factor	A_0	1,000,000
Bias current	I_0	10 μA
Transfer conductance	g_m	1.2 mS
Gate-to-source capacitance	C_{gs}	100 fF
Parasitic capacitance	C_T	1 pF

V. EMI Evaluation

Based on the above consideration, the predication of EMI voltage at the analogue sensing circuit output of pacemaker takes two steps:

1) Calculate the open voltage V_M using the FDTD method by modeling the pacemaker as a receiving antenna. Since the radiation impedance Z_R is much smaller than the input impedance Z_I of the opamp circuit, based on Eq. (1), it is reasonable to use the open voltage V_M as the input voltage V_I, which actually considers a worst case.

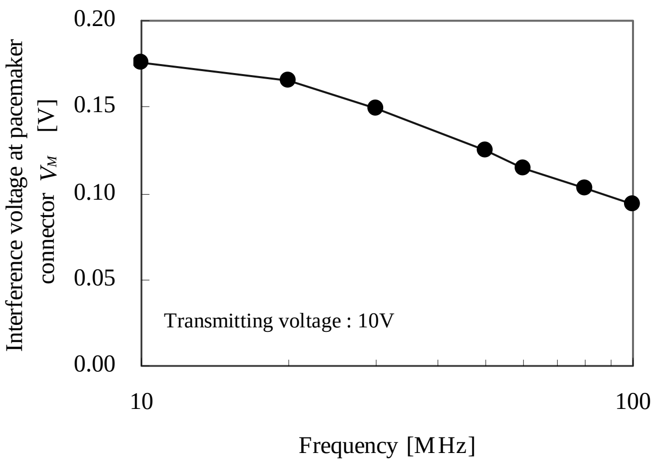

Fig. 7 FDTD-calculated interference voltage V_M at the pacemaker connector. This voltage is almost equal to the input voltage V_I .

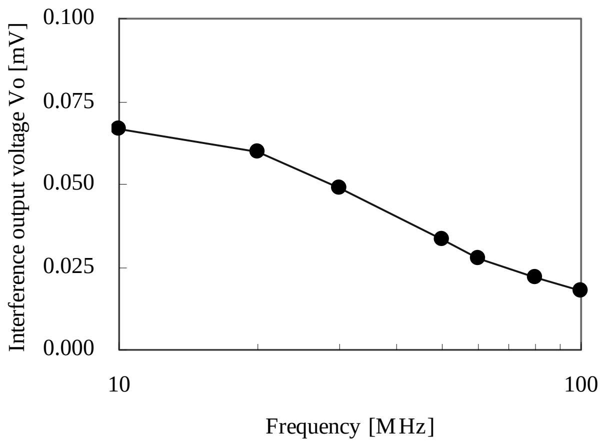

Fig. 8 Predicted interference output voltage of the pacemaker circuit in on-body communication.

2) Calculate the output voltage V_O using Eq. (3) at the carrier frequency and compare it with the sensing threshold V_t. If $V_O > V_t$, the heart-beat-simulating pulse will be triggered and a malfunction may occur.

This approach was applied to a user certification situation as shown in Fig. 3, in which a pacemaker user adorned his chest with an on-body transmitter. Since the transmitter was just on the chest surface, induced EMI voltage to the implanted pacemaker should have a significant level. Fig. 7 shows the FDTD-calculated open voltage V_M as a function of frequency from 10 to 100 MHz for an exciting voltage of 10 V at the transmitter. This exciting voltage should be the maximum level in usual on-body communications. As can be seen, the induced interference input voltage decreases with the frequency, which may be attributed to the more difficult penetration into the body tissue at higher frequencies. Within the interested frequency band, the open voltage V_M ranges from 0.18 V to 0.09 V. These voltages were added to the sensing circuit input as V_I approximately.

Fig. 8 shows the predicted output voltage of the analogue sensing circuit using the nonlinear opamp model, i.e., Eq. (3). It can be found that the output voltage ranges from 0.07 mV to 0.02 mV within this frequency band. Compared to the sensing threshold V_t of 2.2 mV, there is a safety margin of at least 30 dB between 10 to 100 MHz. This result suggests that it is unlikely to have a possibility of pacemaker malfunction under usual transmitter voltage level in on-body communications.

VI. CONCLUSION

EMC consideration is essential for the design of an on-body communication system. The potential of EMI to an implanted cardiac pacemaker by on-body communication signals is especially watched with interest. As a method to predict the EMI level of pacemaker, we have proposed a two-step approach, consisting of an EM field analysis and a nonlinear electric circuit analysis. The EM field analysis employs the FDTD method to calculate the interference voltage which couples into the pacemaker circuit through the connector, while the electric circuit analysis employs a nonlinear opamp model with Volterra series representation to predict the output voltage of the analogue sensing circuit of pacemaker. The output voltage of the sensing circuit may switch the pulse generator and yield a malfunction when it exceeds a threshold. The predicted interference output voltage has shown good agreement with the measured one in literature, which confirms the validity of the proposed approach. Furthermore, applying this approach to a user certification situation, we have obtained the output voltage from 10 to 100 MHz under a considerably strong signal intensity of on-body communication, and demonstrated that it is much smaller than the actual sensing threshold of pacemaker, i.e., with a safety margin of at least 30 dB.

The future subject is to extend this approach to higher frequency bands.

REFERENCES

[1] T.G. Zimmerman, "Personal area networks: Near-field intrabody communications," IBM System Journal, vol.35, nos. 3&4, pp.609-617, 1996.

[2] P.S. Hall and Y. Hao, Eds., "Antennas and Propagation for Body-Centric Wireless Communications," Artech House, 2006.

[3] J. Wang, O. Fujiwara and T. Nojima, "A model for predicting electromagnetic interference of implanted cardiac pacemaker by mobile telephones," IEEE Trans. Microwave Theory Tech., vol.48, no.11, pp.2121-2125, Nov. 2000.

[4] S. Schenke, L.O. Fichte and S. Dickmann, "EMC modeling of cardiac pacemakers," Proc. Int. EMC Zurich Symp., Munich, Germany, Sept. 2007.

[5] F. Fiori and P.S. Crovetti, "Prediction of EMI effects in operational amplifier by a two-input Volterra series model," IEE Proc. Circuits Devices Syst., vol.150, no.3, pp. 185-193, June 2003.

[6] W. Irnich, L. Batz, R. Muller and R. Tobisch, "Electromagnetic interference of pacemakers by mobile phones," PACE, vol.19, pp. 1431-1446, vol.19, Oct. 1996.