

A COCHLEAR SYSTEM WITH IMPLANT DSP

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ABSTRACT

Cochlear implants have achieved great success in restoring hearing to profoundly deaf people. New generation implants pose more restrict requirements on processing precision and make traditional cochlear systems insufficient to deal with the large amount of data which should be transmitted via the wireless link. A new cochlear system with implant DSP is proposed to address the data rate problem. This system needs to transmit only voice-band signals with a data rate of 100 Kbps and the data rate bottleneck is removed for future development. By optimizing the speech processing algorithm and the inductive wireless link, the power consumption of this system increases less than 10% when compared with the traditional system.

1. INTRODUCTION

A cochlear implant is a prosthetic device implanted in the inner ear that can restore normal hearing to profoundly deaf people [1]. Fig. 1 shows a block diagram of a traditional cochlear system. Normal sound is first picked up by the microphone and digitalized. The digital signal processor (DSP) is then used to analyze the sound and encode results into instruction trains. The transmitter modulates the trains and sends both data and energy to the implant circuit. The implant receiver recovers data and energy from the RF signals, decodes the instructions and accordingly directs the stimulator to generate electrical stimuli. Finally, the stimuli are applied to electrodes array to excite the auditory nerve. The fully implementation of this system is detailed in [2] and [3].

However, with regard to the absorption and the protection of human body, the frequency of the carrier wave in the wireless link can't be high [4]. The data rate is hence limited. For example, a data rate of 500 Kbps in 10 MHz carrier is used for a 16-channel implant system with stimulating precision of 10 bits and stimulating rate of only about 900 pulses/channel [2]. The data rate should be proportionally increased with channel number and pulse rate. Although there are demands for more channels and higher pulse-rate [1], little space is left to raise the data rate. For

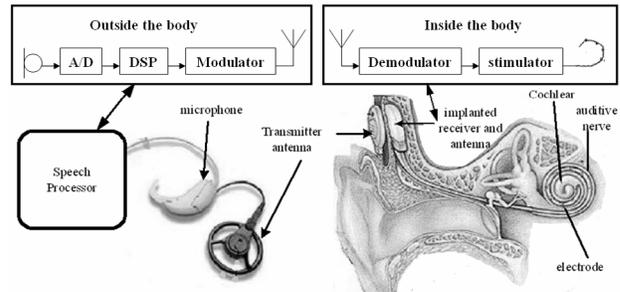


Fig. 1. Block diagram of traditional cochlear system

any modulating method, there is always a maximum data rate with a certain carrier frequency.

Fully implanted cochlear systems are considered as the next generation cochlear devices and have no trouble with the data rate. Because of the very stringent requirements on power consumption, fully implanted systems usually use analog processing method [5][6]. Nevertheless there are obvious disadvantages for analog processors: they are dedicated to a certain processing algorithm and vulnerable to noise. Although A/D-then-DSP systems can provide a flexible and robust solution, they seem to consume too much power for fully implants even with the state-of-art technology.

Based on these considerations, this paper proposes a new system with an implant DSP that can solve the data rate problem and also alleviate the power requirement. The system is powered by external battery and only voice-band signals with data rate of 100Kbps are required to be transmitted via the wireless link. Optimizations on the speech processing algorithm and the inductive wireless link are carried out to ensure the low power dissipation of the system.

This paper is organized as follows. In Section 2, the system overview is given. In Section 3, an improved implementation of the CIS algorithm is described. In Section 4, the power transmission efficiency of the antenna is discussed. A comparison of the proposed and the traditional systems is made in Section 5. And the conclusion is finally drawn in Section 6.

2. SYSTEM OVERVIEW

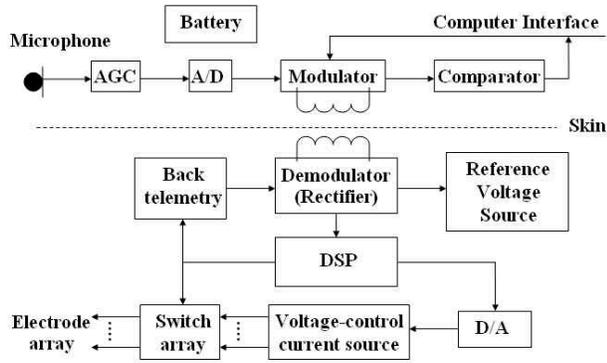


Fig. 2. Block diagram of proposed cochlear system

The proposed system consists of an external part and an implant part. The external part combines microphone, automatic gain control amplifier (AGC), A/D converter and modulator while the implant part is composed of demodulator, DSP, D/A converter and stimulating circuit. These two parts are connected by an inductive link, via which power and data are transferred from the external circuit to the implant using a PWM scheme. A block diagram of the complete system is shown in Fig. 2. The system works similarly as the traditional system. A computer interface is provided to program the implant DSP and perform back telemetry.

The most essential advantage of this system is that only voice-band signals with very low data rate, instead of large amount of processed stimulating data, should be transmitted via the wireless link regardless of channel number and stimulating frequency. But the cost is that it may consume much more power when compared with the traditional system. So to reduce the power consumption is the most concerned factor for the design of the proposed system.

The implant DSP is the key component because it may be the most power-hungry part. Careful considerations on signal-processing algorithm optimization and hardware architecture design of the DSP should be carried out to reduce the power consumption. An application-specific instruction set and proper data bus width can bring the DSP to a very low power status.

The wireless link is another key component, of which the power transmission efficiency is critical to the power consumption of the whole system. Design efforts should be exerted to raise the power transmission efficiency.

3. SIGNAL-PROCESSING STRATEGY

3.1. CIS algorithm

Among various signal-processing strategies developed for multi-channel cochlear prosthesis, the Continuous Interleaved Sampling (CIS) approach [7] achieves good

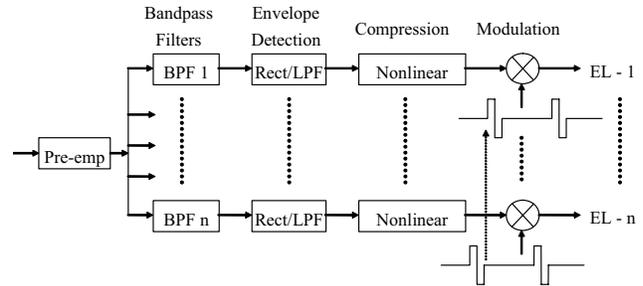


Fig. 3. Structure of CIS algorithm

performance and is being widely used in implant devices. The structure of the CIS strategy is shown in Fig. 3.

The sound signals are first pre-emphasized and passed through a bank of band-pass filters. The envelopes of the filtered waveforms are then extracted by full-wave rectification and low-pass filtering. The envelope outputs are finally compressed and then used to modulate biphasic pulses. A nonlinear compression function (e.g., logarithmic) is used to ensure that the envelope outputs fit the patient's dynamic range of electrically evoked hearing. Trains of balanced biphasic pulses, with amplitudes proportional to the envelopes, are delivered to the electrodes at a constant rate in a non-overlapping fashion.

3.2. Filter bank realization

The filter bank is the most complex component and usually takes more than 60% execution time of the CIS algorithm. Fig. 4 shows the prototype structure of a frequency-domain band-pass filter bank. Firstly, an N-Point FFT transforms the time-domain signals to frequency domain. Windows are then used to extract desired sub-band signals and frequency shifters deliver each sub-band signal to base band. These base-band signals are then transformed to time domain by N-Point IFFT. Finally decimation filters with a decimation factor of M are applied since the bandwidth of each sub-

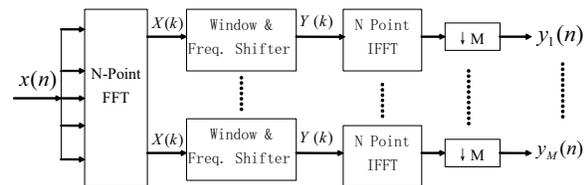


Fig. 4. Primordial structure of filter bank

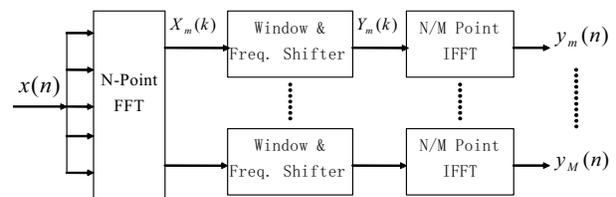


Fig. 5. Block diagram of simplified filter bank

band signal has been decreased to $1/M$ of the original signal in the previous process of frequency division. This filter bank can realize any shape of frequency responds with less computational complexity than IIR or FIR filters.

In fact, the IFFT and the decimation filters can be merged to further decrease the computational complexity. As we have proven in [3], an N/M Point IFFT can be used instead and the block diagram of the simplified realization of the filter bank is shown in Fig. 5.

3.3. Implementation results

In the proposed system, a 10-bit A/D converter is used to sample the sound signals at a rate of $f_s = 10\text{KHz}$, the point of FFT is $N = 1024$, and the number of sub-band is $M = 16$. The algorithm is implemented using ADSP-2189 platform [8]. Table 1 summarizes the implementation results. The CIS algorithm is realized with only 2.8 MIPS which can be mainly attributed to the simplified filter bank. It reduces the computational complexity by more than 80% when compared with traditional IIR and FIR filter banks.

Table 1. Implementation results of filter banks and CIS algorithm

| Type | Parameter | Complexity |
|---------------|--|------------|
| IIR-FB | $N_{order} = 8$ | 6.7 MIPS |
| FIR-FB | $N_{order} = 128$ | 10.5 MIPS |
| Simplified-FB | 1/2 Overlap $C_{FFT(64)} = 1405$ $C_{FFT(1024)} = 37203$ | 1.3 MIPS |
| CIS | With simplified-FB | 2.8 MIPS |

The duration of the computing window used above is 100 ms ($N = 1024$), and the calculating delay for the data within each window is about 7.5 ms (0.3 mega instructions @ 40 MIPS using ADSP-2189). A narrower window can be adopted to improve the system delay, but will also degrade the frequency resolution.

4. POWER TRANSMISSION EFFICIENCY

To achieve high power-transmission efficiency, high data-transmission bandwidth, and coupling insensitivity are three major challenges in the design of an inductive transmission link, which have been fully discussed in [9-11]. In practice, the three aspects of performance trade off with each other, making the design a multi-dimensional optimization problem. Since the proposed system has enormously reduced the requirement of data-transmission bandwidth, the design of the transmission link becomes less difficult.

Fig. 6(a) shows the equivalent circuit of an inductive transmission link, in which L_1 and L_2 are two inductors with

coupling factor k . In the circuit, R_{L1} and R_{L2} are the serial resistors of L_1 and L_2 respectively, i.e., $R_{L1} = \omega L_1 / Q_1$, $R_{L2} = \omega L_2 / Q_2$, where Q_1 and Q_2 are corresponding quality factors of L_1 and L_2 . For convenience, some variables are defined as follows

$$Z = R_{L2} + \frac{1}{j\omega C_2 + \frac{1}{R_L}} = R_{L2} + \frac{R_L}{1 + j\omega R_L C_2} \quad (1)$$

$$Z_{L1} = j\omega L_1, Z_{L2} = j\omega L_2, Z_M = j\omega k \sqrt{L_1 L_2}$$

Hence, the equations of the circuit in Fig. 6(a) can be expressed as

$$\begin{cases} Z_{L1} I_1 + Z_M I_2 = V_1 \\ Z_M I_1 + (Z_{L2} + Z) I_2 = 0 \end{cases} \quad (2)$$

The result of Eq. (2) is

$$\begin{cases} Z_m = Z_{L1} - \frac{Z_M^2}{Z_{L2} + Z} \\ V_2 = \frac{Z_M (Z_{L2} + Z)}{(Z_{L2} + Z) Z_{L1} - Z_M^2} V_1 \end{cases} \quad (3)$$

The expression of the power transmission efficiency η can be deduced from Eq. (3) along with

$$\begin{cases} V_L = \frac{Z - R_{L2}}{Z} V_2 \\ V_1 = \frac{I_0}{\frac{1}{Z_m + R_{L1}} + j\omega C_1 + \frac{1}{R_s}} \cdot \frac{1}{Z_m + R_{L1}} \\ \eta = \frac{|V_L|^2 / R_L}{\text{real}(I_s V_1)} \end{cases} \quad (4)$$

Finally, analytical result of η is

$$\eta = \frac{\left| \frac{Z_M (Z_{L2} + Z)}{(Z_{L2} + Z) Z_{L1} - Z_M^2} \cdot \frac{Z - R_{L2}}{Z} \right|^2 \left(\left(j\omega C_1 + \frac{1}{R_s} \right) (Z_m + R_{L1}) + 1 \right)^2}{\text{real} \left(\left(j\omega C_1 + \frac{1}{R_s} \right) (Z_m + R_{L1}) R_L \right) + R_L} \quad (5)$$

Optimization towards a higher η based on Eq. (5) has

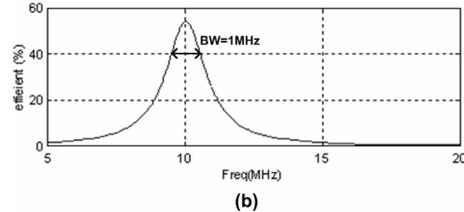
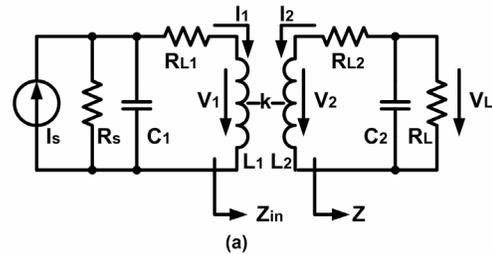


Fig. 6. (a) Equivalent circuit for frequency analysis, (b) Frequency response of transmission efficiency

been carried out by altering parameters of the components in Fig. 6(a), and the best result is shown in Fig. 6(b). As illustrated in the figure, bandwidth more than 1MHz can be provided for data transmission while guaranteeing the power transmission efficiency is above 40%. And the choice of 10 MHz as the carrier frequency is low enough to ensure low absorption of the skin and thus can avoid any long-term damage to the human body [4].

5. COMPARISON OF THE PROPOSED SYSTEM WITH THE TRADITIONAL SYSTEM

5.1. Additional power consumption

Simplified block diagrams in Fig. 7 depict the difference between the traditional and the proposed systems. To simplify the analysis, it is assumed that the power consumption of each component will not change while working in two different systems and the power transmission efficiency η is also identical.

The power consumption of the traditional and the proposed systems can be respectively expressed as

$$W_{old} = W_b + W_{ad} + W_{dsp} + \frac{1}{\eta}(W_{da} + W_a)$$

$$W_{new} = W_b + W_{ad} + \frac{1}{\eta}(W_{dsp} + W_{da} + W_a)$$

Thus, a difference can be derived as follows

$$W_{dif} = W_{new} - W_{old} = \frac{1-\eta}{\eta}W_{dsp}$$

The power consumption of the DSP can be estimated with the power indicator for general-purpose ADSP-2189 that is 0.55 mW/MIPS [12]. The power transmission efficiency η is obtained as 40% from Section 4. So compared with the traditional system, the increased power dissipation of the proposed system is

$$W_{dif} = (1-40\%) \times (0.55 \times 2.8) / 40\% = 2.31mW$$

It is less than 10% of the power consumption of the traditional system and can be further reduced with the proposed application-specific DSP.

5.2. Performance gain

In the proposed system, only voice-band signals with low data rate (100Kbps, 10-bit@10KHz) should be transmitted via the wireless link and the data rate doesn't necessarily increase with future rise of channel number or pulse rate. It eases the design of the antenna and favors the improvement of the power transmission efficiency. The speech processing is also free of the data rate, which increases the versatility of the processing strategy and the precision of the stimuli.

6. CONCLUSION

A cochlear system with implant DSP is presented in this paper. Only voice-band signals with very low data rate (100

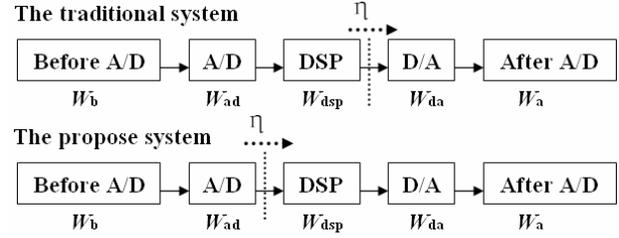


Fig. 7. Power comparison of two systems

Kbps) are required to be transmitted via the wireless link regardless of channel number and stimulating frequency, which makes this system suitable for future development. The CIS algorithm is optimized and implemented in 2.8 MIPS using a general-purpose DSP. The power transmission efficiency of the wireless link is promoted above 40%. By combining these two factors, the increased power consumption of the proposed system can be reduced to less than 10% when compared with the traditional system.

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