Current Distribution Measurements Within the Human Cochlea

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Abstract—The magnitudes of the currents returned through each ground electrode line of a multiple-electrode cochlear implant array were determined during surgical implantations on two patients. These were often found to be distributed widely to points far from the stimulus electrode site. Further measurements made in *in vitro* solutions

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demonstrated that the distributions were due largely to the ground electrode interface impedances being significantly larger than the fluid-path impedances, and demonstrated that distributions could be changed by modification of the ground electrode interface impedances.

INTRODUCTION

Electrical stimulation of the auditory nerve is now being actively investigated as a means of treating profoundly or totally deaf patients in whom there are residual auditory nerve fiber populations. Furthermore, the auditory nerve is extremely homogeneous in terms of its single fiber discharge characteristics, in contradistinction to higher auditory centers, and also exhibits a very simple tonotopic organization around the cochlear spiral. Therefore, excitation of the auditory nerve from an electrode array placed within the cochlea presents the possibility of exciting fiber populations of different frequency bands, but which are otherwise functionally homogeneous [1]. Implants incorporating such arrays have provided significant speech intelligibility scores for two implanted patients [2], [3].

This paper addresses the issue of localizing the stimulus currents to individual electrode sites as this is of primary importance to the successful functioning of the multiple-channel cochlear prosthesis. If excitation is truly tonotopic, with unique fiber populations excited by each active electrode, a much greater transfer of information to the auditory nerve would be possible than with single-channel stimulation.

The electrode array used in the investigation reported here is a modified version of the one described elsewhere [4], and is shown in Fig. 1. It consists of ten active electrodes interleaved with ten ground electrodes that are shorted together to form a common ground. Physiological studies have shown that arrays incorporating intracochlear ground electrodes result in current distributions that are significantly more localized to the active electrode than occurs for extra-cochlear ground electrodes [5], [1]. The current distributions for an interleaved ground system used in the present array, however, have not previously been investigated.

Measurements were made with this array during cochlear implant operations on two patients. The ground current distributions along the length of the array were determined by measuring the magnitude of the current in each ground electrode lead wire. Subsequently, human cochlear electrical impedances were studied using saline-filled tubes to provide an *in vitro* model.

METHODS

In carrying out cochlear implant operations [6], our procedure has involved inserting a dummy electrode array along the scala tympani to help select an implantable device with an array of the correct length. In the present study, the dummy array was used to measure ground current distributions in the cochleas of two patients, and as the investigations were carried out in 10–15 min, they did not significantly disadvantage them.

The dummy array used for the present measurements is shown in Fig. 1, and was similar to the normal array except that a number of active electrodes were removed. It consisted of two sections: a uniform silicon rubber tube of outside diameter 0.64 mm, and a tapered section at the tip which varied from 0.64 to 0.35 mm over a distance of 5.25 mm. Three active electrodes (A1, A2, and A3) and five common electrodes (C1–C5) were situated on the uniform section; A4 and C6–C9 were situated on the taper. Each electrode consisted of a band of pure platinum foil (99.5 percent Pt, Goodfellow's Metals) 0.3 mm wide and 0.0075 mm thick, tightly welded around the tube. A single Teflon-coated wire (Pt–10 percent Ir) of outside diameter 0.025 mm was welded to each electrode and passed back through the tube to the connector.

The magnitude of the ground currents in each ground electrode line was determined by switching a current sensor ($Z_{in} < 1 \Omega$) into each line. A current source with amplitude 0.5 mA, 0.2 ms pulse width, and repetition rate 100 pulses/s stimulated the active electrodes.

RESULTS

Results for the two patients were similar. Ground currents for one patient when stimulating electrodes A1, A3, and A4 are shown in Fig. 2.

The main features to be noted from these plots and from other measurements are the following.

1) There was considerable propagation of the ground currents along the length of the array which, in most cases, did not decrease monotonically with distance from the stimulus site.

2) The propagation was greatest in the basal direction (i.e., in the direction of C1). This was most clearly seen when stimulating the apical active electrode A4 where the ground current in C8, which was a common electrode adjacent to A4, was approximately the same as the ground current in C1, the basal-most ground electrode. Conversely, stimulating the basal electrode A1 resulted in ground currents which were relatively well localized to the stimulus site.

3) Measurements made for two positions of the electrode in the cochlea, viz. insertion depths of 18.5 and 14.5 mm from the round window, yielded similar current distributions.

4) Ground currents summed linearly for coincident stimulation of two electrodes.

It is, of course, not possible to deduce the total current distribution from the results above. The total current distribution must be considered to arise from the total source current at the active electrode, which propagates through the cochlear fluids to be returned to the ground electrodes. The results of Fig. 2 contain only the discrete values of the total sink currents returned to each ground electrode. Although the ground currents were distributed in some cases almost uniformly along the length of the array, discrete neural excitation was nevertheless presumably possible since psychoacoustic tests performed with this patient (and with one other patient on which electrode characteristics were not measured) revealed that a unique spectral color percept was achieved when stimulating each electrode [2], [3]. This percept was generally graded along the length of the array in a manner related to the normal tonotopic organization of the cochlea. Since the total current was sourced at the active electrode, the current density in the tissues would have been largest adjacent to the active electrode, and the ground currents would presumably have been mainly subthreshold. However, in some cases it is possible that ground currents may become excitatory. The stimulus thresholds in these two patients were in the range 200–500 μ A. Because the electrode array was somewhat smaller than the scala tympani, the greater part of this range was most likely due to different distances between the active electrodes and the terminal nerve fibers. It is therefore conceivable that ground electrodes adjacent to active electrodes which exhibit low stimulus thresholds will lie close to the nerve fibers. Therefore, if stimulation from other active electrodes with higher thresholds results in large ground currents being induced in such ground electrodes, they may conceivably become excitatory. In addition, the ground currents may become suprathreshold for coincident stimulation from more than one active electrode since they will sum linearly, in which case they will result in rapid recruitment of stimulated nerve fibers due to their large spatial distributions. It is therefore necessary to reduce the magnitude of these ground currents.

The gross distribution and nonmonotonic nature of the ground currents suggest that they are largely determined by the ground electrode interface impedances. The amount of

Fig. 1. Cochlear implant test electrode showing ground current measurement scheme. A current monitor is successively switched into each ground electrode line during stimulation of active electrodes A1-A4.

Fig. 2. Ground currents arising from stimulation of A1, A3, and A4 in a single patient. Positions of actives are indicated by the corresponding symbols on the abscissa.

current returned to a given ground electrode will be determined by both the fluid-path impedance from the active to the ground electrode and the impedance of the tissue-fluid interface for that ground electrode. Hence, if the interface impedances are significantly larger than the fluid-path impedances, the ground current distribution will be large in extent since the impedance paths to each ground electrode will be similar. Furthermore, small variations in shape and size of ground electrodes will be reflected in variable interface impedances and concomitant ground current variations as observed.

Further evidence for this is found in the results of Fig. 3 in which the ground currents are plotted for stimulating A3 when the electrode was placed in saline-filled tubes of diameters 5, 3.5, and 1.5 mm, and in a saline-filled beaker. The results for stimulation in the beaker show nonmonotonic features identical to those measured in the human cochlea, although the ground currents were more widely spread. Measurements made in the 5 and 3.5 mm tubes were almost identical to those made in patient 1, and the current distributions for the 1.5 mm tube were significantly more localized to the stimulus site and essentially monotonic. All of these results are consistent with the hypothesis that in the human cochlea, the ground current

Fig. 3. Ground currents for stimulation of A3 when measured in the human cochlea, saline-filled tubes of diameters 5, 3.5, and 1.5 mm, and in a saline-filled beaker. C9 was open circuit when measurements were made in saline beaker and 5 and 1.5 mm saline tubes.

distributions for the present array are determined mainly by the interface impedances. For example, fluid-path impedances in the saline beaker will be very low due to the large saline volume. Since the ground current distribution in this case exhibits characteristics similar to those for measurement in the human cochlea, it follows that the interface impedances must be significantly higher than the fluid-path impedances in both cases. However, in the 1.5 mm diameter tube, the fluid impedances will be relatively more significant, and the overall ground current distribution will depend less on the interface impedances as seen in the essentially monotonic decrease of ground current with distance from the stimulus site. It would appear that saline-filled tubes of diameters 3.5 and 5.0 mm represent good *in vitro* models of the human cochlear electroanatomy for the investigation of the properties of this electrode.

Other experiments performed with arrays containing electrodes of different shapes and surface areas indicated that individual variations in impedances of the ground electrodes for the present array were almost entirely due to differences in the surface areas and the degree of tightness to which the electrode was welded around the silicon rubber tube. Large electrode surface areas and loosely fitting electrodes resulted in large ground current sinking. This phenomenon explains the basal propagation of the ground currents where the electrode surface areas were greatest due to the larger diameters in this

Fig. 4. Ground currents measured in a 5 mm diameter saline-filled tube for a uniform diameter (0.64 mm) electrode array with smooth platinum ground electrodes, and with subsequent platinization of the ground electrodes.

position. From measurements made on single electrode pairs, the interface impedances of the ground electrodes are 1–2 k Ω for the range of ground currents measured. However, a 3.5 mm diameter saline-filled tube, which is an adequate model of the electrical environment of the human cochlea for investigating the properties of the array, exhibits a longitudinal impedance of 20 Ω /mm (or 30 Ω between ground electrodes).

It follows that in order to reduce ground current spread, one method is to reduce the ground electrode interface impedance. Fig. 4 presents results made on a uniform diameter (0.64 mm) electrode array where ground currents were measured in the usual way and then again after the ground electrodes were plated with a layer of platinum-platinum black to reduce the interface impedances. There was clearly a greater attenuation of the ground current in the latter case. While platinum-platinum black is not necessarily suitable as a long-term implant material, some form of mechanically roughened platinum surface [7] may suffice in lowering the interface impedance. Alternatively, the problem may be solved electronically where switching circuits can be used to "open circuit" all ground electrodes apart from the ground electrode adjacent to the active electrode to provide a true bipolar stimulus.

DISCUSSION

The measurements presented expose some potential problems should it prove necessary to use coincident stimulation from two or more electrodes. If discrete bipolar stimulation is attempted using ground electrode switching, which makes common the ground electrode adjacent to each active electrode and open circuits the remainder, large stimulus interac-

tions would be expected with coincident stimulation because of the high interface impedances of these ground electrodes. However, coincident stimulation, when performed with an array which has a ground electrode with a low interface impedance, will produce ground currents which are reasonably well localized to the site of each active electrode, with the possibility that the stimulus currents will also be localized to the stimulus sites.

Another possible method of reducing such interactions would be to use an electrode which displaces a greater volume of perilymph than the present electrode, thus increasing the tissue impedances. However, from the data presented, this is unlikely to occur. The electrical properties of the present electrode are quite accurately modeled by saline-filled tubes of diameters 3.5–5.0 mm. The average cross-sectional area of the human scala tympani is 1.5 mm² [11], which is equivalent to a circular tube of diameter 1.4 mm. Since the resistivity of perilymph is almost identical to that of saline, and the tube dimensions are much larger than the human scala tympani, it appears that other cochlear structures also determine the electrode load impedance. These structures could conceivably disperse the ground currents, even if the electrode system were designed to completely displace all the perilymph from the scala tympani. Thus, while electrode arrays have been designed with moulded Silastic carriers which fit the scala tympani [8], [9], and which exhibit well-localized currents for single electrode stimulation [10], electrode interactions could be expected with coincident stimuli. This is all the more so since the interface impedances are presumably much higher than for the present electrode because of the smaller electrode surface areas employed.

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