EXTRACTING ELECTRIC POWER FROM HUMAN BODY FOR SUPPLYING NEURAL RECORDING SYSTEM

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Abstract - A powerful approach to the characterization of cellular electrical activity is electrical recording from cells or living tissues. The human central and / or peripheral nervous system has been a subject of study and fascination of the neuroscience and biomedical engineering communities for many decades. In this paper, we propose a new approach to feed implantable neural recording system, which based on extracting electrical power from human tissue warmth in order to supply a biomedical neural recording system. The major issue to overcome, in the design of a system that is aimed at being implant into the human body, is having a low power consumption, low noise circuit and small dimension to minimize tissue damage.

Index terms*:* Neural recording system, thermoelectric power generation, low power, neural amplifier.

I. INTRODUCTION

One of the crucial aspects is supplying energy to the implantable neural recording system. There are three different approaches; the first one is total implantable battery [1] [2], the second one is percutaneous transfer, the third one is transcutaneous transfer [3].

Batteries, in the first approach, are not the optimal choice for implantable devices, because their lifetime is limited, and they are usually large and leaks can pose a hazard to tissues. This would require periodically replacing the battery and require additional surgeries, and is not a viable solution. The second approach, has the disadvantage that wires cause a permanent breach of the skin's natural barrier to infection. Bacterial infection is therefore a common complication of such implants. For most of these long-term implantable devices, an external transcutaneous wireless link is preferred to an internal battery and to percutaneous wires for the power supply. Crucially, the implant will be powered by inductive coupling by an external power source. This paper faces the third aspect by designing a system capable of supplying the neural recording system by means of a mixture between external battery and energy harvested from human body temperature [4] [5].

The measurement and processing of electrophysiological signals are a fundamental task performed by neuroscientists, as a means to gaining an insight into how biological systems respond to external stimuli and communicate with each other. Recently, the advent of microelectrode arrays has created the need to design low-power implantable electronic systems that are capable of recording neural activity while not causing permanent damage to the neural systems under observation. Such systems would prove invaluable in furthering our understanding of the electrical function of neurons with respect to drug interactions and mental functions and their interactions with other neurons and muscles. This has made possible the developing of a variety of implantable bio-system applications such as treating spinal cord injuries, deep brain stimulation to treat Parkinson's disease and detection of epilepsy [6]-[11]. These systems require a long term simultaneous recording of neural activity from many neurons simultaneously. The ideal system for long term recording would be a fully implantable device which is capable of amplifying the neural signals and transmitting them to the outside world [12] [13] [14]. One of the most important parts in the development of the neural recording system is the neural signal amplifier. Neural signals from extracellular recording are very weak (typically between 50μV and 500 μV). As a result, amplification is needed before they can be processed further. This in turn drives the need for the integration of low power electronic circuitry. We describe a low voltage low power CMOS amplifier that addresses this need by rejecting DC offsets, and has tunable bandwidths.

This work investigates the design of the electronic components of such an implantable recording system. Low-power and low-noise neural recording amplifier is designed.

The organization of the paper is as follows. After a general introduction of the aspects for supplying energy to the implantable system power system and the need to get a fully implantable device which is capable of amplifying the neural signals and transmitting them to the outside world wirelessly. The block diagram of a wireless neural recording system has been discussed in section II. Energy harvesting from human body temperature has been presented in section III. In section IV, the characteristics of neural recording amplifier have been presented. The simulation results, of the neural amplifier with adjustable gain and low cutoff frequency, have been discussed in section V. The paper has been concluded in section VI.

II. OVERVIEW OF THE WIRELESS NEURAL RECORDING SYSTEM

The block diagram of the neural recording system is shown in Figure 1. The extracellular electrical activities of single or multiple neurons, in proximity of small recording sites on a micro-electrode array, create neural signals in the range of 50~500 µV. Low-power and low-noise amplifiers with built-in bandpass filtering capability amplify the neural signals in the desired frequency range [15] [16]. The amplified signal will be processed and transmitted to external unit wirelessly through a miniaturized antenna. The implantable part is supplied by power and data via an inductive link [17]. It consists of two coils, forming a loosely coupled transformer. The primary coil, placed outside the body, generates a magnetic field which is partly picked up by the secondary coil that is implanted. In this way, power and data can be transferred wirelessly, overcoming the skin barrier. The implanted device is externally controlled and powered by a modulated radio frequency (RF) signal. The inductive antenna (receiver coil) picks up the transmitted RF carrier, followed by a rectifier, an AC/DC-voltage converter and regulator, an ASK demodulator for recovering transmitted data and a clock recovery. The clock is generated by rectifying the received RF signal and dividing it down. Generating the clock, in this way, ensures that it does not drift with respect to the external transmitter and control circuitry. The external receiver unit picks up the neural signal, digitizes it, and transfers it to the PC for further signal processing, storing and visualization.

The external control unit is responsible of the mode selection and generation of command words to the implant. It contains a control data encoder, an ASK (Amplitude Shift Keying) modulator and a class E power amplifier. The new approach discussed in this paper consists of the use of two batteries. Battery 1 is a commercial one, while the battery 2 is used to stocks the power from the thermoelectric power generation Sensor array ‘MPG-D602’.

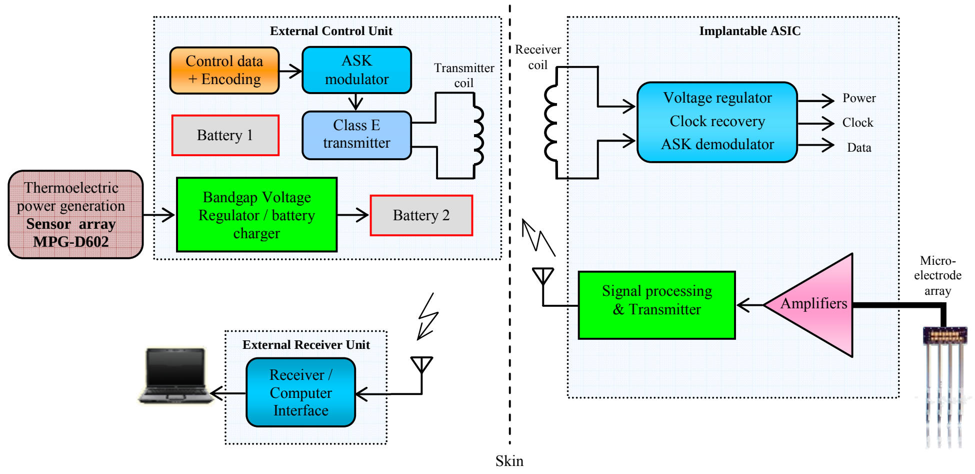

Figure 1. Overview of fully implantable neural recording system using a thermoelectric power

III. ENERGY HARVESTING FROM HUMAN BODY TEMPERATURE

The human body is subject to the same laws of physics as other objects, gaining and losing heat by conduction, convection and radiation. Conduction between bodies and/or substances in contact; convection involving the transfer of heat from a warm body to a body of air above it or inside the human body and here the blood, gases and other fluids is the medium, radiant heat transfer is a major mechanism of thermal exchange between human body and the surface surrounding environment. These three effects in most situation operate together. In human body, metabolic processes generate its own heat as well, similar to a heat-producing engine. Human body behavior try to be in stable state therefore, it absorbs and emits energy to be in equilibrium, stimulation is applied to the body surface, this make the activity of metabolism induced to body surface[18].

In different works, the changes in a part of human body being have been studied and analyzed before and after stimulation and compared between them, then simulate the bio heat transfer mechanism using 2nd order circuit, which designed based on 1st order introduced by Guotai et al. [19], and analyze the human thermo response. G.Jiang, A.Shang and N.Umekawa in [20] studied the diagnosis of diabetes by thermograph. Since the human body emits energy as heat, it follows naturally to try to harness this energy. However, Carnot efficiency puts an upper limit on how well this waste heat can be recovered. Assuming normal body temperature and a relatively low room temperature (20° C), the Carnot efficiency is

$$\frac{T_{body} - T_{ambient}}{T_{body}} = \frac{(310K - 293K)}{310K} = 5.5\% \tag{1}$$

In a hot environment (27°C) the Carnot efficiency falls to

$$\frac{T_{body} - T_{ambient}}{T_{body}} = \frac{(310K - 300K)}{310K} = 3.2\% \tag{2}$$

According to mathematical viewpoint, heat diffusion in human body can be represented according to a system of eight differential equations of a structural human body model [21], and the human body head is the area with the highest temperature. Consequently, we exploit the head skin to locate the sensors for gathering electric energy from body temperature. We used a thin film generator named MPG-D family and specifically MPG-D602 according to the characteristics of Table 1.

Table 1: MPG-D602 characteristics

Type	Dimension (mm)	Number of leg pairs	Thermal resistance	Electrical resistance	Substrate type	Thickness
MPG-D602	Cold side: 2.47x2.47 Hot side: 2.47x2.47	450	9.6 K/W	189Ω	Silicon	500µm

The MPG-D is a thermoelectric power generator based on the transfer of the thermal energy through a minimum of one leg pair consisting of p-type and n-type thermoelectric material. Micropelt utilizes Bismuth (Bi), Antimony (Sb), Tellurium (Te) and Selenium (Se) compounds that have the best material properties with operating temperatures around room temperature and up to 200 °C. The produced output voltage is direct proportional with the number of leg pairs and the applied temperature difference ΔT over the element. The resulting voltage U is given by the following equation, where α is the Seebeck coefficient in µV/K (material related) that influences the output voltage (see figure 2)

$$U = N_{legpairs} . \Delta T . \alpha \tag{3}$$

The circuit connections of the MPG-D are illustrated in figure 3 and the real dimensions of MPG-D602 is depicted in figure 4. The efficiency of a thermoelectric device is given by the material properties which are combined in a figure of merit F given by the following equation

$$F = \alpha^2 T \frac{\sigma}{k} \tag{4}$$

where T is the absolute temperature, σ is the electrical conductivity and k the thermal conductivity. As aforementioned, the most widely used material for the fabrication of thermoelectric generators operating at room temperature is BiTe, which exhibits a F of 1. PolySiGe (F=0.12) has also been used, especially for micromachined thermoelectric generators [22, 23]. Research on nanostructured materials and multilayers is ongoing worldwide in order to optimize thermoelectric properties and F values as large as 3.5 have been reported in many researches. These encouraging results may replace BiTe in the long term. Apart from improving the material properties, miniaturization using micromachining is ongoing [24, 25] and the main challenges of micromachined energy harvesters are listed in [26]. Selected device results reported in literature [27]. The reported power levels however cannot be directly compared, as output values are often calculated using a well-defined temperature drop across the thermopile (i.e. the temperatures of both plates have been fixed). In real applications the temperature drop across the thermopile is lower than the one between the hot plate and the ambient, and therefore the extrapolated results are too optimistic. It has been shown that the most challenging task in designing an efficient thermoelectric converter consists in maximizing this temperature drop across the thermopiles [28].

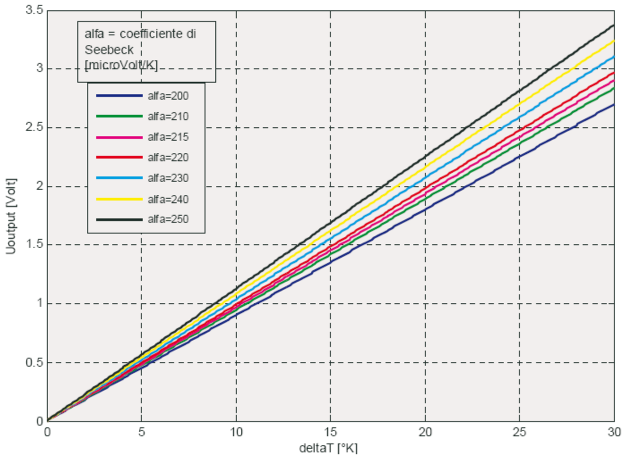

Figure 2. MPG output in function of Seebeck coefficient

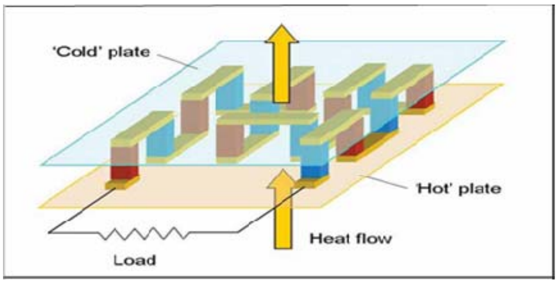

Figure 3. Circuit connections

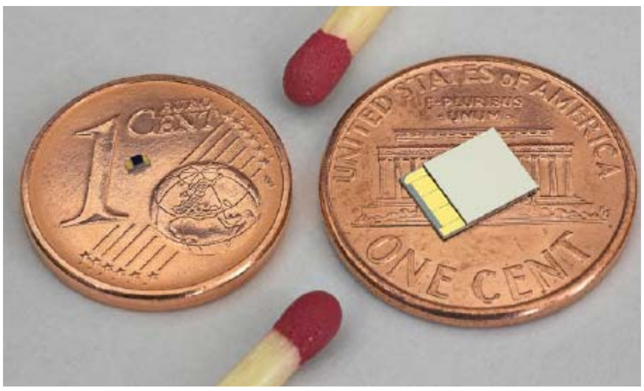

Figure 4. MPG-D601 real dimensions

IV. NEURAL RECORDING AMPLIFIER

The most critical block in neural recording system is low-power low-noise neural amplifier which is the first stage in the neural recording system. There have been considerable research efforts in the design of low-power low-noise neural amplifiers in recent years.
Figure 5 shows the schematic of the neural recording amplifier is composed of a differential stage and an output gain stage.

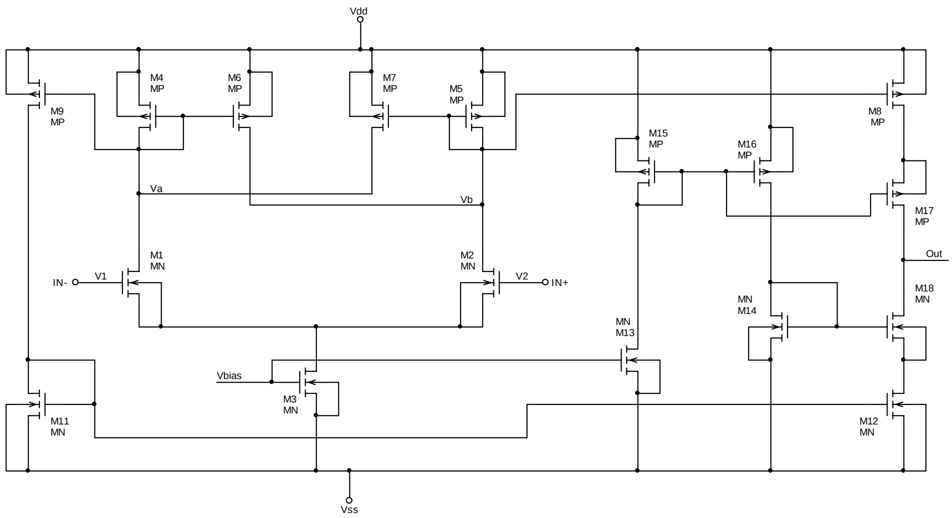

Figure 5. Schematic of the neural amplifier circuit used in this design

Referring to figure 5, we calculate the size of different devices. Table 2 illustrates the parameters values of the neural recording amplifier.

Table 2 neural amplifier parameter values

Devices	Value	
M1-M2	W=100μm	L=1μm
M3, M10, M11, M12, M13, M18, M14	W=1.5μm	L=1μm
M4, M5, M8, M9, M15, M16, M17	W=6μm	L=1μm
M6-M7	W=5μm	L=1μm
Supply voltage	Vdd=1V	Vss=-1V
Load capacitance	C_L=7pF	
Bias current	Ibias =1μA	

The circuit, given by figure 5, is simulated using PSPICE simulator with MOS transistors models BSIM3v3 level 7. We establish various characteristics such as gain, phase margin, power, noise, CMRR, PSRR, and offset. Table 3 summarizes the simulation results of the open loop neural recording amplifier.

Table 3. Simulated Performance Characteristics of neural amplifier

Parameters	Value
Gain	$G_{dB} = 43.95$ dB
Phase margin	$M\varphi = 64.6°$
Gain-BandWidth	GBW = 1.1294 MHz
Power Consumption	$P_{tot}= 9.22$ µW
Total Current absorbed	$I_{tot}=4.617$ µA
Common Mode Rejection Ratio	CMRR=113.27dB
	CMRR > 92dB @ f < 10KHz
Power Supply Rejection Ratio	$PSRR_{Vss}=73.11$dB
	$PSRR_{Vdd}=75.29$dB
Output-voltage swing	CMR+=806mV
	CMR-=992mV
DC Offset	196µV
Input Referred Noise	Vni,rms = 14.8 µVrms
Noise Efficiency Factor	NEF=13,22

V. NEURAL AMPLIFIER WITH ADJUSTABLE GAIN AND LOW CUTOFF FREQUENCY

A- Closed loop configuration

Figure 6 shows the schematic of the closed loop neural amplifier configuration which composed of an amplifier and a feedback network. The negative capacitive feedback is widely used because there are electrochemical effects at the electrode-tissue interface, DC offset of 1-2 V are common across differential recording electrodes [29].

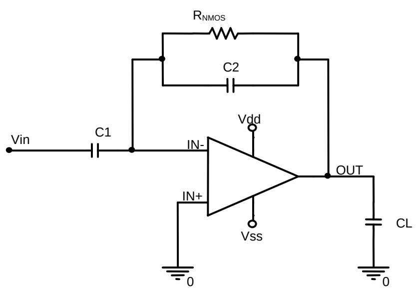

Figure 6. Closed loop amplifier

The gain A_v of the amplifier is given by the following equation,

$$A_v = \frac{-R_{NMOS}C_1}{\frac{1}{j\omega} + R_{NMOS}C_2} \tag{4}$$

The midband gain is set by the ratio of the two capacitances in the feedback network (5),

$$A_v \approx \frac{-C_1}{C_2} \tag{5}$$

The low cutoff frequency (f_L) is given by equation (6),

$$f_L = \frac{1}{2\pi R_{NMOS}C_2} \tag{6}$$

- **Frequency response**

For A_M=150 (so A_M=43.52 dB) and C2=0.1pF, by using equation 16 then C1=15pF. In the case of a low cutoff frequency equal to 100 Hz, so R_{NMOS}=19.92GΩ (by using equation 6). Figure 7 shows the frequency response of the closed loop configuration.

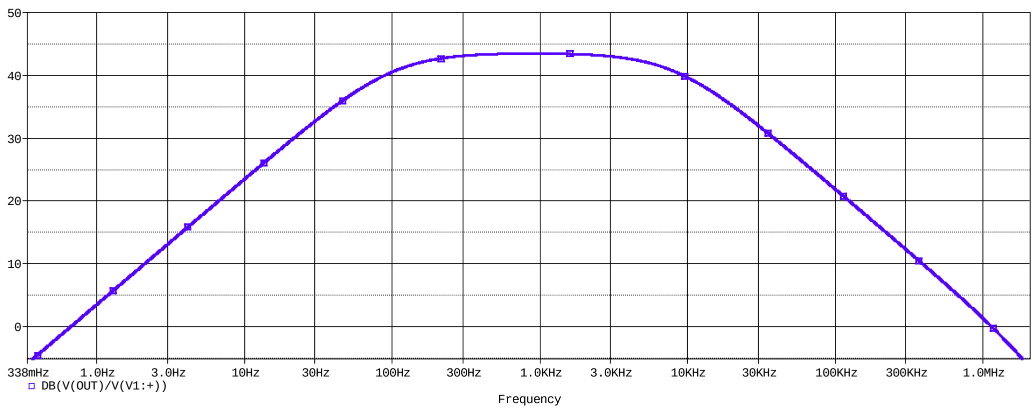

Figure 7. Closed loop amplifier's frequency response

From figure 7, the maximal gain is 43.458 dB, the low cut off frequency is equal to 98.175 Hz and the high cut off frequency is equal to 8.37 kHz.

- **Temporal response**

Extracellular neural action potentials contain frequency components from few Hz to 10 kHz and amplitudes in 50~500µV range. Temporal response are given in figure 8 and figure 9 respectively for $V_{in,max}$=50µV and $V_{in,max}$=500µV at frequency of 1 kHz.

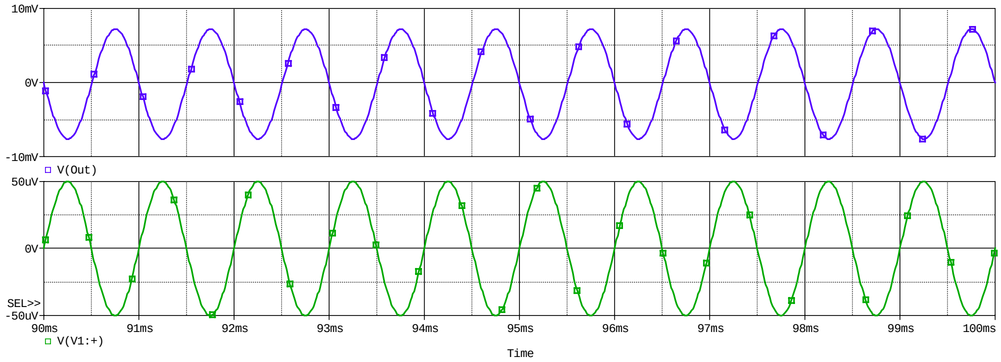

Figure 8. (a) Output voltage (b) Input voltage (50µV)

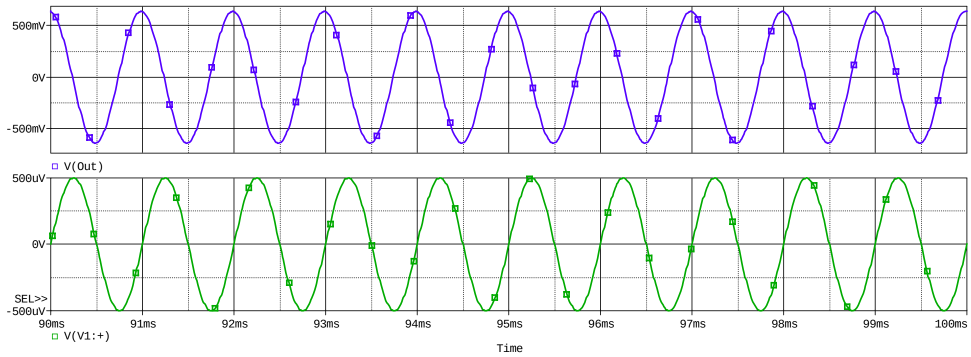

Figure 9. (a) Output voltage (b) Input voltage (500µV)

The figure 8. (b) gives an output maximum voltage value of 7.88 mV. And figure 9. (b) gives an output maximum voltage value of 788.8 mV.

B- Tunable Low Cutoff Frequency

The bio-potential signals are near to each other in particular at low frequencies. Hence, the idea is to vary the low cutoff frequency of the amplifier. The closed loop amplifier configuration can be used with more flexibility by varying the low cutoff frequency. By referring to equation 6, and for changing the low cutoff frequency, it is possible by varying the value of R_{NMOS}. The subthreshold-biased MOSFET's used to implement R_{NMOS} in figure 10 is adjustable via the tuning voltage, V_{band} [30]. The low- cutoff frequency can be trimmed in this design by adjusting V_{band}.

R_{NMOS} is done by two transistors M19 and M20 given in figure 10. By varying the voltage V_{band}, the value of the pseudo-resistance varies, therefore the low cutoff frequency varies. Simulation results show that if V_{band} sweep from 70 to 125 mV with a step of 5 mV, then the low cutoff frequency change from 21.36 Hz (for V_{band} = 70 mV) to 100.45 Hz (for V_{band} = 125 mV). Figure 11 shows the gain curve of the amplifier for V_{band} changing from 70 to 125 mV with a step of 5 mV. V_{band} voltage is generated by Digital to Analog Converter, encoding the value of the low cutoff frequency for amplifiers.

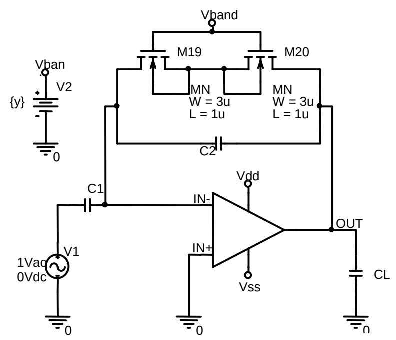

Figure 10. Closed loop amplifier with pseudo-resistor

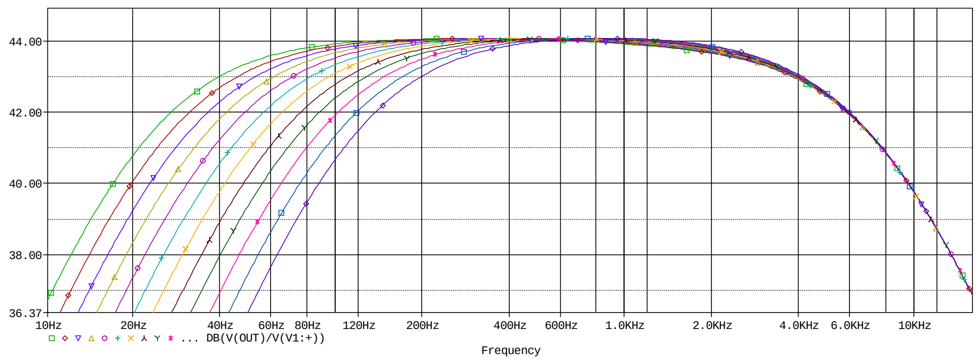

Figure 11. Variable low cutoff frequency

C- Adjustable Gain

By analyzing the equation 5, and to vary the gain there are two solutions: varying the value of C2 or the value of C1. If we vary C2, the low cutoff frequency will change. Hence the solution is to vary C1. For V_{band} = 125 mV, and by setting the value of the capacity C1 {15pF, 25pF, 35pF, 45pF}, we obtain a gain in band equal to 43.6dB, 48dB, 50dB and 52.8dB, respectively (see figure 12).

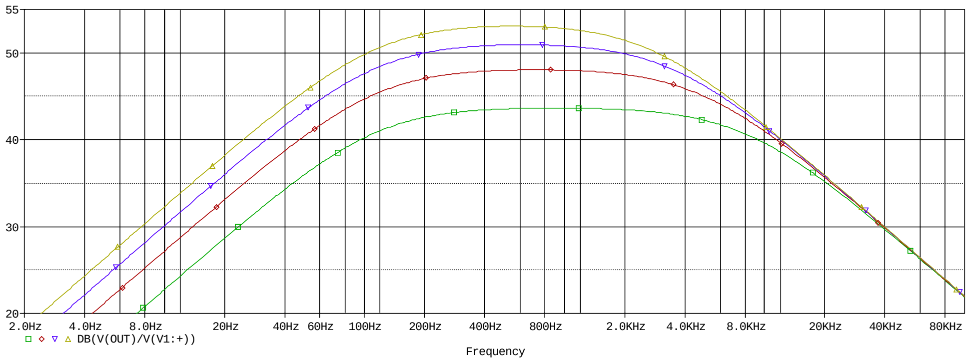

Figure 12. Variable gain

VI. CONCLUSION

In this paper, we proposed a new approach to feed implantable neural recording system, which based on extracting electrical power from human tissue warmth in order to supply a biomedical neural recording system. We presented a low power instrumentation amplifier for fully implantable neural recording system. The architecture of the neural recording system was discussed. Furthermore, Pspice simulation using a real transistor model was presented. A power dissipation of about 9.22 μW. The low cutoff frequency is adjustable from 21 Hz to 100 Hz, with four tunable gains of 43.6 dB, 48 dB, 50 dB and 52.8 dB.

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