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Alternative pulse shapes in electrical hearing

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ABSTRACT

Cochlear implants (CIs) stimulate the auditory nerve with trains of symmetric biphasic (BI) pulses. We review studies showing that more efficient stimulation can be achieved by modifying these pulses by (1) increasing the inter-phase gap (IPG) between the two phases of each pulse, thereby delaying the recovery of charge, (2) increasing the duration and decreasing the amplitude of one phase – so-called "pseudomonophasic (PS)" waveforms, and (3) combining the pseudomonophasic stimulus with an IPG in a "delayed pseudomonophasic" waveform (PS_IPG). These efficiency gains, measured using changes in threshold and loudness, occur at a wide range of pulse rates, including those commonly used in current CI systems. In monopolar mode, dynamic ranges are larger for PS and for long-IPG pulse shapes than for BI pulses, but this increase in DR is not accompanied by a higher number of discriminable loudness steps, and hence, in a better coding of loudness. Moreover, waveforms with relatively short and long interphase gaps do not yield different patterns of excitation despite the relatively large differences in threshold.
Two important findings are that, contrary to data obtained in animal experiments, anodic currents are more effective than cathodic stimulation for human CI patients and that the thresholds decrease with increases in IPG over a much longer time course (more than 3 ms) than for animals. In this review it is discussed how these alternative pulse shapes may be beneficial in terms of reducing power consumption and channel interactions, which issues remain to be addressed, and how models contribute to guiding our research.

1. Introduction

Cochlear implants (CIs) make use of trains of symmetric biphasic (BI) electrical pulses to provoke action potentials in the auditory nerve (AN). These pulses consist of a first phase of one polarity immediately or shortly followed by a second phase of opposite polarity (Fig. 1a and b). Because CI stimulation is delivered extracellularly, every nerve fiber that is locally depolarized is also hyperpolarized at another or several other sites, and vice-versa. As a result, both phases (i.e. both polarities of a BI pulse) can depolarize the nerve fibers and potentially initiate action potentials (van den Honert and Stypulkowski, 1987; Miller et al., 1998, 1999). However, because the two polarity phases have counteracting effects on the polarization state of the neural membrane, BI pulses may not be the most efficient way of stimulating neurons (van den Honert and Mortimer, 1979; Miller et al., 2001; Rubinstein et al., 2001). The way two consecutive opposite polarity phases counteract each other's effects may be two-fold. First, consider a site on the fiber that is hyperpolarized by the first phase of a BI pulse. In order to generate an action potential, the second phase will first have to repolarize the membrane to its resting potential and then further depolarize it to its excitation threshold. Second, consider a site now *depolarized* by the first stimulus phase. At this specific site, the sodium channels start to open, allowing positive charges to flow inside the neural membrane. If the current amplitude is large enough, this triggers a regenerative process during which more and more sodium channels activate and further depolarize the neural membrane, eventually leading to the generation of an action potential. If a second, hyperpolarizing phase is sent during this regenerative process, this may cause a rapid repolarization of the membrane towards its resting state, thereby abolishing the generation of the action potential. This abolition phenomenon was extensively described by van den Honert and Mortimer (1979). They showed that the hyperpolarizing pulse amplitude

Abbreviations: CI, cochlear implant; IPG, interphase gap; AN, auditory nerve; C level, comfortable level; BI, biphasic; PS, pseudomonophasic; ALT-M, alternating-monophasic; PS_IPG, pseudomonophasic with an interphase gap; BP, bipolar; BP+1, bipolar with 1 intervening electrode; pps, pulses per second; DR, dynamic range; IDL, intensity difference limen; ECAP, electrically evoked compound action potential

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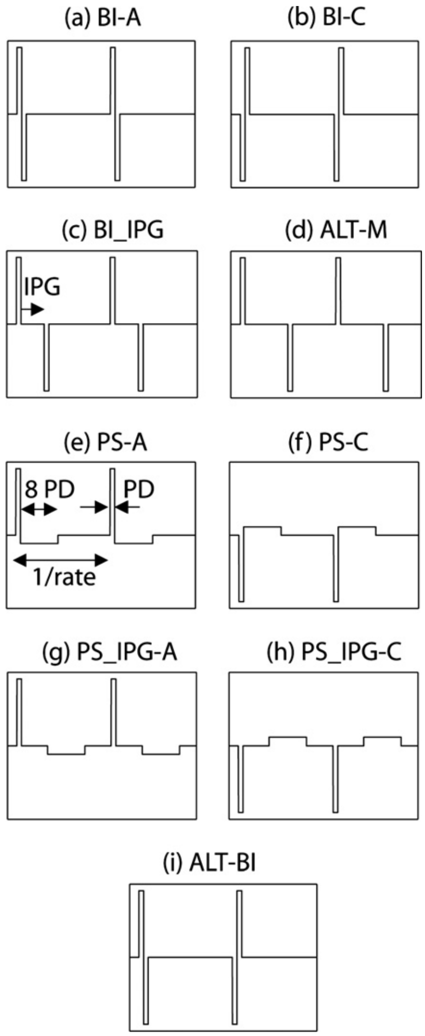

Fig. 1. Overview of the pulse shapes used in the study. (a) Symmetric biphasic anodic-first (BI-A), (b) symmetric biphasic cathodic-first (BI-C), (c) symmetric biphasic with an inter-phase gap (BI_IPG), (d) Alternating-monophasic (ALT-M; equivalent to a BI_IPG with an IPG such that the negative phase is midway between two consecutive positive phases), (e) pseudomonophasic anodic-first (PS-A), (f) pseudomonophasic cathodic-first (PS-C), (g) delayed pseudomonophasic anodic-first (PS_IPG-A; also with an IPG such that the negative phase is midway between two consecutive positive phases), (h) delayed pseudomonophasic cathodic-first (PS_IPG-C), and (i) alternating biphasic (ALT-BI) pulses. All asymmetric pulses had a long/low phase eight times longer (and eight times lower in amplitude) than the short/high one (PD = pulse duration).

required to prevent action potential generation was a function of both the amplitude of the depolarizing phase and of the gap between the two phases. They also reported that this abolition phenomenon could be eliminated by introducing a 100-μs gap between the two phases of a BI pulse. Although this behavior arises at least partly from the non-linear properties of the neurons' ionic currents, Rubinstein et al. (2001) showed that it could be modeled linearly as a leaky integrator of charge with a variable time delay.

Monophasic pulses (which consist of a single-polarity pulse) have been shown to be more efficient than BI pulses in eliciting neural responses. AN single-cell recordings in animals showed that thresholds could be reduced by up to about 4 dB by using monophasic rather than BI pulses (Shepherd and Javel, 1999). However, monophasic current pulses are not charge-balanced and cannot, therefore, be used with human subjects for safety reasons (Shepherd et al., 1999). Safety studies on neural electrical stimulation showed that the transfer of charge between the electrode and the electrolyte should be made essentially via reversible reactions, i.e. chemical reactions for which the products remain bound to the electrode and can be reversed to their initial state by an opposite current (Brummer and Turner, 1977; Robblee and Rose, 1990). In monophasic stimulation, charge accumulates up to a point where irreversible reactions may occur, releasing products potentially harmful for the tissue. The purpose of this paper is to review several pulse shapes which may be as efficient as monophasic stimulation while still being charge-balanced. We will concentrate on the effects of delaying each pulse's charge recovery by using (1) BI pulses with an interphase gap (BI_IPG, ALT-M; Fig. 1c and d), (2) "pseudomonophasic" (PS; Fig. 1e and f) pulses, where the charge of the second phase is re-distributed by a reduction in amplitude and an increase in duration (Miller et al., 2001; Rubinstein, 1993), and (3) pseudomonophasic pulses with an IPG (PS_IPG; Fig. 1g and h).

We will review in Section 2 the effects of these pulse shapes on thresholds, C levels and intensity discrimination of CI listeners and the way in which they are influenced by different stimulus parameters, such as phase duration, stimulation rate, phase polarity, or electrode configuration (McKay and Henshall, 2003; Carlyon et al., 2005; van Wieringen et al., 2005a, 2006; Macherey et al., 2006, 2007, 2008). In Section 3, we will describe the relative phase polarity contributions to neural excitation and show how they may affect the spatial selectivity of CI stimulation. Finally, in Section 4, the underlying physiological mechanisms, the implications for future speech processing strategies and some new perspectives of research will be discussed.

2. Effects of pulse shape on thresholds, C levels and intensity discrimination

2.1. Waveforms with different IPGs

Animal physiological studies have shown the threshold for a biphasic pulse to approach that of a monophasic pulse with increases in IPG (Shepherd and Javel, 1999). The magnitude of the threshold reduction depends on the phase duration and is larger for short durations. Specifically, for a 100-μs/phase BI pulse, the introduction of an 80-μs IPG decreased the threshold by 0.9 dB, whereas a 60-μs IPG produced a larger, 2.4-dB threshold drop (Shepherd and Javel, 1999). Similar results were obtained in behavioral experiments with implanted persons, showing an average difference of 2.5 dB between the "no IPG" and the "100-μs IPG" condition and a larger effect of IPG for a 26-μs than for a 52-μs phase duration (McKay and Henshall, 2003). Carlyon et al. (2005) extended the range of IPGs tested in CI users up to several milliseconds and showed that thresholds for biphasic pulse trains continued to decrease as the IPG increased up to at least 2.9 ms. Fig. 2 illustrates thresholds averaged over four Nucleus CI24 users (filled symbols) and standard errors of biphasic pulse trains of which the IPG varied between 8 μs and 2900 μs in monopolar mode. Thresholds are normalized to the (BI-A) condition with the 8 μs IPG. The solid and dashed lines plotted without symbols show the predictions of two phenomenological models that will be described in Section 4 (Carlyon et al., 2005; Macherey et al., 2007). The drop in thresholds with increases in IPG up to 2900–4900 μs appears, at first sight, inconsistent with the results of a previous study by McKay and Henshall (2003). They reported effects of IPG on threshold that appeared to asymptote at 100 μs, which was the longest gap studied by them. However, Carlyon et al. (2005) argued that the difference between the two sets of data was due to the difference in pulse rate, which was 1000-pps in the McKay