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Sensing Auditory Evoked Potentials with Non-invasive Electrodes and Low-Cost Headphones

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Abstract

This paper presents a sensor for measuring auditory brainstem responses to help diagnose hearing problems away from specialist clinical settings using non-invasive electrodes and commercially available headphones. The challenge of reliably measuring low level electronic signals in the presence of significant noise is addressed via a precision analog processing circuit which includes a novel impedance measurement approach to verify good electrode contact. Results are presented showing that the new sensor was able to reliably sense auditory brainstem responses using non-invasive electrodes, even at lower stimuli levels.
Keywords—auditory evoked potentials; tinnitus; bio-sensor; electrocochleography; auditory brainstem response

Introduction

We live in a world where there is potentially dangerous noise all around us, from loud equipment in the workplace to powerful music at concerts or even through personal headphones. Whether enjoyable or not, this constant noise risks problems with the health of our hearing. From tinnitus to Meniere's disease these conditions can be seriously debilitating with a significant and prolonged impact on peoples' lives. Diagnosis and investigation of such problems require complex sensors to examine our auditory system.

When stimulated, the auditory system in our body produces a range of electrical signals called auditory evoked potentials (AEPs), which can be used for diagnosing a range of conditions, including Meniere's disease [1] and potentially tinnitus [2]. However, current sensor methods for detecting AEPs require expensive equipment, expert fitting of invasive electrodes and manual analysis by a trained physician.

Electrodes used to measure AEPs have historically been invasive, using either needles or tympanic electrodes, limiting their use in many scenarios. Recent studies have shown that far less drastic electrodes in the outer ear canal, mastoid and the concha are able to provide satisfactory results [3]. The audio stimulus provoking AEPs in experimental conditions is typicaly provided by specialist audiometric insert earphones, which generate a controlled input signal some distance from the ear and couple this input into the ear via a tube. This separation is used to avoid adding electrical interference. Such equipment has limited the technique to use in specialist clinical laboratories.

AEPs produced by the cochlear nerve are called the auditory brainstem response (ABR). The level of these signals is very low, even when measured invasively. When measured non-invasively ABR signals range from 100 nV to 1 µV peak to peak and these waveforms are measured in an electrically noisy environment. Key noise sources are other local body signals e.g. brain (EEG) and muscle (EMG) signals, as well as external noise due to the body acting as an antenna for mains hum and harmonics from sources such as lighting, mobile phones and other radio signals. Many of these unwanted signals are several orders of magnitude larger than the target and hence any system needs to filter and remove these unwanted interferers with advanced analog electronics prior to digital sampling.

In this paper, a new sensor is developed which can sense AEPs such as ABR and cochlear response (known as Electrocochleography (ECochG)) using non-invasive re-usable electrodes integrated within commercially available headphones. The new sensor makes use of a new processing technique and includes a novel impedance measurement to verify electrode connectivity. The new system is shown to allow AEP monitoring in a range of non-clinical environments where standard commercial headphones can be used, opening up the technique to more medical practitioners and researchers working in the fields of occupational safety and hearing loss.

System Requirements And Design

In order to measure ABR with electrodes placed in the concha region of each ear, a sub micro-volt signal must be measured whilst filtering out environmental noise and any noise from the stimulus headphones. Therefore an analog front end is needed that minimises noise and amplifies the wanted signal such that it sits well within the dynamic range of any sampling ADCs.

One of the main methods of reducing noise will be using the common mode rejection of the circuit, significantly reducing any noise sensed by both electrodes (one in each ear). Due to the likely levels of interference, the first amplifier would need to have significant CMRR (common mode rejection ratio), while also having a low input noise. Masood highlights that the “amplifier requires at least 113 dB of CMRR to acquire a signal of magnitude 0.1µV” [4].

A low-noise amplifier will be used to amplify the signal above the noise floor, with a second amplifier providing high gain so that the signal uses a significant part of the dynamic range of the ADC – in this work a Focusrite 2i2, a 24-bit USB sound card, is used for sampling. This allows a higher resolution image of the signal to be captured, and keeps the majority of the noise added by the second amplifier below levels that affect signal measurement.

A. Front End Design

The block diagram shown in Fig. 1 reflects the analog front end which amplifies and filters the signals before the sampler. With the use of dry electrodes, a high input impedance amplifier is required; hence the selection of the instrumentation amplifier AD8224 from Analog Devices, which has input impedance of 10 TΩ, along with an excellent CMRR of 120 dB up to 10 kHz (more than our required bandwidth). Most amplifiers struggle with CMRR at frequencies above 1kHz, which is one of the main driving factors behind any bioamplifier. Additionally, the input current noise of the AD82224 is low, specified at 1fA/√Hz, which given the high source impedance of the signals would be expected to be the largest contributor to input noise.

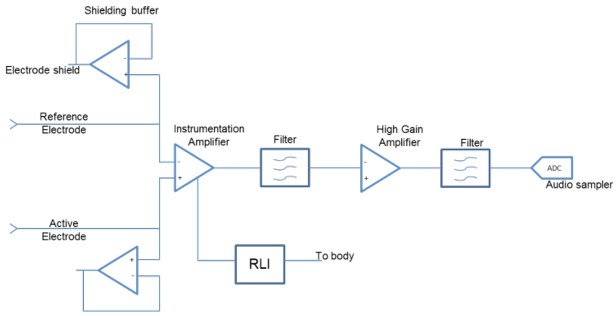

Fig. 1. Block circuit design for analogue front end

After the instrumentation amplifier, the output is filtered, first with a DC RC filter with a cut-off of a few Hertz, then by an active notch filter at 50 Hz for mains hum. Due to the large amount of additional electrical noise after the instrumentation amplifier stage (shown in Fig. 2), an additional common-mode low pass filter was placed in front of the amplifier, with high precision resistors used to maintain the CMRR of the AD8224.

A gain stage was then added to amplify the measured signals so that a detailed picture could be examined using the chosen ADC, then a low-pass sallen-key filter to remove any out of band signals. A final stage converts from single-ended to differential for the ADC using a Texas Instruments OPA1632D op-amp. This gave the system a final overall gain of 53 dB from the input to the sampler.

B. Impedance measurement
One of the main factors which would reduce the CMRR of the circuit would be mismatched skin-electrode impedance between the two electrodes. An additional factor to be avoided is a poor contact which will reduce the signal amplitude and quality overall. To avoid these issues, our new sensor measures circuit to body contact impedance by injecting a small fixed AC current into the body, set by a single op-amp circuit in a classic Howland current source configuration. The resultant voltage across the body is then measured by the ADC to verify the system is configured to produce valid results.

Due to the nature of putting current through the body, two 100k resistors are also placed in series with the electrode contact so that any single fault will result in a maximum of 45µAp-p through the body, and 90µAp-p if one of the resistors also fails. This level is well below recommended safety limits, with most medical standards allowing up to 500µAp-p for a single fault condition. 10^{-3}.

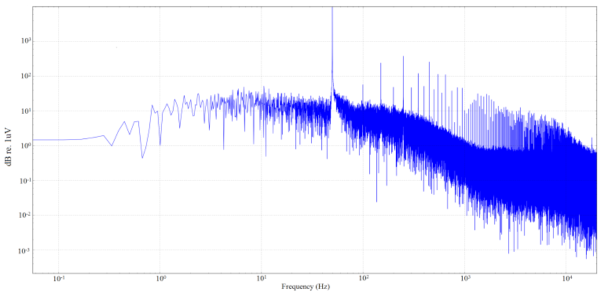

Fig. 2. Typical noise spectrum of signal subject to EM inteference, measured after the instrumentation gain stage.

C. Driven right leg circuit
It is common to improve the CMRR of an ECG system by using a driven right leg (DRL) circuit [5]. The idea is to sense and then invert the input common-mode signal at the instrumentation amplifier, then (with some gain and filtering for stability) drive this inverted signal back into the body. This then reduces the common mode interference for the local environment and hence improves the overall CMRR of the system. A DRL option was implemented, and the complete design was manufactured, as shown in Fig. 3.

Headphone And Electrode Integration
Initially, a range of 8 different commercially available headphone sets were tested to find the most suitable models, with each being calibrated

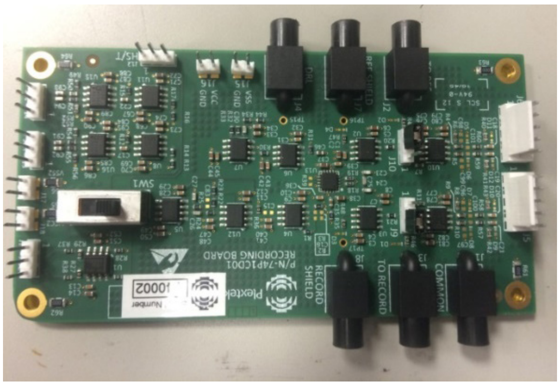

Fig. 3. Analog front end board built to generate drive signals and condition received audio for sampling

using a B&K 4128 Head and Torso Simulator (HATS). This allowed consistent measurements, the correct amount of voltage required for each headphone to produce the same sound pressure at the ear. The transient and frequency response of each headphone was recorded in order to down select a smaller subset for further testing, removing headphones with ringing or skewed frequency response. A reference set of audiometric insert earphones was also calibrated and used to compare between commercial headphones and those used in a clinical environment. A range of electrodes were also tested in combination with the headphones, including a gold plated tiptrode, disposable stick-on mastoid electrodes and a custom material earbud.

Experimental Results
The complete measurement system including laptop, the Focusrite 2i2 sound card, the analogue front end board and the headphones and electrodes as described above were used to take sample readings at a range of stimulus amplitudes.

Initial measurements showed that the DRL circuit was more effective than a 3rd connection to ground, improving the overall CMRR by ≈15 dB. The Focusrite sound card was used to both produce the stimuli, and to record the filtered and amplified output. By timing the recording to stimuli generation, a large number of repetitions can be taken and then averaged together to reveal the ABR signal. The stimulus was a 100µS pulse, which is defined by IEC guidelines [6], with a repetition rate of 17.7 Hz.

Multiple subjects were pre-selected by using pure tone audiometry to test that all subjects have little to no hearing loss from 1 to 8 kHz. All subjects were then tested with the ER-3a reference headphones and mastoid electrodes, to both prove that the custom PCB could sense AEPs (ABRs in this case), and identify the results for reference when using commercially available headphone sets and custom electrode ear-buds.

The goal was to identify a wave-V ABR waveform, which can be seen clearly (and is labelled) in Fig. 4. Here the output of the custom PCB is shown for different stimuli amplitude levels. The system was able to reliably sense ABRs using non-invasive electrodes even at lower stimuli levels. A control test for zero stimuli is displayed as the final waveform, showing a flat response compared to the wave-V waveforms.

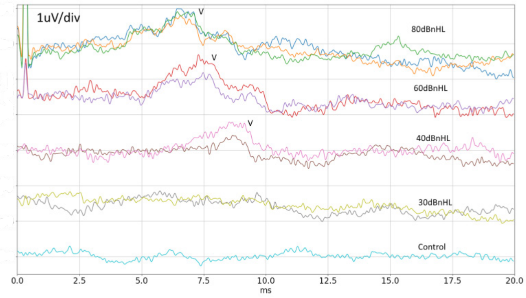

Fig. 4. ABR for ER-3 refrence electrodes for different stimuli levels

Using some Silver-Aluminium silicone material, ear-buds were created, shown in Fig. 5, to place on both the ER-3 reference headphones and Sony XBA-C10 headphones.

The users then used the different headphones while a stimulus was produced and the voltage at the electrodes was recorded by the circuit. The results recorded for the two headphones are shown in Fig. 6, with the ER-3 results shown for a range of different electrodes including custom silicone ear-buds and the Sony headphones with custom ear-buds. From the results we can see that all of the electrodes were usable with the reference system, and importantly the Sony electrodes were able to reliably sense ABR wave-Vs (appearing at ≈6.5ms).

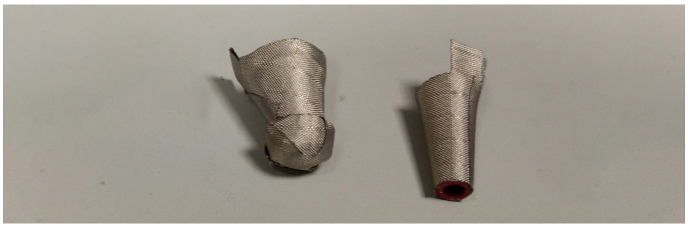

Fig. 5. Ultra-soft Silver-Aluminium silicone material around plastic ear-buds

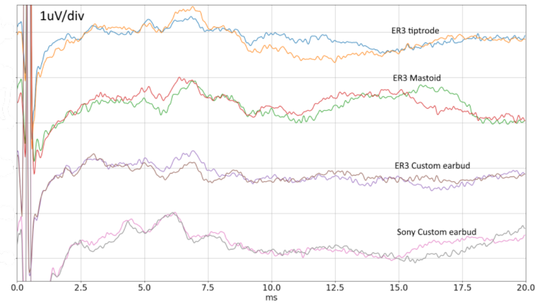

Fig. 6. ABR at 80DBnHL for different electrodes and sony headphones

Conclusion

A sensor architecture for measuring auditory brainstem responses with non-invasive electrodes and commercial grade headphones has been presented. A precision analog processing circuit has been described including a novel impedance measurement approach to verify good electrode contact. Results have shown that the new sensor is able to reliably measure ABRs using non-invasive electrodes even at lower stimuli levels, demonstrating comparable performance to expensive clinic-only systems in use today.

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