

## COMBINATION AND COMPARISON OF ELECTRIC STIMULATION AND RESIDUAL HEARING

*Peter Blamey, Gary Dooley, and Elvira Parisi*

Cooperative Research Centre for Cochlear Implant, Speech and Hearing Research,  
384-388 Albert St, East Melbourne, Victoria 3002, Australia

**ABSTRACT**

Speech processing for cochlear implant users has now reached a level where some severely hearing-impaired hearing aid users may be better aided by a cochlear implant, or a hearing aid and implant together. This paper reviews studies comparing the loudness, pitch, and vowel perception in opposite ears of adults using cochlear implants and hearing aids. A study of nine subjects showed narrow dynamic ranges and steep loudness growth in both ears. Mismatches in aided thresholds and dynamic ranges at different frequencies resulted in highly variable loudness differences between the ears for some subjects. A comparison using pure tones showed that the electric pitch depended on both rate and electrode site. Pitch of electrodes was lower than expected from the characteristic frequency distribution in a normal cochlea. Synthetic vowels were used to show that signals presented via the implant and hearing aid may be perceived as different vowels in the two ears.

**I. INTRODUCTION**

Over the last decade, advances in speech processing for cochlear implants have led to improved speech perception results: mean open-set sentence scores for groups of postlinguistically deafened adult implant users at the Melbourne Cochlear Implant Clinic are shown in Table 1.

Years	Processing strategy	Number of patients	Mean score
1982-85	F0F2	13	16%
1985-89	F0F1F2	32	39%
1989-93	MPEAK	21	57%

*Table 1 - Mean scores on the CID Sentence Test without lipreading for groups of cochlear implant users assessed 3 months postoperatively*

The processing strategies evolved from one that coded only fundamental frequency, second formant frequency and amplitude (F0F2) [1], to include first formant information (F0F1F2) and finally, amplitude information for three higher frequency bands (MPEAK) [2]. The most recent strategy (SPEAK) has produced further improve-

ments in perception [3] but clinical data are not available.

These advances suggest that there may be a sizeable population of severely-to-profoundly hearing-impaired people now using hearing aids who may obtain improved speech perception from an implant. At least some of this group will benefit from combined use of a hearing aid and implant together. In Melbourne there has been a variety of responses: One person continued to use a hearing aid because the combined effect was "more natural". One person rejected the implant because it "interfered with the hearing aid use". A third person rejected the hearing aid because "it didn't contribute anything extra".

This paper reviews three studies using acoustic and electric stimulation in opposite ears and discusses the consequences for combined hearing aid and implant use.

**II. LOUDNESS**

This study measured the loudness of monaural electric and acoustic stimuli as a function of intensity across a range of electrode positions and frequencies. The results were used to compare the perceived loudness of signals processed by the cochlear implant and the hearing aid.

**2.1 Method**

Fifteen stimulus levels were chosen to cover the range from threshold to very loud for five different frequencies, and five different electrodes for nine listeners who used a hearing aid and a cochlear implant in opposite ears. Stimuli were presented in random order and classified as "too soft", "very soft", "soft", "medium/comfortable", "loud", "very loud" or "too loud". Three responses for each stimulus were obtained from each subject, and mean stimulus levels were calculated for each response category to produce iso-response contours and loudness growth functions [4].

**2.2 Main Results**

Figures 1 and 2 show iso-response contours for one subject. Averaged across subjects, the standard deviations of the iso-response levels were 1.4 dB for acoustic and 3.1 SL for electric stimuli (one Stimulus Level is an increment of approximately 2% in charge per pulse). There was a high degree of variability between subjects, but some consistent patterns emerged. For acoustic stimuli, iso-response lines tended to be flatter for

louder categories. This implies a variation in dynamic range across frequency, with greatest dynamic range at frequencies with lowest thresholds. For electric stimuli, dynamic range and threshold were not related in the same way. When expressed as a proportion of the dynamic range from "very soft" to "loud", the mean levels for "soft" and "medium" were consistent across subjects and stimulus type (electric or acoustic), indicating that the results are well-fitted by a single loudness growth function with dynamic range as the only parameter.

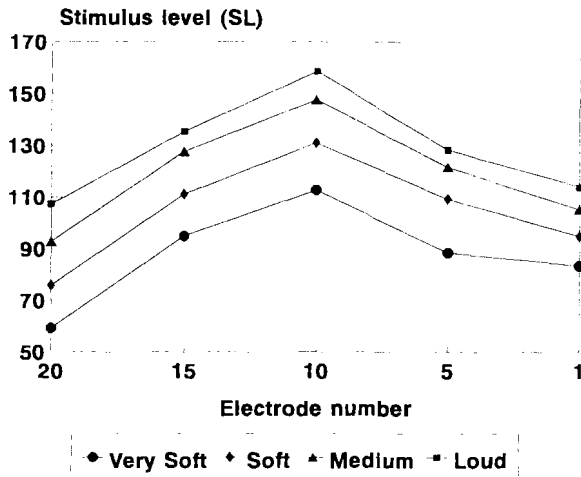


Fig. 1 - Electric iso-response loudness contours for one listener.

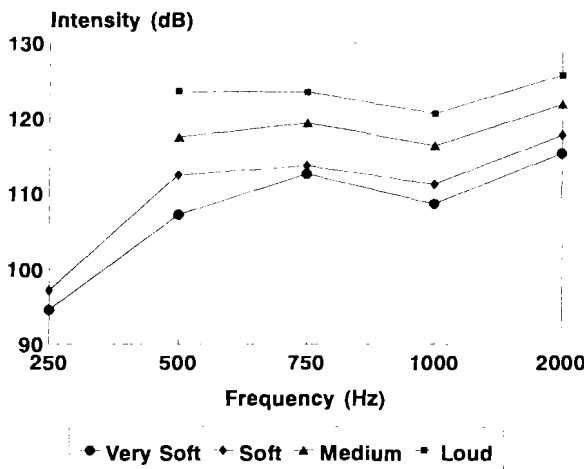


Fig. 2 - Acoustic iso-response loudness contours for one listener. Intensity is in dB SPL for a 2cc coupler.

### 2.3 Loudness Mapping

In the cochlear implant speech processor, amplitudes of formants or fixed spectral bands are measured by an analog-to-digital converter (ADC) and converted to stimulus levels by a look-up table. Thus the 30 dB input range is mapped (non-linearly) onto the dynamic range of each electrode. This gives flexible control of loudness, ensuring that signals are both audible and comfortable. In a conventional hearing aid, gain is specified as a function of frequency, and amplification is a linear function of

amplitude up to the maximum power output. Fig. 3 shows intensities at the microphone of the implant and hearing aid that would result in "very soft" or "medium" loudness for the subject whose results are shown in Figs 1 and 2.

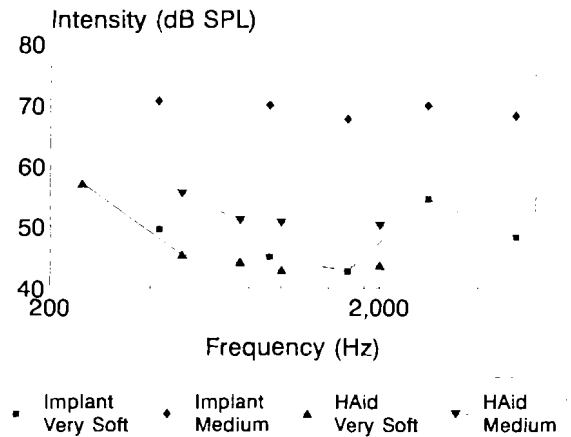


Fig. 3 - Iso-response loudness contours expressed as free field intensities at the microphone of a hearing aid and cochlear implant for one listener.

Because of the loudness mapping strategy used by the cochlear implant, the iso-response curves for all implant users are similar when expressed in terms of input frequency and amplitude as in Fig 3. For hearing aids, different thresholds and dynamic ranges can lead to very different iso-response contours.

In Fig. 3, the loudness of sounds in the two ears is well matched across frequency close to the "very soft" contour, but loudness grows more rapidly in the hearing aid ear than in the implant ear, resulting in a loudness mismatch at higher input levels. In practice, this listener would probably reduce the gain of the hearing aid to achieve a better balance at higher input levels. It is not possible, however, to achieve a loudness balance across all levels because of the difference in dynamic range for the two ears. Other listeners may have an added problem if the aided iso-response contours in the two ears are not parallel across frequency. In this case, which ear is louder may depend on both the frequency and amplitude of the incoming signal.

Clearly, the optimum binaural fitting may be different from the combination of independently optimised monaural fittings of the implant and hearing aid. The effects of loudness mismatches have not been evaluated experimentally, but it is expected that localisation and fusion of the binaural signal into a single auditory stream of information may be affected. Separate studies have demonstrated that inter-aural masking and loudness summation may need to be taken into account also.

### III. PITCH

This study compared the perceived pitch of electric stimuli presented to one ear with acoustic pure tones presented to the other ear. There is very little prior data on the absolute pitch of electric stimuli.

### 3.1 Method

Electric stimuli varying in electrode position at a fixed pulse rate, or varying in rate at a fixed position were first adjusted to be approximately the same loudness as an acoustic pure tone in the opposite ear. Pure tone frequencies were chosen for each subject to cover the range of residual hearing. Electric stimuli were presented alternately with the pure tone and the subject was asked which sound was "higher" in pitch. The two sounds heard were often different in aspects such as timbre, roughness, fullness, etc, making it difficult for listeners to isolate the pitch differences. Repeated judgements for each stimulus (in randomised order) were made by each subject.

### 3.2 Main Results

Fig. 4 shows the results from the experiment in which electric pulse rate was held constant and electrode position was varied. The symbols represent the electrode positions at which responses changed from "implant higher" to "hearing aid higher" as the electric stimulus was shifted in a basal to apical direction in the cochlea. Electrode positions are specified as angles measured from the round window, derived from postoperative X-rays [5]. This compensates for different depths of insertion of the electrode array. The solid line in Fig. 4 shows frequency coordinates on the basilar membrane derived from other studies [6,7]. At the apical end of the array, the pitch matches are two to three octaves lower than the frequencies for the same angular position in normally-hearing listeners. At the basal end, the difference is less.

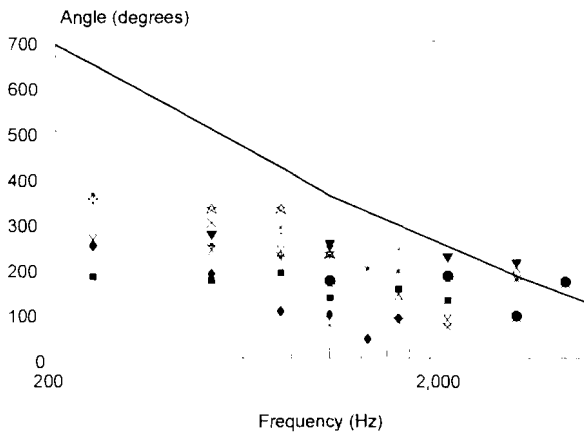


Fig 4 - Frequency vs position in the cochlea from a pitch matching experiment with 9 cochlear implant users (symbols) compared to normal listeners (diagonal line). Pulse rate was 250 pps.

Fig. 5 shows the results of pitch matching when electric pulse rate was varied on the most apical electrode. There was a spread of results across subjects, with a tendency for the "matched" pulse rate to exceed the acoustic frequency at the lowest frequency tested (62 Hz). At higher frequencies, the matched rate was within one octave of the pure tone for most subjects. Temporal coding of pitch in terms of electric pulse rate seems to be

less precise than normal, but does not involve a shift like the place coding of pitch in terms of electrode position.

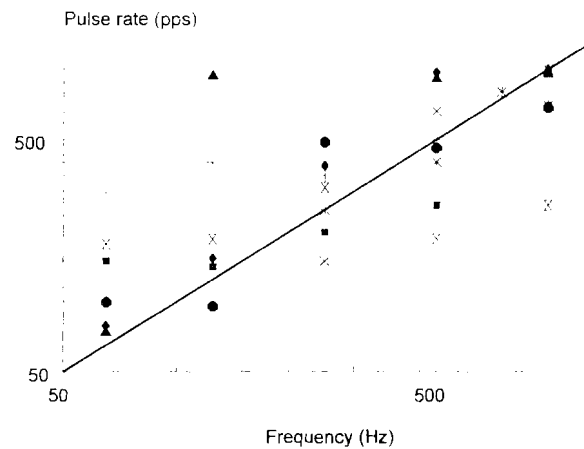


Fig 5 - Frequency vs electric pulse rate derived from a pitch matching experiment with 8 implant users (symbols). The most apical electrode was used.

### 3.3 Frequency Mapping

In the implant speech processor, frequency bands are "mapped" to particular electrodes, and the fundamental frequency of the voice, F0, is represented by the pulse rate. Until recently, it was assumed that the pitch of an electric stimulus would approximate the pitch of a pure tone which produces a maximum of excitation on the basilar membrane at the corresponding position in a normally-hearing listener [7] as shown in Fig. 4. Because the electrode array cannot be inserted into the full length of the cochlear spiral, the mapping of frequencies from 300 Hz to 6 kHz onto electrodes in the basal turn of the cochlea (0-360° in Fig. 4) would have implied a considerable upward pitch shift. The results of Fig. 4 imply that the perceived pitch range for many implant users is actually much closer to the normal range covered by speech stimuli. This result helps to explain why postlinguistically deafened adults can recognise speech at high levels of accuracy within a short time of the implant operation (Table 1).

## IV. VOWEL PERCEPTION

This study used synthetic vowels to investigate the effects of pitch and loudness differences on the perception of speech with implants and hearing aids.

### 4.1 Method

A series of two-formant vowels in /hVd/ frame was synthesised with first formant (F1) frequencies from 300 to 900 Hz at 100 Hz intervals and second formant (F2) frequencies from 600 to 2400 Hz at 200 Hz intervals using the Klatt parallel synthesiser [8]. The resulting stimuli were normalised to the same level in dBA. Each stimulus was presented ten times to the subject, listening through one ear only (implant or hearing aid). The listener responded with one of the eleven steady state vowels of Australian English: /i,ɪ,e,æ,a,ʌ,ɔ,ɒ,u,ʊ,ɜ/.

### 4.2 Main Results

Table 2 shows the stimuli and responses for which at

least six out of ten responses were the same for a listener using a cochlear implant. In this way, a "perceptual vowel space" has been mapped out for the listener. Those stimuli for which no vowel is shown in Table 2 produced no dominant response. Table 3 shows the response pattern obtained when the same listener heard the same stimuli through his hearing aid. For comparison, Table 4 shows the pattern obtained when the responses for four normally-hearing listeners were combined.

600	u	o					
800	u	o		ɒ	ɒ		
1000	u	o		ɒ	ɒ		
1200		o		ɒ		ʌ	ʌ
1400						ʌ	ʌ
1600	u		ɜ		æ		
1800	u					ʌ	æ
2000	u	u	ɜ	ɛ	æ	æ	æ
2200	u		ɛ	ɛ	æ	æ	
2400	u	u	ɛ	ɛ		æ	æ
	300	400	500	600	700	800	900

Table 2 - Synthetic vowels perceived by a cochlear implant user. The first column shows F2 frequency. F1 frequency is shown in the last row.

600							
800					ʌ		
1000							i
1200		i					
1400						ʌ	
1600						ʌ	
1800	i						
2000					ʌ		
2200		i	i	i			
2400				i		ʌ	
	300	400	500	600	700	800	900

Table 3 - Synthetic vowels perceived with a hearing aid in the non-implanted ear of the same listener.

600	u						
800	u	o		ɒ	ɒ		
1000		o		ɒ	ɒ	ɒ	
1200	u	o		ɒ	ɒ		
1400	u		ɜ			ʌ	ʌ
1600	u	ɜ	ɜ	ɜ		ʌ	ʌ
1800	u	ɜ	ɜ	ɜ		æ	
2000	u				æ	æ	æ
2200			ɛ	ɛ	æ	æ	æ
2400	i	ɛ	ɛ		æ	æ	æ
	300	400	500	600	700	800	900

Table 4 - Synthetic vowels perceived by 4 normally hearing listeners.

The response patterns for the implanted ear and the normal listeners are similar, and the hearing aid pattern is

quite different. The number of filled entries in Tables 2-4 are 40, 12, and 43 respectively. This probably indicates that the formant information is perceived more clearly with the implant than the hearing aid. For eight stimuli, the listener heard different vowels in the two ears. These differences may have arisen from differences in the perceived pitch and relative loudness of the F1 and F2 components in the signals.

## V. BIMODAL SPEECH PROCESSING

It seems reasonable to suppose that sounds from the two ears will combine best if they are well matched in their pitch and loudness variations. A combined implant/hearing aid speech processor has been developed to investigate this hypothesis [9]. The processor has the flexibility to "map" amplitudes and frequencies in both ears to compensate for differences such as those observed in the studies described in this paper.

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**Author/s:**

BLAMEY, PETER; Dooley, Gary; Parisi, Elvira

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