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Low-bandwidth binaural beamforming

S. Srinivasan

An efficient beamforming scheme for wireless binaural hearing aids is proposed that provides a trade-off between the transmission bit rate and the amount of noise reduction. It is proposed to transmit only the low-frequency part of the signal from one hearing aid to the other, which is used in a binaural beamformer to generate the low-frequency part of the output. The high-frequency part is generated by a monaural beamformer using only the locally available microphone signals. The trade-off can be attained by adjusting the cutoff frequency of the lowpass filter. For speech sources with a 8 kHz bandwidth in the presence of an interfering source, it is shown that good performance can be achieved with a cutoff frequency of 4 kHz.

Introduction: Recently, beamforming schemes using signals from both left and right hearing aids (referred to as binaural beamforming), have received much attention [1]. Compared to a monaural solution (using signals from a single hearing aid), they offer improved noise reduction and the ability to preserve the auditory scene. A drawback of binaural beamforming in hearing aids is that it requires the transmission of audio signals from one device to the other. For aesthetic reasons, a wired connection is unacceptable and wireless transmission is necessary. Wireless transmission is power intensive, with each additional quantisation bit resulting in a doubling of the power dissipation in an analogue-to-digital converter [2]. Thus efficient transmission schemes are of interest.

Modern hearing aids typically contain two or more closely spaced microphones in end-fire configuration, which offer good noise reduction properties at high frequencies [3]. The microphones in a binaural array are spaced further apart, with the inter-element spacing being equal to the distance between the two ears. As a result, binaural systems offer good noise reduction performance at low frequencies. This Letter combines the two above-mentioned features. It is proposed to transmit only the low-frequency part of the signal from one hearing aid to the other, which is then used in a binaural beamformer together with the locally available microphone signals at the receiving end. The high-frequency part is processed by a monaural beamformer using only the locally available signals. Transmitting only the low-frequency portion of the signal requires less bandwidth and power than transmitting the entire signal. By adjusting the cutoff frequency of the lowpass filter, any desired trade-off between noise reduction performance and bandwidth/power consumption can be attained.

A binaural enhancement system with reduced bandwidth requirements is discussed in [4], where instead of transmitting all microphone signals from one device to the other, only a filtered combination is transmitted. However, the entire frequency range of the signal is transmitted. This Letter relies on the fact that the binaural array is most effective at low frequencies and thus only the low-frequency part of the signal needs to be transmitted.

Signal model: A frequency-domain model is considered. Assume that there exists a desired source $S(\omega)$ with power spectral density (PSD) $E\{S(\omega) S^*(\omega)\} = \Phi_s(\omega)$, and an interfering source $I(\omega)$ with PSD $E\{I(\omega) I^*(\omega)\} = \Phi_i(\omega)$, where ω is the normalised angular frequency, $E\{\cdot\}$ is the statistical expectation operator, and $\dagger$ indicates complex conjugate transpose. The signal observed at the k th microphone on the left hearing aid can be written as

$$X_L^k(\omega) = H_L^k(\omega)S(\omega) + G_L^k(\omega)I(\omega) + U_L^k(\omega), \quad k = 1 \dots, N \quad (1)$$

where $H_L^k(\omega)$ and $G_L^k(\omega)$ are the transfer functions between the k th microphone on the left hearing aid and the desired and interfering sources, respectively, and N is the number of microphones on the left hearing aid. $U_L^k(\omega)$ corresponds to uncorrelated (e.g. sensor) noise at the k th microphone on the left hearing aid, with the assumption $E\{U_L^k(\omega) U_L^l(\omega)\} = \delta(k-l)\Phi_u(\omega)\forall k$. Further, $S(\omega)$, $I(\omega)$ and $U(\omega)$ are assumed to be pairwise independent. A similar right ear model can be written as

$$X_R^k(\omega) = H_R^k(\omega)S(\omega) + G_R^k(\omega)I(\omega) + U_R^k(\omega), \quad k = 1 \dots, N \quad (2)$$

where the relevant terms are defined analogously to those for the left ear. Define $\mathbf{X}_L(\omega) = [X_L^1(\omega), X_L^2(\omega), \dots, X_L^N(\omega)]^T$, $\mathbf{H}_L(\omega) = [H_L^1(\omega), H_L^2(\omega), \dots, H_L^N(\omega)]^T$, and $\mathbf{G}_L(\omega) = [G_L^1(\omega), G_L^2(\omega), \dots, G_L^N(\omega)]^T$. The

right ear quantities $\mathbf{X}_R(\omega)$, $\mathbf{H}_R(\omega)$, and $\mathbf{G}_R(\omega)$ are defined similarly. The head related transfer functions (HRTFs) $\mathbf{H}_L(\omega)$, $\mathbf{H}_R(\omega)$, $\mathbf{G}_L(\omega)$, and $\mathbf{G}_R(\omega)$ can be obtained using the spherical head shadow model described in [5].

Beamforming scheme with low bandwidth requirements: In the binaural scheme considered in this Letter, it is assumed that each hearing aid transmits part of its signal received at one of its microphones to the other hearing aid. The beamformer used is a multichannel Wiener filter so that a minimum mean squared error (MMSE) estimate of the desired signal is obtained using the received signal and the locally available microphone signals. In the remainder of this Letter, it is assumed that the beamforming is performed at the right hearing aid using the locally available microphone signals, and the signal received from the left hearing aid over a wireless link. Owing to symmetry, the discussion also applies to beamforming at the left hearing aid. It is further assumed that only one microphone signal is transmitted from the left hearing aid. It is straightforward to extend the discussion to the case when signals from multiple microphones are transmitted.

The signal received at one of the microphones (without loss of generality, the first microphone) at the left hearing aid is lowpass filtered with a cutoff frequency ω_c and transmitted to the right hearing aid over a wireless link. Let $\tilde{\mathbf{X}}_R(\omega) = [X_L^1(\omega), \mathbf{X}_R^T(\omega)]^T$, $\tilde{\mathbf{H}}_R(\omega) = [H_L^1(\omega), \mathbf{H}_R^T(\omega)]^T$, and $\tilde{\mathbf{G}}_R(\omega) = [G_L^1(\omega), \mathbf{G}_R^T(\omega)]^T$. At the right hearing aid, an MMSE estimate of the desired signal as observed at the first microphone can be obtained as

$$\hat{S}_R(\omega) = \begin{cases} E\{H_R^1(\omega)S(\omega)|\tilde{\mathbf{X}}_R(\omega)\} & \text{for } |\omega| \leq \omega_c \\ E\{H_R^1(\omega)S(\omega)|\mathbf{X}_R(\omega)\} & \text{for } \omega_c < |\omega| \leq \pi \end{cases} \quad (3)$$

The expectations in (3) can be evaluated as

$$\begin{aligned} E\{H_R^1(\omega)S(\omega)|\tilde{\mathbf{X}}_R(\omega)\} &= \mathbf{W}_B(\omega)\tilde{\mathbf{X}}_R(\omega) \\ E\{H_R^1(\omega)S(\omega)|\mathbf{X}_R(\omega)\} &= \mathbf{W}_M(\omega)\mathbf{X}_R(\omega) \end{aligned} \quad (4)$$

where $\mathbf{W}_B(\omega)$ and $\mathbf{W}_M(\omega)$ are the binaural and monaural multichannel Wiener filters given by

$$\begin{aligned} \mathbf{W}_B(\omega) &= E\{H_R^1(\omega)S(\omega)\tilde{\mathbf{X}}_R^\dagger(\omega)\}(E\{\tilde{\mathbf{X}}_R(\omega)\tilde{\mathbf{X}}_R^\dagger(\omega)\})^{-1} \\ \mathbf{W}_M(\omega) &= E\{H_R^1(\omega)S(\omega)\mathbf{X}_R^\dagger(\omega)\}(E\{\mathbf{X}_R(\omega)\mathbf{X}_R^\dagger(\omega)\})^{-1} \end{aligned}$$

The completely binaural and completely monaural estimates can be seen as special cases of the above scheme where the cutoff frequency ω_c is π and 0, respectively. The mean squared error (MSE) in estimating the desired signal, $\xi_M(\omega)$ in the completely monaural case, $\xi_B(\omega)$ in the completely binaural case, and $\xi(\omega)$ in the proposed scheme, can be written as

$$\begin{aligned} \xi_M(\omega) &= E\{|H_R^1(\omega)S(\omega) - \mathbf{W}_M(\omega)\mathbf{X}_R(\omega)|^2\} \\ &= |H_R^1(\omega)|^2\Phi_s(\omega)[1 - \Phi_s(\omega)\mathbf{H}_R^\dagger(\omega)\Phi_{x,R}^{-1}(\omega)\mathbf{H}_R(\omega)] \\ \xi_B(\omega) &= E\{|H_R^1(\omega)S(\omega) - \mathbf{W}_B(\omega)\tilde{\mathbf{X}}_R(\omega)|^2\} \\ &= |H_R^1(\omega)|^2\Phi_s(\omega)[1 - \Phi_s(\omega)\tilde{\mathbf{H}}_R^\dagger(\omega)\tilde{\Phi}_{x,R}^{-1}(\omega)\tilde{\mathbf{H}}_R(\omega)] \\ \xi(\omega) &= \begin{cases} \xi_B(\omega) & \text{for } |\omega| \leq \omega_c \\ \xi_M(\omega) & \text{for } \omega_c < |\omega| \leq \pi \end{cases} \end{aligned} \quad (5)$$

where

$$\begin{aligned} \Phi_{x,R}(\omega) &= \Phi_s(\omega)\mathbf{H}_R(\omega)\mathbf{H}_R^\dagger(\omega) \\ &\quad + \Phi_i(\omega)\mathbf{G}_R(\omega)\mathbf{G}_R^\dagger(\omega) + \Phi_u\mathbf{I}_{2N} \\ \Phi_{x,R}(\omega) &= \Phi_s(\omega)\tilde{\mathbf{H}}_R(\omega)\tilde{\mathbf{H}}_R^\dagger(\omega) \\ &\quad + \Phi_i(\omega)\tilde{\mathbf{G}}_R(\omega)\tilde{\mathbf{G}}_R^\dagger(\omega) + \Phi_u\mathbf{I}_{3N} \end{aligned} \quad (6)$$

where $\mathbf{I}_M$ is defined as the $M \times M$ identity matrix.

The output signal-to-interference-plus-noise ratio (SINR) per frequency in the three cases is given by $10 \log |H_R^1(\omega)|^2\Phi_s(\omega)/\xi_M(\omega)$, $10 \log |H_R^1(\omega)|^2\Phi_s(\omega)/\xi_B(\omega)$, and $10 \log |H_R^1(\omega)|^2\Phi_s(\omega)/\xi(\omega)$, respectively. Note that SINR is measured in dB. Fig. 1 shows the improvement in SINR averaged over frequency for each of the three cases, where the improvement is defined as the difference between the output SINR

defined above, and the input SINR is given by $10 \log(|H_R^1(\omega)|^2 \Phi_s(\omega) / |G_R^1(\omega)|^2 \Phi_i(\omega + \Phi_u))$. The desired source is assumed to be located at 0° as is common in hearing aid applications, and the Figure plots the results for different locations of the interferer. To obtain these results, the HRTFs were obtained using the spherical head shadow model of [5], with the radius of the sphere set to 0.0875 m, the distance between the two microphones on a single hearing aid set to 0.008 m, and a sampling frequency of 16 kHz. The desired and interfering sources were assumed to be located 1.5 m away from the head. An averaged speech spectrum was used as the PSD for the two sources. The signal-to-interference ratio (SIR) was 0 dB, and Φ_u was set to 0.01, corresponding to a signal-to-noise ratio (SNR) of 20 dB. Fig. 1a corresponds to a cutoff frequency of 2 kHz, and Fig. 1b to a cutoff frequency of 4 kHz. It can be seen that, at a cutoff frequency of 4 kHz, the proposed scheme performs close to the completely binaural scheme. For interferences located in the rear half plane, good performance is obtained even at the lower cutoff of 2 kHz, as the end-fire monaural array provided performs well at high frequencies. When the interferer is co-located with the desired source at 0° , there is no benefit due to spatial processing, and the only benefit is from spectral processing. When the interferer is located at 180° , due to the front-back ambiguity of the broadside array, availability of the signal from the left ear does not significantly improve the output SINR compared to the two microphone end-fire monaural array.

Fig. 1 Improvement in SINR averaged over 0–8 kHz for completely monaural (dash-dot), completely binaural (dashed), and proposed scheme (solid) for different locations of the interferer; desired source located at 0°
a Cutoff frequency 2 kHz
b Cutoff frequency 4 kHz

Fig. 2 shows the improvement in SINR corresponding to different cutoff frequencies. The desired source is located at 0° , and three locations of the interferer are considered: 45° (dashed), 90° (dash-dot), and 135° (solid). For interference in the rear half plane, the monaural system performs well in the high frequencies, and the binaural system adds value at the low frequencies. Performance increases with the cutoff frequency up to approximately 4 kHz, after which the gain is only marginal. For interferers located close to the desired source, e.g. 45° , the performance gain resulting from closely spaced monaural array is limited, and the benefit provided by the binaural system increases with increasing cutoff frequency.

Fig. 2 Improvement in SINR averaged over 0–8 kHz resulting from proposed scheme for different cutoff frequencies

Values plotted for different interferer locations θ_i : 45° , 90° , 135° . Desired source located at 0°

Conclusions: Transmitting only the low-frequency portion of the observed microphone signal from one hearing aid to the other provides a good trade-off between transmission bit-rate and noise reduction performance. The performance of the resulting system is better than the monaural system and approaches that of the binaural system as the cutoff frequency increases. Simulations using speech sources with 8 kHz bandwidth show that a 4 kHz cutoff provides a good trade-off. The analysis in this Letter has considered the presence of a single interferer. Topics for future work include considering multiple interferers and the effect of reverberation.

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