Investigation of Wireless Data Transmission between Hearing Aids

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ABSTRACT
Electrodes were developed for use in a wireless data-sharing system between hearing aids. Electrodes were placed on the outsides of custom-fit, in-the-ear (ITE) and one-size-fits-all behind-the-ear (BTE) hearing aid shells. Impedance measurements and gains were collected for six electrode-pair configurations at 21.5, 64.8, 128.3, and 216.4 MHz for multiple subjects. Electrodes were coated with a standard hearing aid lacquer or with Galxyl-parylene. Preliminary comparisons of impedance values of similar electrodes inserted in subjects' ears showed that for each design, impedance values were fairly constant for all subjects, so mismatch loss variation is not significant across the range of impedances obtained for different subjects. Thus a standard impedance can be chosen for ease of manufacturing hearing aids using this technology. Transducer losses across subjects' heads were generally between 60 and 80 dB at all tested frequencies, but were lower at higher frequencies.

1. INTRODUCTION
Applications for data sharing hearing aids have proliferated in recent years. In particular, there are multiple algorithms that improve the ability of hearing impaired individuals to localize sound and hear in noisy situations [1, 2, 3, 4]. These algorithms require that signals from at least two microphones be compared. In the past, this was done using a wired array of microphones, for example in [5] [6], which was cumbersome. In a step toward wireless communication, a low-power transceiver system specifically for hearing aid applications was developed [7]. However, no study of antennas suitable for hearing aid communication across a wearer's head had been published. To connect devices such as wristwatches and PDAs, other groups have proposed larger antennas that use the skin as a waveguide or depend on a closed circuit with return current conducted over a path outside of the body. These systems are not practical for communication between two hearing aids because of size and power requirements [8, 9, 10, 11].

The most useful suggestion for antennas came from Luethi [12], who proposed electrodes forming a capacitor on the surface of an in-the-ear (ITE) hearing aid for wireless remote control of hearing aid functions, but no details of implementation had been made public. In our investigation, it became apparent that the capacitance between each electrode and the skin nearest it was the most important parameter, but measuring losses across a human head with pairs of such electrodes poses unique challenges in terms of measurements. This work focuses on those challenges and their results with human subjects.

2. ELECTRODES
Coated electrode pairs were constructed on ITE and behind-the-ear (BTE) hearing-aid shells. Initial attempts to characterize the impedance of these electrodes in human subjects' ears demonstrated that some care must be taken in electrode shape and placement.

2.1 ITE Configuration
Electrode pairs were made for each of nine subjects using two strips of 90 μm thick copper tape on the outside of custom-fit ITE hearing aid shells. Originally, we chose to make the electrodes as large as possible, covering roughly two-thirds of the sides of the shells. They were insulated from the skin by a thin layer of conventional hearing-aid lacquer or Galxyl-Parylene, a newer lacquer. With this configuration, the capacitance, and thus the impedance of the electrode pairs and mismatch losses in later measurements, changed if the skin moved relative to the electrodes. This change came from the fact that the sides of the ear canal changed shape when the subject's jaw moved. Thus, 3 by 10 mm copper strips were used on the top and bottom of the shells (Figure 1). These parts of the ear canal are relatively stable, providing less variation in skin capacitance and thus, electrode impedance.

2.2 BTE Electrodes
For ease of future production, symmetrical electrodes were fitted on the body of the BTE so that it could be used on either ear.

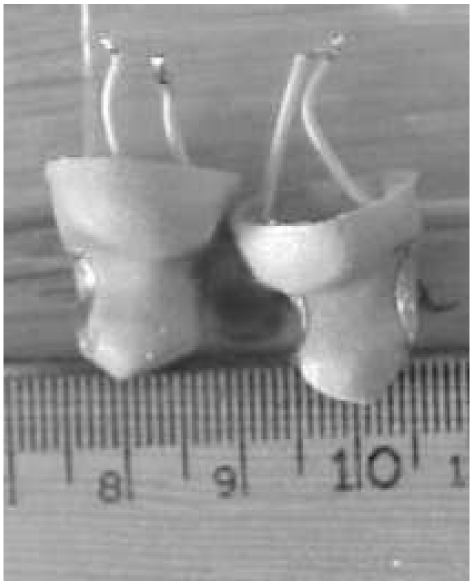

Figure 1: Example of ITE electrodes

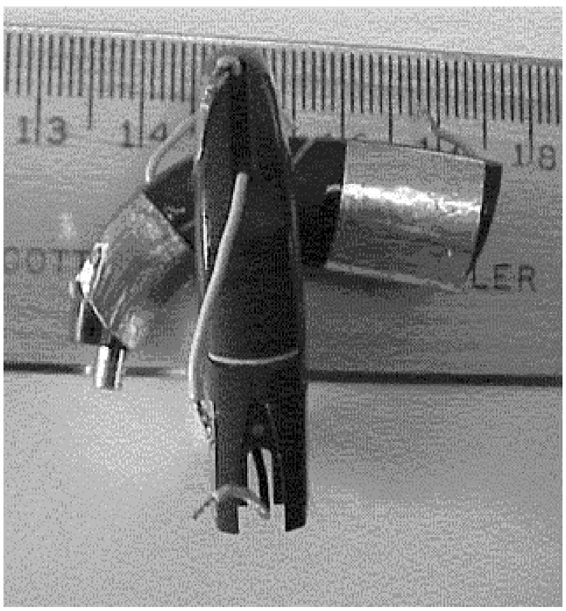

Figure 2: Example of BTE electrodes

As with the ITE electrodes, consistent skin contact was required to minimize impedance changes. We determined that for different ear shapes among our subjects, the most reliable contact was along the inside of the curve of the BTE body. Thus, electrodes were wrapped around the inside curve of the shell perpendicular to the plane of symmetry and separated by as much as possible to reduce the capacitance between them. Electrodes were placed at the bottom of the aid around the battery case and at the narrow end near the hook (Figure 2). Electrodes were made of strips of copper tape 9 or 15 mm wide and cut to fit around the shells. They were coated with conventional hearing aid lacquer or Galxyl-Parylene.

3. MEASUREMENTS

Electrode pair impedances were measured using an HP 8753ES Network Analyzer with an Agilent 85047A S-parameter test set.

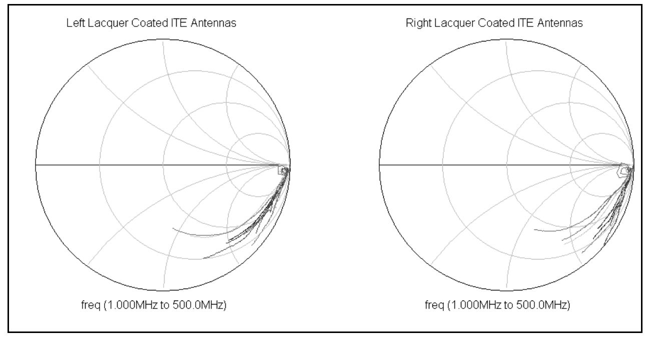

Figure 3: Examples of ITE electrode impedances between 1 and 500 MHz when they were worn in the nine subjects' ears.

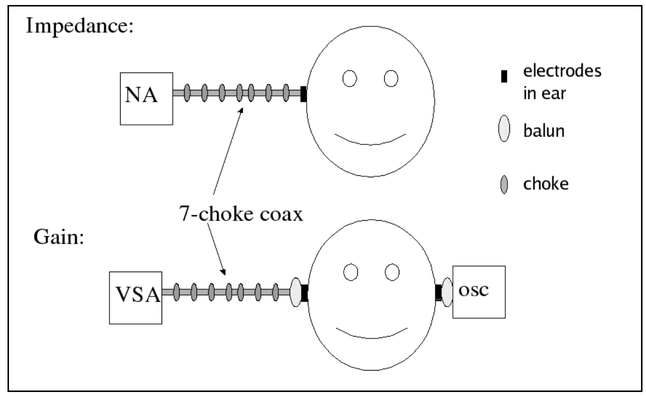

Figure 4: Diagrams of measurement configurations for electrode impedance and gain across the head.

Electrodes on custom-fit hearing aid shells were inserted in subjects' ears during the measurement, which made it necessary to make connections to the machine using a roughly 100 cm length of small-diameter coaxial cable. To minimize currents on the outside of the coaxial cable and minimize the resulting effects on the measured electrode impedance, seven chokes were placed along the cable (Figure 4). The chokes were realized by wrapping 3 turns of the cable around toroidal ferrite cores (FT-50 #61). Efforts were made to place the chokes on either end of the cable as close to the ends as possible, with the remaining five chokes evenly spaced between them, leaving 3 to 4 cm between them. We calibrated at the end of the coax using a 50 Ω chip resistor. Impedances changed by less than 10% at all points in the range of interest when the wire was moved or touched. 21 MHz impedance measurements were obscured by the resonance between the inductance of the chokes and the capacitance of the antennas. ITE impedances were extremely repeatable, but BTEs were more difficult to position, creating uncertainties in the impedances (Figure 3).

For each gain measurement across a subject's head, the left electrodes were attached through a balun to a Colpitts oscillator with a power output between –3 and –6 dBm at each of four frequencies: 21.5, 64.8, 128.3, or 216.4 MHz (Figures 4 and 5). Each oscillator was constructed on a 1.5 × 2 cm circuit board and was powered by a standard hearing-aid battery. The oscillator impedances were approximately 50 Ω across frequencies of interest. The balun was a current balun consisting a bifilar winding of 28-gauge (0.32 mm) wire wound nine times around an FT-50 #63 ferrite core. The right electrodes were attached

Figure 5: An example of an ITE in a subject's ear with an oscillator attached through the balun.

through a similar balun to the small diameter coax with seven chokes described above. The coax was connected to an Agilent 89441A Vector Signal Analyzer. Baluns were required on both sides of the system to minimize unwanted coupling between the oscillator circuitry extending out of the test subject's ear on one side of the head and the cable between the receiving electrodes and the Vector Signal Analyzer on the other side of the head.

Several arrangements were tried on the oscillator side to minimize coupling. The balun was essential because without it, the gain changed by at least 5 dB when the oscillator battery was touched, and by much more at the lower frequencies. With the balun, when the battery was touched, gains changed more than 5 dB at 21.5 MHz, but by only 3 or 4 dB at 64.8, sometimes 1 or 2 dB at 128.3, and not at all at 216 MHz. Rotating the ground plane of the oscillator (the side with less circuitry and more ground plane than the other side) relative to the balun also caused a shift in the gain. By measuring gain changes when we tilted a subject's head to change the distance from the oscillator to the head, we concluded that the ground plane should be directed away from the subject's head. With that orientation, tilting the head did not produce any gain change that could not be accounted for by the earshell shifting in the ear because of the weight of the oscillator, so we did not attempt to keep the oscillator farther from the subject's body.

4. RESULTS

The received signal levels that we measured combined with the known output power of the oscillator used provided a measure of the insertion loss of the electrodes/head 2-port in the 50 Ω measurement system. These results are shown for ITEs in Figure 6 and BTEs in Figure 7. Outlying points in both figures are due to ill-fitting hearing aid shells. The outlier in the ITE data came

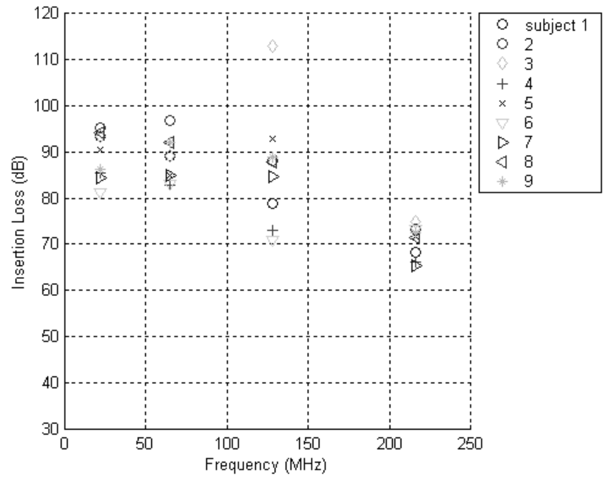

Figure 6: Insertion loss with lacquer-coated ITE electrodes for 9 subjects. Uncertainties were ±3 dB.

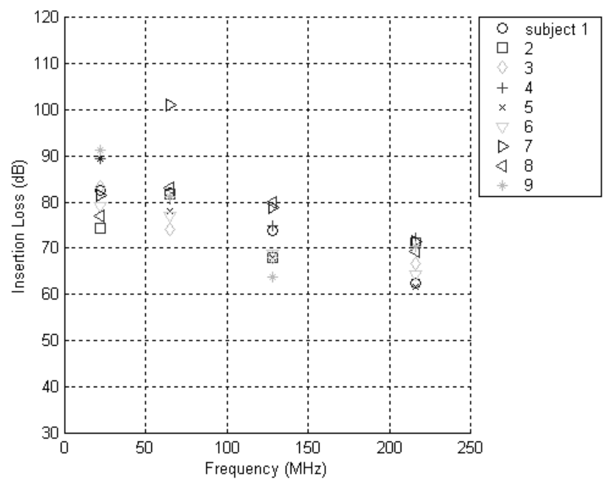

Figure 7: Insertion loss with lacquer-coated BTE electrodes for 9 subjects. Uncertainties were ±5 dB.

from a subject who, had he actually required a hearing aid, would have needed his audiologist to send away for a better-fitting shell. Common hearing-aid fit-adjustment procedures are insufficient on occasion, and his data highlights the importance of fit and electrode stability. The BTE data outlier came from a subject with large ears. The one-size-fits-all BTE shells did not wrap as far behind his ears as they had for other subjects, and were thus less stable. Real hearing aids presumably would not suffer from this difficulty because they would be secured by the insert for the ear canal, a component we chose to leave out to avoid further custom fitting.

Our DSP collaborators informed us that a loss of no more than 80 dB would be acceptable for their algorithm to function [13]. Our insertion losses were acceptable for both ITEs and BTEs only at 216 MHz. A lower frequency would be preferable in terms of power usage. In a step toward reducing that loss, better impedance matching at the source and load would be required. It would not be economical to build custom matching networks for

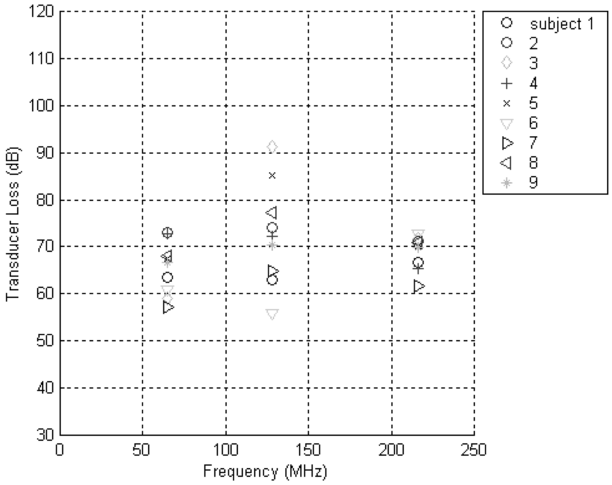

Figure 8: Transducer loss with lacquer-coated ITE electrodes for 9 subjects. Uncertainties were ±3 dB.

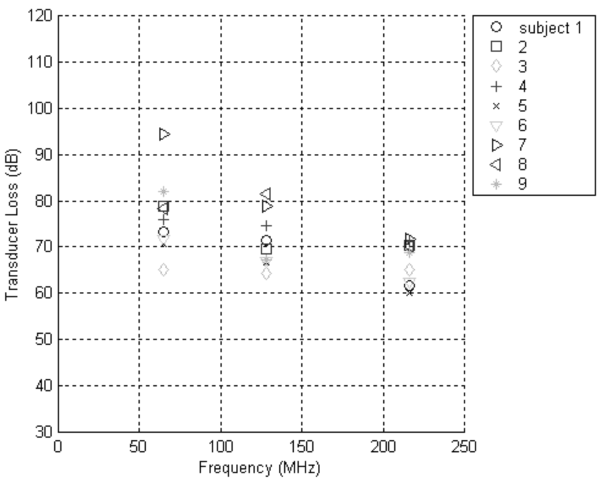

Figure 9: loss with lacquer-coated BTE electrodes for 9 subjects. Uncertainties were ±3 dB.

every hearing aid produced, so to provide a realistic measure of the transducer loss that would be present in a real system, we calculated the transducer loss in a system with source and load impedances equal to the conjugate of the average of the electrode impedances measured for all subjects for each configuration. Based on the measured impedances, a fixed matching network designed to match to the average electrode impedance would result in total mismatch losses of less than 2 dB. The transducer loss results are shown in Figures 8 and 9. Losses at 21 MHz could not be calculated and are not included, but losses at the other three frequencies were lower than 80 dB. Losses remained lowest at 216 MHz, where the transducer losses were below 75 dB for every antenna configuration. The differences between Galxyl-Parylene and lacquer coatings were not significant.

These results demonstrate that low-power, electrode-based body area networks are indeed feasible and worthy of continuing investigation.

5. ACKNOWLEDGMENTS

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