Real-Time Digital Signal Processing of Electroneurographic Signals

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ABSTRACT

A battery powered, isolated, Digital Signal Processor (DSP) based system was developed to extract real-time information from electroneurographic (ENG) signals received from a nerve-cuff electrode implanted around the *sural* nerve of a human subject. The processed ENG signal, acting as an indication of foot contact, has been used to control a Functional Electrical Stimulator (FES) system, in an attempt to correct "drop-foot" gait in a hemiplegically paralyzed, spastic patient. A significant improvement in signal-to-noise ratio over a previous system that was based solely on analog electronics has been achieved through the use of digitally implemented Bessel filters, rectification, and integration.

INTRODUCTION

With the advent of low-power, high performance Digital Signal Processors (DSPs), it has become feasible to implement many of the processing functions typically performed on bio-potential signals in the digital domain. Also, functions that have typically been performed "off-line", can now be implemented with real-time digital systems. The question remains: Is the increased complexity of such DSP based systems, warranted by an adequate increase in the information content derived from the processed signals, and the additional flexibility provided? To answer this question, we have quantified and compared the effectiveness of an analog neural prosthetic system with a DSP based system, in analyzing electroneurographic (ENG) signals.

METHODS

In order to detect foot contact, ENG signals were obtained from the *sural* nerve of a hemiplegic spastic subject using an implanted silicone tube/stainless steel wire nerve-cuff electrode. The cuff was connected via transcutaneous leads to a custom designed, isolated amplifier [1] with a gain of 140,000. The resulting signal was input into a battery powered, analog ENG processing/FES stimulator system [2], and simultaneously to a digitizing, 8-channel DAT recorder, with a 22 kHz sampling rate and 14 bit resolution (TEAC model RD-135T). The analog processing system filtered, rectified, and bin-integrated (RBI) the amplified ENG signal, using a synchronizing signal from the muscle stimulator (also recorded) to blank the input signal for a constant time (up to 25mS) after the stimulation pulse. The stimulation pulse width was fixed at 330µS, with a 30 Hz repetition rate. Stimulation current was held constant throughout the trial, at a level determined by the patient to be adequate for gait restoration. The RBI output from the analog system, used as an indicator of heel-contact (to turn off the stimulator), was recorded on another DAT channel. Although the DSP based system was designed to replace the analog system, the objective of this trial was the direct comparison of the two systems using the **same** raw ENG signal. Therefore, the amplified ENG signal, recorded to DAT during the gait tests, was played back into the DSP system through a TMS32046 A/D-D/A converter chip, filtered (using the chip's built-in eighth-order elliptic antialiasing filter), and digitized to 14 bit resolution with a 20 kHz sample rate. The digitized signal was then processed by an Analog Devices ADSP-2101 16 bit, fixed-point DSP running at 16 MHz. The recorded stimulator sync. pulse was used to generate interrupts, allowing blanking (similar to that implemented in hardware in the analog system) to be performed as part of the DSP algorithm. The algorithm consisted of two sequential 8th order Bessel filters: a low-pass filter with a cutoff (-3dB) frequency of 4 kHz, followed by a high-pass filter with a cutoff frequency of 1.5 kHz. The resulting filtered signal was rectified, and bin-integrated in double precision (32 bits), using a bin time corresponding to the interval between stimulator sync. pulses. The filter input was zero-padded for 3.5mS after each sync. pulse, blanking the ENG signal analogously to the blanking performed in the analog system, but for a shorter time. The rectified, and bin-integrated signal was then output to the D/A converter portion of the TMS32046, again at a 20 kHz conversion rate. This real-time analog output from the DSP system was then digitized, along with the RBI output from the analog system using a standard PC based data sampling system (with 12 bit accuracy, no input antialiasing filters, 200 Hz sample rate). A comparison of the RBI signals from the digital and analog systems was performed in Matlab, using this digitized data.

Of principle significance in the digital implementation, are considerations of filter order, type, topology, and precision. Although they exhibit the least amount of attenuation for a given order, Bessel filters were chosen due to their favorable impulse response characteristics. Bessel filters have maximally flat group delay characteristics, and, consequentially, little ringing in their impulse and step responses. The digital realization was achieved using a **normalized** analog Bessel prototype [3]. The equation yielding *n*th order prototype Bessel filter coefficients is:

$$H(s) = \frac{b_0}{q_n(s)}, \text{ where } q_n(s) = \sum_{k=1}^{n} b_k s^k, \text{ and } b_k = \frac{(2n-k)!}{2^{n-k}\, k!\,(n-k)!}$$

This prototype filter is specified to give unit delay at ω=0, with the result that the -3dB point varies with filter order. It was desired that the cutoff frequencies be precisely specified, independently of filter order. Therefore, the Bessel coefficients were normalized (for -3dB at ω=1) using:

$$b_k' = A^{n-k} b_k, \text{ where } A = 0.31546 \text{ for } 8\text{th order filters}$$

The resulting transfer function for the prototype low-pass normalized Bessel filter was thus:

$$H(s) = \frac{199}{s^8 + 11.4s^7 + 62.7s^6 + 218s^5 + 515s^4 + 844s^3 + 932s^2 + 630s + 199}$$

The high-pass filter analog transfer function was generated by using the standard low-pass to high-pass transformation:

$$H(s) = H_0(\omega_c^2 / s), \text{ where } H_0(s) \text{ is the lowpass response}$$

Due to the limited (16 bit) precision of the DSP used, these filters were realized as a cascade of 2nd order Direct Form II (Biquad) sections, resulting in the transfer function:

$$H(z) = \prod_{n=1}^{4} \frac{B_{0n} + B_{1n} z^{-1} + B_{2n} z^{-2}}{1 - A_{1n} z^{-1} - A_{2n} z^{-2}}$$

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The digital (z) domain transfer functions were derived from the analog transfer functions using a pole-only version of the factored form Bilinear Transform. The analog transfer functions were factored (using Matlab root finding routines) into complex conjugate pairs, and the z domain denominator factors found from the analog prototype poles (p_n) using:

$$Den(z) = (z - z_{p1})(z - z_{p2}), \text{ where } z_{pn} = \frac{2 + p_n T}{2 - p_n T}, n = 1,2$$

Since the Bessel analog transfer functions contain multiple zeros at s=∞ (low pass case) or s=-∞ (high pass case), the digital zeros were fixed at z=±1, resulting in numerator (B_n) coefficients of 1,2,1 for low-pass filters, and 1,-2,1 for high-pass filters. The overall filter gain was divided equally amongst the sections and factored into the numerator coefficients. The resulting coefficients were then expressed in the DSP as Q14 format numbers (a sign bit, 1 integer and 14 fractional bits) to accommodate a range of $-2<n<+2$.

RESULTS

In order to quantify the effectiveness of the two systems, a measure of signal-to-noise ratio was developed, where the 0dB, "noise" reference point was defined to be the **peak** value of the RBI ENG signal during the swing phase of the subject's gait, and "signal" was defined to be the **peak** value of the RBI ENG signal during the stance phase. Both systems detected heel-contact using a simple comparator. A fixed threshold level was set as a reference, and any RBI ENG peak value over the threshold was seen as an indicator of heel-contact. A single noise peak over the threshold level occurring during swing phase would be perceived, erroneously, as a heel-contact. Conversely, if the threshold level was set too high, valid heel-contact signals were missed. Thus it was desirable that the ratio of the peak RBI ENG signal, occurring with heel-contact at the start of stance phase, to the peak noise signal during swing phase, be as large as possible. This ratio, expressed in dB, was calculated for both the analog and digital systems, for each individual gait cycle, using gait data from 125 consecutive cycles. The result is shown as a histogram in Figure 1. The analog system's average SNR was 8.9dB, with 85% of the data falling between 9 and 15dB. The digital system's average SNR was 17.3dB, with 85% of the data falling **above** 15dB. The digital system's SNR was better than the analog system's in 123 of the 125 cycles analyzed.

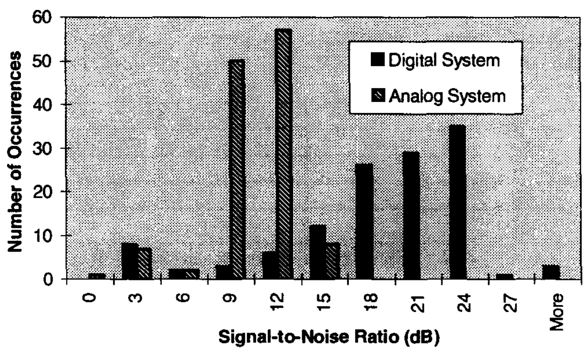

Fig. 1 - Performance of the Digital vs. Analog System

DISCUSSION

The most significant problem in correlating ENG signal activity with the applied mechanical stimulus, is the elimination of noise. This noise appears in a variety of forms, dependent upon where in the gait cycle measurements are made. During swing phase, while the foot dorsiflexors (primarily *Tibialis anterior*) are being electrically stimulated, noise results from two sources: 1) A large artifact spike, due to the limited ability of the nerve-cuff to reject radial potential gradients [4], appears nearly simultaneous (delayed by the various tissue conduction velocities between the stimulation electrode site, and the nerve-cuff) with the externally applied stimulation pulse (up to 400µS in width, 170V in amplitude). 2) In response to the induced muscle stimulation, a significant (>10µV) electromyographic (EMG) signal is present. Since the desired ENG signal is in the range of 1µV, these noise signals result in a SNR of less than **minus** 20dB! The artifact spike acts essentially as a Dirac impulse, initiating the impulse response of any filter it is passed through. Therefore, processing must be halted (blanked) during the artifact. The EMG response signal generally contains lower frequency components (up to 2 kHz) than the desired ENG components (800-3000 Hz), and can thus be substantially filtered out, while still leaving significant ENG information. The analog system, with its lower order filters, was unable to adequately eliminate this EMG noise. Thus a long blanking interval was required to allow residual EMG to die out. This meant that only a very short sample (3-4mS) of ENG data was available for analysis during each stimulus period. With its substantially better filters, the DSP system was able to eliminate much more EMG interference, and thus operate with a reduced blanking interval, allowing more ENG data to be integrated per stimulus frame. Trials with the analog system indeed showed that increasing filter order resulted in improved SNR. Also significant, was the ability of the DSP system to integrate signals over a fixed noise threshold. It was also found that integration of the squared RBI signal (proportional to signal power) further increased SNR.

CONCLUSIONS

We have shown that a significant increase in the information content obtained from ENG signals can be realized using digital signal processing. Although, much of this increase can be attributed to improved filtering, which could also be performed in the analog domain, it is anticipated that further improvements will be achieved through the use of more sophisticated algorithms. Such algorithms (higher order filters, frequency domain analysis, etc.) can not be implemented with analog circuitry. Given continual reductions in cost and power consumption, coupled with ever improving integration and performance in successive generations of digital signal processors, it seems likely that they will be used more frequently as a viable and competative alternative to standard analog electronics in battery-powered bio-potential signal processing systems.

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ACKNOWLEDGEMENTS

We would like to thank Dr. Morten Haugland for the use of his ENG system, also, The Danish Foundation for Physical Disability, The Danish National Research Foundation, and The Danish Ministry of Industry for their financial support.