Frequency Shifting for Patients with High Frequency Hearing Loss

(Using Digital Signal Processing Chips from Texas Instruments)

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Abstract

Digital Signal Processing is the study of the digital representation of an analog signal and the conversion and processing of those signals. There are many fields in which digital signal processing is becoming important, such as for communications, medical instrumentation, video, and audio. Our group will use digital signal processing chips from Texas Instruments to create an improved hearing device for patients with sensorineural hearing loss (also known as high frequency hearing loss). Our device will first ascertain the maximum and minimum frequency that the individual is capable of hearing. Once these values are known, the device will compress and shift all the sounds a typical human can hear into the impaired range of our patient. This semester, we planned on building the hardware to control input and output, and learn the software kit and programming the actual DSP chip. For input, we will use an electrot condenser microphone for its tiny size. The DSP chip we have selected, the C5509, is designed for use with audio signals and has a built in ADC. We have worked on the conceptual design and because we still have not received our order from TI we decided to use LabVIEW in place of the development kit to construct a working model of the hardware. In future semesters, we will receive and put together the hardware for our device, attempt to streamline the code, reduce background noise, shift voices into the hearing fovea, and compact our device to look like a Bluetooth headset.

Background

Types of Hearing Loss

Our sense of hearing is only possible due to a complex set of organs, all with a specific and necessary purpose. Because there is so much that can go wrong in the ear with so many different parts, doctors have divided hearing impairment and loss into two broad categories: conductive and sensorineural hearing loss.

Conductive

Damage or obstruction to either the outer or middle ear is classified as conduction hearing loss. Obstructing the external canal with either wax or a foreign object is one of the most common causes of conductive hearing loss, but damage or loss of the Pinna is also possible. The other major source of conductive hearing loss is infection in the middle ear. Infections occur in the Eustachian tube (which is narrow and more curved in babies) that will lead to a wax build up in the middle ear, and will dampen middle ear bone vibrations. Other cause of conductive hearing loss include an abnormal pinna, irregularly curved ear canal, scratches or scars of the ear drum, middle ear bone damage, or tight middle ear bone muscles (Figure 1).

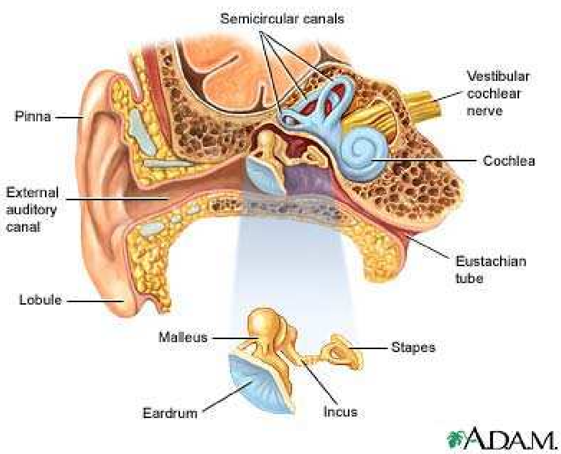

Figure 1. Damage to either the outer ear or middle ear is considered conductive loss. The middle ear is considered everything from the eardrum to the stapes. Sensorineural loss is damage to the cochlea or the auditory nerve. <http://www.nlm.nih.gov/medlineplus/ency/images/ency/fullsize/1092.jpg>

Sensorineural

Sensorineural hearing loss is classified as pretty much anything else that can go wrong. Any damage to the cochlea, auditory nerve, or temporal lobe of the brain may cause hearing loss. Any lesions in the auditory nerve can destroy hearing in both ears, or hearing in just one ear, or certain bands of frequencies with no specific preference. There are many areas of the brain that cause also be harmed and this can lead to a multitude of other symptoms, too many to list them all here. However, the most common type of sensorineural hearing loss is damage to the inner hair cells of the cochlea.

Inner hair cells line the entire cochlea from base to apex, and the curling structure is designed to innervate certain regions of its length for certain frequencies. When the hair cells of one stretch of cochlea become damaged, signals for this frequency range are no longer sent to the brain and are, therefore, not observed. High frequency waves of the same amplitude as a lower frequency wave have more energy. Ergo, the high frequency hair cells are more likely to be damaged if the waves that innervate them are of greater energy. For this reason, high frequency hearing loss is the most common. In fact, our high frequency hearing begins to decrease after we pass 20 years of age. However, the frequency ranges that are used to detect speech will not be affected until, at the very earliest, the mid 70's (if completely natural). With the loud sounds and cheap headphones used today this process is natural process is being rapidly accelerated.

Prevalence of Hearing Loss

Americans don't seem to realize how valuable their hearing is until it is gone. Because of this attitude we continue to ignore the advice of our elders as they yell at us to, "Turn that loud noise down!" It is estimated that currently ten million Americans have an irreversible hearing disorder caused by excessive noise. Also, 30 million Americans are exposed to potentially damaging sound levels every day (NIDCD website, 2007). For many reasons, the number of Americans with a hearing disorder has doubled over the past 30 years. From 13.2 million in 1971 (Ries, 1994) to an estimated 28.6 million in 2000 (Benson & Marano, 1995). Furthermore, the most recent study estimates that 31.5 million people in the United States currently have a hearing disorder (Kochkin, 2005), that is nearly 10% of the population.

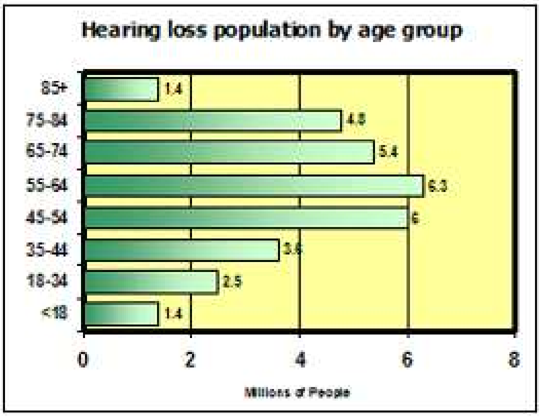

Figure 2. Hearing loss is not only an elderly condition. This graph shows the number of Americans with hearing loss in each age range. (Better Hearing Institute website)

The incidence of hearing loss increases per age group from 17 per 1,000 for those under 18, about 71 per 1,000 for those between 29 and 40, and 167 per 1,000 for those between 41 and 59, up to 314 in 1,000 for those above 65, and up to 50 percent for those older than 75 years of age. (NIDCD website, 2007; Kochkin, 2005) However, this is not an affliction that only affects the elderly. As Figure 2 shows, over 65% of those with a hearing disorder are under the age of 65.

It is most important to treat the hearing loss of the young that are just developing language skills. This is because a hearing deficit will cause children more difficulty when learning vocabulary, grammar, word order, idiomatic expressions, and other verbal skills (NICCYD, 2004). The Department of Education counted 5,775,722 with a known hearing loss attending public schools; of these only 70,767 (1.2%) received help for their disability (DoE, 2002). Studies show that up to 6 in 1,000 newborns are affected by a hearing disorder, and these studies admit the difficulties in detecting hearing loss in those without language skills (Cunningham, 2003). Those under 18 also make up over 8% of those with severe or profound hearing loss (Blanchfield, 2001).

Problem Statement

Our client, Texas Instruments (TI), has given us the task of engineering their digital signal processing (DSP) chips to work in medical applications. This project will give the team the opportunity to use and learn from TI's DSP chips in order to apply them to medical applications. We have selected to use a DSP in implementing a hearing aid that can compensate for sensorineural hearing loss. As mentioned earlier, millions of people suffer from sensorineural hearing loss; however, the intensity of the

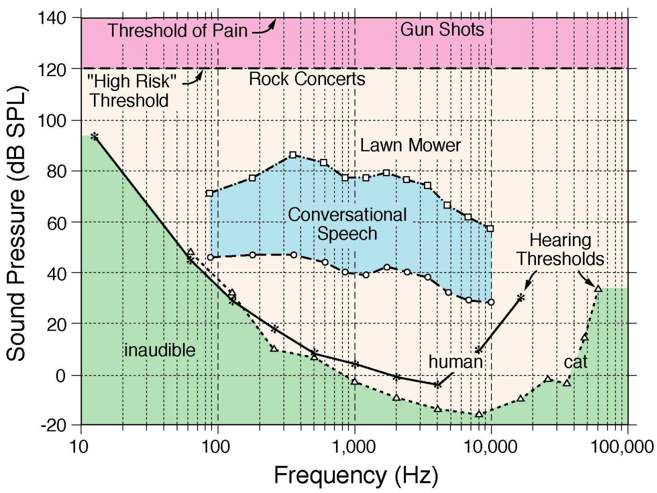

Figure 3. This graph shows the full range of what an average human can hear before any hearing disorders. The blue band indicates the range of normal speech. This will be our targeted region to begin our processing.

hearing loss can vary from person to person. The normal human hearing range is from 10 – 20,000 Hz with normal speech in the range 100 to 10,000 Hz (Figure 3).

Our objective is to design a device that can shift frequencies that are beyond the wearer's hearing range into a frequency range that they can hear. Using DSP technology, will give us enough processing power to be able to shift frequencies in real-time and eliminate lag-time due to processing. Since shifting frequencies can produce some error and noise, especially shifting large frequency range, we will begin work on the normal human speech range, giving us a smaller range to deal with and reduce the chance of electrical noise.

Since sensorineural hearing loss affects different people in varying degrees, the device will need to be calibrated to each user. We plan to develop a frequency scan to let the device calibrate itself to the user. The user will push a button on the device to initiate the frequency scan, which will produce a signal that progressively goes from a low frequency to high. The user will then push another button to indicate when they can begin to hear the signal and once more we they are no longer able to hear the signal. This will allow the device to identify the audible range of the wearer. Any sound outside the this frequency range will be processed and shifted into the wearer's audible range, allowing them to hear what they, by themselves, could not.

We have selected to use TI's C5509 DSP chip. According to TI, the C55 line of chips is power-efficient DSP, which are ideally designed for portable devices such as cell phones, PDA's, and MP3 players (TI website, 2008). These DSP's have ultra-low standby power, which will allow us to minimize the size of the device and maximize the usage life, as we ultimately will be able to supply power with a small battery. Because this device is meant to be worn, it is important to take aesthetical considerations into account. Not only does the device need to work, but also it shouldn't attract negative attention to itself. We would like to make it minimalist or fashionable, and ultimately hope to be able to make the device small enough to resemble a Bluetooth hands-free device for cell phones.

Design

Platform

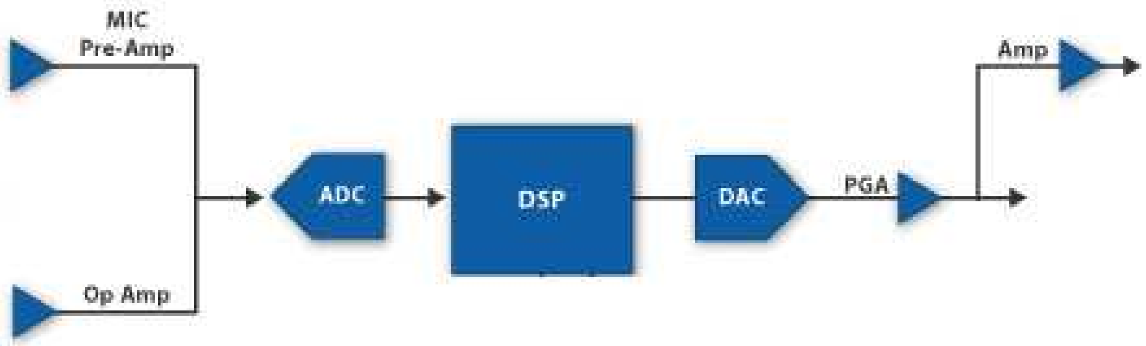

Figure 4. A block diagram for the design of the hardware, or platform, of our digital hearing aid.

A platform was selected based on the project requirements. The original signal is received by the microphone which will be connected to the DSP chip. Because this specific DSP has an analog to digital converter in the chip, there is no need to use an analog to digital converter before the DSP. The analog signal is passed into the DSP chip where it is converted to a digital signal and the necessary operations are performed on the signal (for example, compressing or shifting the frequency so that all the speech frequency will be in hearing range that a person with sensorineural hearing loss.) After the signal modification in the DSP, the processed digital signal is converted to an analog signal using a digital to analog signal converter, which will send the signal to an output, such as a speaker.

Signal Processing

There were several audio signal processing methods for a frequency modification such as the phase vocoder method, pitch scaling, and re-sampling method. Based on the design requirement of this device, signal modification processing should be ideally real time (modification of signal, especially compressing the frequency of the input signal to optimal human hearing range while keeping the duration of signal unchanged). To perform a modification of the signal's frequency the input time-domain signal was converted to frequency domain. For this project the method of frequency compression was chosen using the Short-Time Fourier Transform (STFT), and phase vocoder algorithm. The phase vocoder is one of the solutions for real time scale modification of signals which has a high quality output.

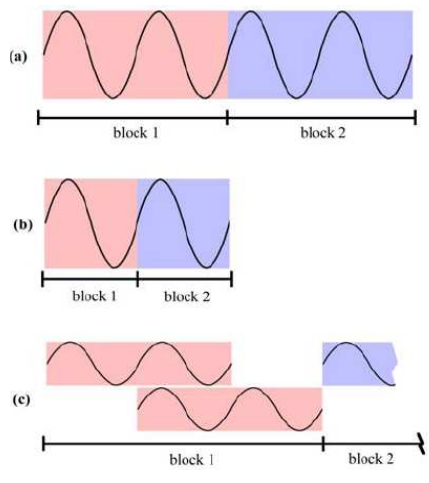

Figure 5. Block-based time domain pitch shifting technique; (a) shows input signal split into blocks; (b) for a down shift, input blocks are truncated by an integer number of periods to create a signal of shorter duration; (c) for an up shift, input blocks are overlapped and added to increase signal duration (from http://www56.homepage.villanova.edu/)

Most of the time pitch scale modification is implemented by time scaling and re-sampling rate conversion (Figure 5).

First, the original signal enters the processer and is divided into smaller segment which allows an STFT to be performed. Then each segment's frequency is modified in the frequency domain. These modified segments would be re-synthesized (inverse Fourier Transform). All of the re-synthesized segments would be resampled and combined.

In the MATLAB test program an alpha factor was chosen which all frequencies are multiplied by. For example, choosing alpha of 0.8 leads to compressing a signal of frequency 10 kHz to 8 kHz. The original signal (.wav format) was then loaded as a matrix. The matrix was divided into segments each with 512 elements. Those segments were then converted to frequency domain using fast Fourier Transform (FFT). The signal was multiplied by the alpha factor in order to compress the frequency in the time domain signal. The resulting signal was re-sampled at a new sampling period (original sampling period multiplied by factor alpha) which resulted in an output signal with the same duration as the original signal.

This program showed great results, providing a proof of concept for our ideas. This program did not take more than several seconds to complete the task of processing about 15 seconds of original audio (this processing duration depends on computer processer). However, there are still some areas of improvement for this method such as a noise problem and actual real time processing. Because the frequency of the entire signal was compressed there was some noise introduced by higher frequency. This problem could be fixed with filtering the high frequency using band pass filter before processing the original signal.

Frequency Sweeper (Automated Hearing Test)

In order to design a hearing device that could be used by people with varying degrees of hearing loss, the device needed to be designed so it could identify each individual wearer's frequency hearing range. To do this, we decided to design a program that could be implemented in the hearing device so that it would be able to find the minimum and maximum frequencies of the wearer's hearing range. We decided to do the initial programming of the hearing test using LabVIEW due to ease of use and the virtual interface LabVIEW provides.

The hearing test runs in two stages, fast and detailed. The initial fast part consists of the hearing tester running a quick frequency sweep (normal sine wave) from 0 – 20k Hz, which is an approximation of the normal human hearing range (20 – 20k Hz). The user pushes a button when they can first hear the signal and once again when they can no longer hear the signal. The device stores the indicated frequencies as approximate minimum and maximum frequencies. The tester then switches to a detailed sweep and runs two slower frequency sweeps of about a 500 Hz range around the previously recorded frequencies for a more precise assessment of the wearer's hearing range. We decided to design the tester as a two-part system because it was the best way to precisely find the min and max frequencies without having to run the frequency sweep for too long of a time. The first sweep runs at 50 Hz steps at a speed of about 150 Hz/s, while the slower sweeps run at 10 Hz steps at a speed of about 10 Hz/s. These values are all easily adjustable and further testing is planned to be done in order to find the most efficient speeds for sweeping.

When running the tester program a major concern that we came across was a clicking noise created by the program and speakers that was audible enough to make it difficult to discern whether or not one was hearing the actual signal. At first we believed it was due to a slight phase offset that occurred when the programs was changing frequencies preventing a smooth continuous signal. We found that increasing the frequency by multiples of ten eliminated the offset problem, but the clicking was still present and we concluded that it was due to the LabView program turning on and off the speakers every time it cycled through the program to increase frequency. We are currently working on eliminating this problem.

Future work

The whole platform is to be built including the circuit, sound receiving device (microphone), and speaker. We can only do this once the parts and development kit from TI finally arrive. When they do come we will also need to learn how the development kit works. In our current programs, we will work on reducing background noise and increasing the Signal to Noise Ratio (SNR). We will want to refine the code to make the processing as fast as possible, and will continue to conduct tests of sound quality. For inspiration we will explore the possibility of combining recent hearing aid features such as noise reduction and feedback reduction in our device. The hearing aid can also be used to determine the person's hearing fovea (the range where a person is best at distinguishing differences between frequencies) and perform compression and shifting to put the most relevant audio information in that band. Housing of the device should be small and if possible, good looking, such as a Bluetooth headset.

References

Benson, V., & Marano, M.A. (1995). *Current estimates from the National Health Interview Survey, 1993.* Vital Health Stat, 10(190).

Blanchfield, B.B. (2001). The severely to profoundly hearing-impaired population in the United States: Prevalence estimates and demographics. *Journal of the American Academy of Audiology*, 12, 183-189.

Cunningham, M., & Cox, E.O. (2003). Hearing assessment in infants and children: Recommendations beyond neonatal screening. *Pediatrics*, 111(2), 436-440.

Kochkin, S. (2005). MarkeTrak VII: Hearing loss population tops 31 million people. *The Hearing Review*, 12(7) , 16-29.

National Information Center for Children and Youth with Disabilities (2004). *Deafness and hearing loss* (Pub. No. FS3). Washington, DC: U.S. Government Printing Office.

National Institute on Deafness and Other Communication Disorders (2007). *Statistics about hearing disorders, ear infections, and deafness*. Retrieved March 3rd, 2008 from http://www.nidcd.nih.gov/health/statistics/hearing.asp.

Ries, P.W. (1994). *Prevalence and characteristics of persons with hearing trouble: United States, 1990-91.* Vital Health Stat, 10(188).

Task Force on Newborn and Infant Hearing (1999). Newborn and infant hearing loss: Detection and intervention. *Pediatrics*, 103(2), 527-530.

Texas Instruments (2008). *Digital signal processing: C55x DSPs.* Retrieved on February 12th, 2008 from http://focus.ti.com/paramsearch/docs/parametricsearch.tsp?family=dsp&sectionId=2&tabId=133&familyId=325&paramCriteria=no

U.S. Department of Education (2002). *Twenty-fourth annual report to Congress on the Implementation of the Individuals with Disabilities Education Act.* Retrieved March 3rd, 2008 from http://www.ed.gov/about/reports/annual/osep/2002/index.html.