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Time Encoding and Reconstruction of Multichannel Data by Brain Implants Using Asynchronous Sigma Delta Modulators

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Abstract— **Recently, information technology and microelectronics have enabled implanting miniature and highly intelligent devices within the brain for in-vitro diagnostic and therapeutic functions. Power and physical size constraints of these devices necessitate novel signal processing methods. In this paper we investigate an effective data acquisition and reconstruction method for brain implants based on Asynchronous Sigma Delta Modulators (ASDMs). The ASDMs are analog non-linear feedback systems capable of time coding signals. The proposed reconstruction algorithm is based on the Prolate Spheroidal Wave Function (PSWF) expansion of the sinc functions and the order of expansion is given by the input signal being coded. Multiplexing and transmission of the different channels of data are accomplished by chirp orthogonal frequency division multiplexing. Computer simulations using multi channel electroencephalographic data are performed for wireless transmission by brain implants for monitoring abnormal brain activities of epilepsy patients.**

I. INTRODUCTION

Currently, medical implants are being developed for a variety of clinical applications. For instance, in the case of focal epilepsy prior to surgical intervention the localization of epileptogenic tissues is needed. As a dynamic system, the brain produces complex cellular activities which involve some latencies in time. Due to a kindling period before the full development of an epileptic seizure [1], the initial ictal activity appears to spread from the epileptogenic zone to the other regions of the brain during a transition period until the seizure is fully developed rather than starting instantaneously in all affected areas. Therefore, by detecting the time at which the ictal activity starts for each recording channel, the epileptogenic zone within the brain can be determined [2]. High performance brain implants are ideal tools for detecting early ictal activity for localizing epileptogenic zones accurately for subsequent surgical treatment. Another application maybe to use them to provide a warning signal to the patient when an incoming seizure is detected. These implants may also be used as treatment tools by delivering anti-epileptic drugs or stimulating the brain to counter-react to developing seizures.

Although brain implants are very useful in medical applications, the design and construction of these devices are technologically challenging. Among a number of unsolved problems, efficient energy management in brain computer interface (BCI) is an important problem. Power requirement and dissipation due to analog to digital conversion as well as to wireless transmission are major limitations in the implementation of human implants. Uniform sampling in the analog to digital converters (ADCs) requires synchronous implementation with a common clock shared with the digital signal processor. The need for a clock is a source of power consumption. In contrast, asynchronous circuits are not governed by a clock and consume low power. The elimination of clocks in these circuits also reduces device sizes and cuts electromagnetic interference (EMI) significantly. Due to these desirable properties, Asynchronous Sigma Delta Modulators (ASDMs) have been proposed for data acquisition in bio-monitoring systems [3].

An ASDM is a non-linear feedback system capable of coding an analog signal using time information [6]. Reconstruction of band-limited signal from the zero crossings of the binary output of an ASDM is possible since the amplitude information of the input signal is encoded in the pulse widths of the output signal. In [3], the authors show how to reconstruct the original signal from the output of the ASDM. In the present work we provide a computationally efficient reconstruction algorithm based on a Prolate Spheroidal Wave Functions (PSWF) projection of the original signal [7], [9]. We show that many signals such as the electroencephalogram (EEG), can be accurately represented by the PSWFs as an interpretation of Shannon's sampling theory using the ASDM time codes. In order to transmit the data collected from a number of signal channels, such as multiple EEG electrodes without spatial blurring, we investigated an efficient multiplexing method for wireless transmission using a modulation system known as Orthogonal Frequency Division Multiplexing (OFDM) [10]. In this paper we also assume volume conduction communication in the human body [4], [5] which, when compared to other methods, requires low power consumption.

II. ASYNCHRONOUS SIGMA DELTA MODULATORS

An Asynchronous Sigma Delta Modulator (ASDM), Fig. (1), is a nonlinear feedback system consisting of an integrator and a non-inverting Schmitt trigger [6]. In the ASDM, amplitude information of a signal $x(t)$ is transformed into time information without the quantization error that exists in the synchronous sigma delta modulators.

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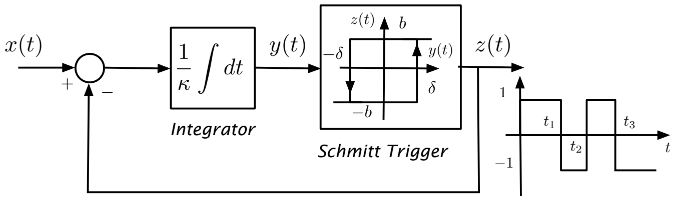

Fig. 1. Asynchronous sigma delta modulator

A. Time-encoding

Assume that the input $x(t)$ of the ASDM is band-limited, with a maximum frequency Ω_{max}, and bounded, $|x(t)| \leq c$. The output of the integrator, $y(t)$, at time $t_{k+1} > t_k$ is

$$y(t_{k+1}) = y(t_k) + \frac{1}{\kappa}\int_{t_k}^{t_{k+1}}[x(u) - z(u)]du.$$

Initially assuming that the Schmitt trigger is in the state $(-b, -\delta)$ at the instant $t = t_o$, ($y(t_o) = -\delta$ and $z(t_o) = -b$), the following equation holds:

$$\delta = -\delta + \frac{1}{\kappa}\int_{t_o}^{t}[x(u) + b]du \tag{1}$$

and since $y(t)$ increases monotonically, the trigger switches to the state (b, δ) at the time $t = t_1 > t_o$ the equation

$$-\delta = \delta + \frac{1}{\kappa}\int_{t_1}^{t}[x(u) - b]du \tag{2}$$

is satisfied for some $t = t_2 > t_1$. Combining (1) and (2), for the strictly increasing sequence t_k, $k \in Z$ the following equation

$$\int_{t_k}^{t_{k+1}} x(u)du = (-1)^k[-b(t_{k+1} - t_k) + 2\kappa\delta] \tag{3}$$

uniquely describes the relationship between $z(t)$ and $x(t)$ for all $t \in R$ and $|y| \leq \delta$ [3]. For future use, we call the right-hand side term of (3) as $v(k)$ such that

$$v(k) = (-1)^k[-b(t_{k+1} - t_k) + 2\kappa\delta].$$

The perfect reconstruction of $x(t)$ is possible provided that the sequence $\{t_k\}$ satisfies the condition [3] :

$$\max_k(t_{k+1} - t_k) \leq T_N \tag{4}$$

where $T_N = \pi/\Omega_{max}$ is the Nyquist sampling period. Since $x(t)$ is bounded, i.e., $|x(t)| \leq c$, equations (3) and (4) give

$$t_{k+1} - t_k \leq \frac{2\kappa\delta}{b + c} \leq T_N$$

providing a way to choose the parameters b, δ, and κ in terms of the Nyquist sampling rate.

B. Reconstruction

The reconstruction of the band-limited signal $x(t)$ from the zero-crossings of $z(t)$ requires a finite length approximation of the sinc function and an approximation of the integral in equation (3). The Shannon sinc-interpolation for the nonuniform times $\{t_k\}$ is

$$x(t) = \sum_{k=-\infty}^{\infty} \gamma_k S(t - t_k) \tag{5}$$

where γ_k are coefficients and $S(t)$ is the sinc function. In [3], the sinc function is approximated by complex exponentials

$$S(t) \approx \sum_{m=-L}^{L} \alpha e^{jm\Omega_0 t} = \alpha\frac{\sin((L + 0.5)\Omega_0 t)}{\sin(0.5\Omega_0 t)} \tag{6}$$

where L is an arbitrarily larger number not connected with the signal $x(t)$, and $\Omega_0 = \frac{\Omega_{max}}{L}$. In the following section we will show that using the Prolate Spheroidal Wave Functions (PSWFs) [7] the maximum frequency of $x(t)$ can be used to determine the representation of the sinc function.

III. Prolate Spheroidal Wave Functions for Sampling and Reconstruction

The Prolate Spheroidal Wave Functions (PSWFs) have maximum energy concentration within a given bandwidth among all time-limited signals. The PSWFs $\{\varphi_n(t)\}$ have been considered for the reconstruction of band-limited signals from uniform and nonuniform samples [8], [9]. The following are important properties of these functions:

- The PSWFs constitute an orthogonal basis for the space of finite energy signals with finite support $[-T, T]$, or signals in $\mathcal{L}^2(-T, T)$, and an orthonormal basis for the space of bandlimited functions [8].
- The sinc function $S(t)$, which belongs to the space of band-limited signals, can be expanded in terms of the basis $\{\varphi_n(t)\}$. The shifted sinc function can be expressed as

$$S(t - kT_s) = \sum_{n=0}^{\infty} \varphi_n(kT_s)\varphi_n(t)$$

- For signals with a finite support, Parseval's theorem and orthogonality on the interval $(-T, T)$ give

$$\int_{-T}^{T} |x(t)|^2 dt = \sum_{m=0}^{\infty} \lambda_m |\gamma_m|^2$$

where the eigenvalues λ_m are related to the energy of the signal and are ordered as $\lambda_0 \geq \lambda_1 \geq \lambda_2 \geq \ldots > 0$ and coefficients γ_m.

The Shannon's sinc interpolation for a bandlimited signal $x(t)$ as an expansion in terms of the PSWFs:

$$\begin{aligned} x(t) &= \sum_{k=-\infty}^{\infty} x(kT_s) \sum_{n=0}^{\infty} \varphi_n(kT_s)\varphi_n(t) \\ &= \sum_{n=0}^{\infty} \left[\sum_{k=-\infty}^{\infty} x(kT_s)\varphi_n(kT_s) \right] \varphi_n(t) \\ &= \sum_{n=0}^{\infty} \gamma_n \varphi_n(t). \end{aligned} \tag{7}$$

Considering a time-limited signal $x(t)$, $0 < t \leq T$, and assuming the signal energy outside the given frequency band $(-\Omega_{max}, \Omega_{max})$, is small enough to be ignored, an approximate of $x(t)$ is given by the PSWF projection

$$\begin{aligned} \hat{x}(t) &= \sum_{n=0}^{M-1} \left[\sum_{k=0}^{N-1} x(kT_s)\varphi_n(kT_s) \right] \varphi_n(t) \\ &= \sum_{n=0}^{M-1} \gamma_{M,n}\varphi_n(t) \end{aligned} \tag{8}$$

where $T_s \leq \pi/\Omega_{max}$ and M is the number of PSWFs that gives a good approximation for the signal depending on Ω_{max}. This indicates that depending on Ω_{max}, we can find a PSWF representation of the sinc function of order M as shown in [9].

IV. ASDM Reconstruction Algorithm

Assume that the input to the ASDM is approximated by a PSWF projection

$$x(t) \approx \hat{x}(t) = \sum_{n=0}^{M-1} \gamma_{M,n}\varphi_n(t)$$

where the value of M is chosen by making the frequency of the $\gamma_{M,M}$ coincides with the frequency of $x(t)$. If we let $t = k\Delta_t$, such that $\Delta_t < \pi/\Omega_{max}$ the projection can be written

$$\hat{\mathbf{x}} = \mathbf{\Phi}\ \gamma_{\mathbf{M}} \tag{9}$$

The integral in (3), for $b = 1$, can be approximated by means of the trapezoidal rule as

$$\begin{aligned} \int_{t_k}^{t_{k+1}} x(t)dt \quad &\approx \quad 0.5x(t_k)\Delta_t + \sum_{i=1}^{N_k-1} x(t_k + i\Delta_t)\Delta_t \\ &\quad +0.5x(t_{k+1})\Delta_t \end{aligned}$$

where $N_k = (t_{k+1} - t_k)/\Delta_t$. Letting $v(k)$ in Eq. (3) be the entries of a vector $\mathbf{v_k}$ computed at each of the $\{t_k\}$ values we have that

$$\mathbf{v_k} \approx \mathbf{q_k}\ \mathbf{\Phi}\ \gamma_{\mathbf{M}}$$

where the entries of the row vector $\mathbf{q_k}$ are given as

$$q_{k,j} = \begin{cases} 0.5\Delta_t & j = N_k, \text{ and } j = N_{k+1} \\ \Delta_t & N_k + 1 \leq j \leq N_{k+1} - 1 \\ 0 & \text{otherwise} \end{cases}$$

Thus for the time sequence $\{t_k,\ k = 1, \cdots, K\}$ equation (3) can be written as

$$\mathbf{v} = \mathbf{Q}\ \mathbf{\Phi}\ \gamma_{\mathbf{M}} \tag{10}$$

where $\mathbf{Q}$ is the matrix composed of the vector $\mathbf{q_k}$ and $\mathbf{v}$ is composed of the term in the right-hand of (3). Computing

$$\gamma_{\mathbf{M}} = [\mathbf{Q}\ \mathbf{\Phi}]^{\dagger}\mathbf{v} \tag{11}$$

we can use it to find the projection $\hat{\mathbf{x}}$ in (9) where $\dagger$ represents the pseudoinverse operation. To see an example of comparison between the same order PSWF and sinc function (with exponential approximation) reconstruction, see Fig.6 for an arbitrary bandlimited signal.

V. Modulation/Multiplexing for ASDM Output

We take advantage of the Orthogonal Frequency Division Multiplexing (OFDM) for a bandwidth efficient transmission of multichannel brain data. OFDM is a multicarrier modulation method of which subcarriers overlap in the frequency domain. The mutual orthogonality of the subcarriers in the frequency domain ensures the recovery of data at the receiver for each substream. It is possible to use a chirp basis that is guaranteed to be orthogonal by the Fractional Fourier Transform (FrFT) method [10] where instantaneous frequency $IF(t)$ of the chirp is:

$$IF_u(t) = u/T - \cot\alpha t \tag{12}$$

for channels $u = 0, 1, \ldots, U-1$ and signal duration T.

To reduce the bandwidth, the output $z_u(t)$ of each of the channels requires pulse shaping. This can be achieved by approximating the derivative of $z_u(t)$ by $w_u(t) = 0.5[z_u(t) - z_u(t - \Delta_t)]$. Using the PSWF as the pulse shaping filter $h(t)$, the output of the filter $s_u(t) = w_u(t) * h(t)$ which is then modulated by the chirp $c_u(t)$. The modulated signal for the U channels is

$$p(t) = \sum_{u=0}^{U-1} s_u(t) \cdot c_u(t) \tag{13}$$

At the receiver, the time codes corresponding to each channel are obtained by multiplying by the conjugate of the corresponding chirp and then low pass filtering. The corresponding demodulated signal $s_{d,u}(t)$ for each channel u is

$$s_{d,u}(t) = p(t) \cdot c_u^*(t). \tag{14}$$

The time codes can be found by finding the peaks of the demodulated waveforms.

VI. SIMULATIONS

In our simulations we used subdural EEG signals of an epilepsy patient. For 4-channels, ASDM sampling was applied giving time codes for each $z_u(t)$, $u = 1, \ldots, 4$. The parameters of the ASDM were chosen for a sampling period of 5 msec. After pulse shaping (see Fig.2) and modulation/multiplexing (see Fig.3), the total signal $p(t)$, as a collection of modulated signals $s_u(t)$, is to be transmitted through a short range communication channel to a PDA mounted on the human body. Each demodulated waveform is shown in Fig. 4. Then a more power consuming digital RF communication system can be used for long distance communications. In a remote clinical environment the signals corresponding to different channels will be reconstructed. Fig. 5 shows the reconstructed 0.14 seconds long subdural EEG for one of the channels using the proposed method.

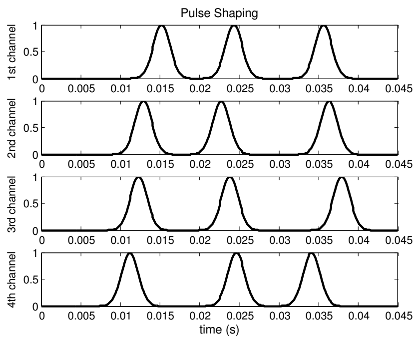

Fig. 2. Pulse-shaped time codes for 4 channels using PSWF pulses.

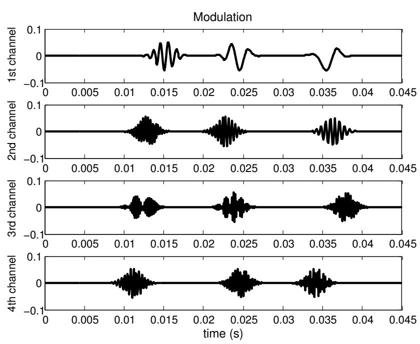

Fig. 3. Chirp-modulated signals from 4 channels.

VII. CONCLUSIONS

We have presented an efficient data acquisition and reconstruction method for biomedical implants, especially for the monitoring of the brain activity of epilepsy patients. Our method is energy efficient and sufficiently accurate, providing a useful tool for detecting the onsets of epilepsy seizure events. The reconstruction is based on a PSWF representation of the sinc function with order given by the maximum frequency of the signal being sampled, rather an using for it an exponential expansion of arbitrary order. The modulation is implemented using chirp OFDM with the bandwidth of each channel reduced by PSWF pulse shaping.

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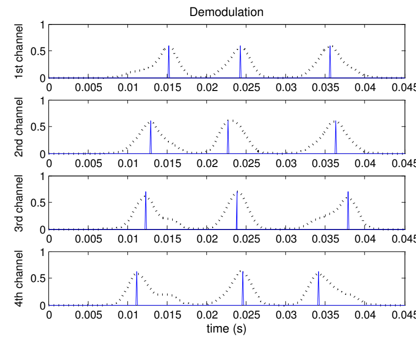

Fig. 4. Demodulation

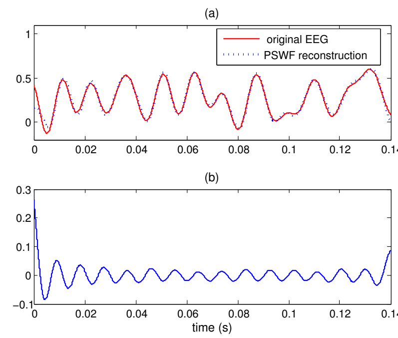

Fig. 5. (a) reconstructed vs. original EEG, (b) reconstruction error

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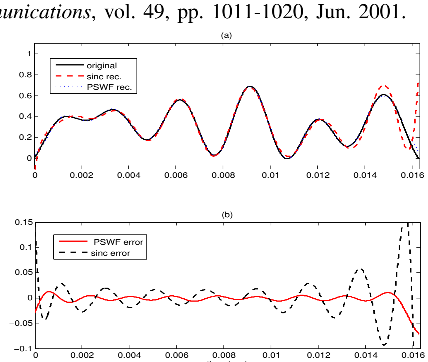

Fig. 6. (a) reconstructed signals vs. original signal, (b) reconstruction errors