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PROGRAMA DE DOCTORADO:
INVESTIGACIÓN APLICADA A LAS CIENCIAS SANITARIAS

INTEGRACIÓN DE PRINCIPIOS FÍSICOS EN UNA UNIDAD DE HIPOACUSIA

INTEGRATION OF PHYSICAL PRINCIPLES IN A HEARING LOSS UNIT

TESIS DOCTORAL PRESENTADA POR Dª MARÍA TERESA PÉREZ ZABALLOS
DIRIGIDA POR:
PROF. DR. ÁNGEL M. RAMOS MACÍAS
PROF. DR. SANTIAGO RODRÍGUEZ FEIJOO

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D JUAN FRANCISCO LORO FERRER, COORDINADOR DEL PROGRAMA DE DOCTORADO INVESTIGACIÓN APLICADA A LAS CIENCIAS SANITARIAS, DE LA ESCUELA DE DOCTORADO DE LA UNIVERSIDAD DE LAS PALMAS DE GRAN CANARIA, EN CALIDAD DE PRESIDENTE DE LA COMISIÓN DE DOCTORADO

CERTIFICA,

Que la citada Comisión en su sesión de fecha diez de Mayo tomó el acuerdo de dar el consentimiento para su tramitación, a la tesis doctoral titulada "**Integración de principios físicos en una unidad de hipoacusia/ Integration of physical principles in a hearing loss unit**" presentada por la doctoranda D/Dª María Teresa Pérez Zaballos y dirigida por los Doctores Ángel Manuel Ramos Macías y Santiago Rodríguez Feijoó.

Y para que así conste, y a efectos de lo previsto en el Artº 8 del Reglamento de Estudios de Doctorado de la Universidad de Las Palmas de Gran Canaria, firmo la presente en Las Palmas de Gran Canaria, a 11 de Mayo de dos mil dieciséis

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DIRECTOR	CODIRECTOR
Ángel M. Ramos Macías	Santiago Rodríguez Feijoó

DOCTORANDO

María Teresa Pérez Zaballos

A mi familia y a Carlos ,que ya es parte de ella

A mis tutores y mis compañeros de trabajo

A Miguel

Pero, sobre todo, a los voluntarios que han participado en los estudios, sin ellos esta tesis nunca hubiera sido posible

EPPUR SI MUOVE

.

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A todos, gracias.

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I. INTRODUCTION

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1.- MOTIVATION AND OVERVIEW OF THIS DISSERTATION

In today's world, where interdisciplinarity plays a key role in research, it is vital to change the traditional schemes that separated professionals according to their degrees. To get a true scientific revolution, different fields of knowledge need to be mixed and hybridized, so that old problems are looked at from new perspectives. This is how innovative solutions emerge.

This dissertation attempts to illustrate this concept. The addition of a physicist to a Hearing Loss Unit has yielded different solutions to clinical problems. The presence of a physicist in a clinical session that listens to the real needs of the health personnel can dramatically speed up innovation regarding surgery, cochlear implant programming, rehabilitation and patient evaluation. In this thesis, work has been done trying to cover all of these aspects.

The thread of the studies presented here is **the improvement of sound perception of cochlear-implanted patients under different conditions**. The work has been divided into the different physical fields that were applied to solve each initial question. Three sound conditions were evaluated. First, speech-in-noise perception mechanisms are studied. Then, how pitch discrimination is affected by electrode proximity was evaluated. Finally, the improvement of sound in patients with high impedances was looked at.

The first part focuses on psychoacoustics, a hybrid field between physics and psychology. Two studies were conducted. The first is a study on the relevance of high frequencies for the perception of speech in noise in normal hearing individuals. This showed that high speech components, above 8 kHz, help to understand words subjected to ambient noise. The second study is a continuation of the first, and explores the effect of long-term speech in noise training in air traffic controllers. This job sector develops speech perception abilities during the course of their work. The special acoustic conditions to which they are subjected train their brains to extract the maximum amount of auditory cues to be able to establish radio communications. These have the same bandwidth as auditory prostheses (implants and hearing aids). Therefore, the study demonstrates how training enhances the ability to understand speech in noise and proposes the inclusion of speech in noise training for people with hearing loss.

The second part relates to electrical psychophysics. It focuses on evaluating the effect of electrode distance to the neural ends as an important factor for pitch discrimination in cochlear implanted patients. This study shows that a closer position to the auditory nerve produces better electrode discrimination.

The third part uses the physical principles of electromagnetism to improve power efficiency of the cochlear implant. It suggests that readjusting the position of the reference electrode at the time of surgery,-placing it as close to the cochlea as possible-, can decrease the electrical current required to produce a response from the auditory nerve. This in turn would bring economic benefits to the patient, as it results in longer battery life.

The results of these studies demonstrate the benefits of including a physicist in the daily routine of a medical department. This can generate solutions *a la carte*. Therefore, the role of the physicist needs to be considered in the new models of organization of health services as a motor of innovation and individual treatments.

2.- THE PHYSICS OF SOUND

2.1.- Anatomy of a sound wave

Sound is a disturbance in the medium caused by a vibrating object and it propagates through a fluid (mechanical wave) in a longitudinal way. This disturbance is due to local pressure changes (p) in the medium, which cause successive regions of high pressure, and therefore high density, and regions of low pressure and low density (figure 2.1a). For vibration to propagate, the particles in the medium must be sufficiently close for their movement to affect their neighbours. Thus, sound propagates better through a solid than through liquid, and through liquid than through gas. Because in vacuum there are no particles that can transmit this vibration, sound waves cannot propagate through it. Hence, sound waves can be described as perturbations that travel through elastic mediums that undergo compression and rarefaction patterns. These vibrations transmitted to the environment generally weaken the longer the wave moves away from its source due to dispersion, reflection and absorption caused by obstacles and by the medium itself. (1)

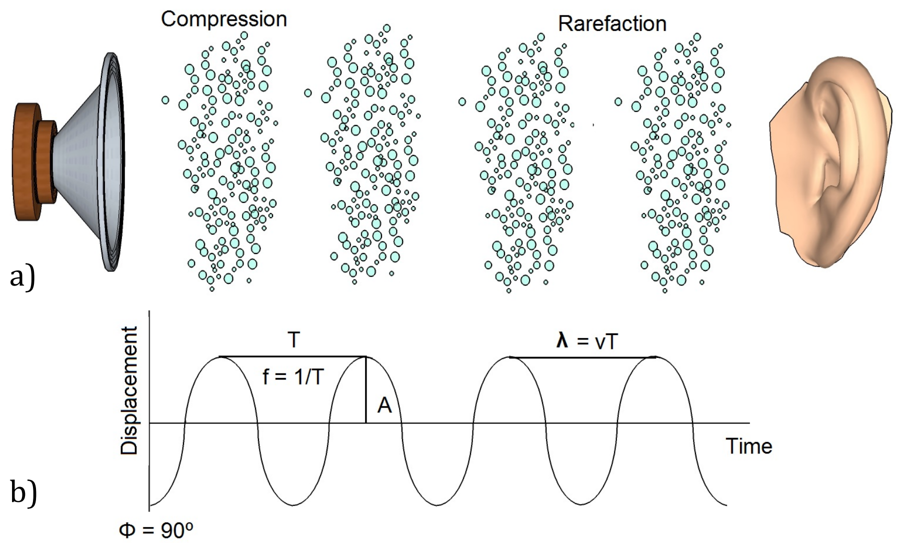

Figure 2.1. A sound wave is caused by the compression and rarefaction of the particles in the medium along which it propagates. This displacement pattern takes the form of a sinusoidal wave when represented along the time axis. Part b shows the parameters that define a wave: amplitude, frequency, period, phase and phase constant. (Own production)

The most usual representation of the sound wave is as the pressure variation of a particle in the medium over time. The simplest wave that it can describe is that of figure 2.1b, which is a sinusoidal variation, resulting in successive, simple compressions and rarefactions. These types of waves represent what is known as pure tones and form the basic building blocks of more complex sounds. To describe the displacement of a particle in the medium (x), six parameters, illustrated in Figure 1b, are required:

1. **Frequency (f):** it is the number of times that a wave repeats itself per second. Pure tones are unique frequencies, since the vibration is always repeated at regular intervals. Its unit is the Herz (Hz).
2. **Period (T):** this is the inverse of the frequency and is the time taken by a particle to complete one full vibration. Its unit is the second (s).
3. **Wavelength (λ):** it is simply the length of one complete wave cycle. The length of one such spatial repetition, known as wave cycle, can be measured as the distance from crest to crest or from trough to trough. In fact, the wavelength of a wave can be measured as the distance from a point on a wave to the corresponding point on the next cycle of the wave. The wavelength is given by the velocity of the wave propagation divided by the period, because this gives the distance traveled by the particle during one cycle ($\lambda = v/T$).
4. **Amplitude (A):** it is the value of the peak pressure variation from the equilibrium ($x=0$) value. It is measured in meters (m).
5. **Phase (2πf•t):** it indicates the instantaneous position of the vibration in the cycle with respect to the time taken as a state of reference (most commonly $t=0$). This way, one full cycle is represented as 360 ° or 2π radians, one half is represented by 180° or π radians, etc. It is measured in radians (rad).
6. **Phase constant (Φ):** this parameter represents the position at which the particle is when time is taken as 0. If this value is 0, then the particle will be exactly at its equilibrium state if represented by

a sine wave, and at its maximum displacement if done so by a cosine function (figure 2.1b). It is also measured in radians.

Using these parameters, a particle movement can be represented by the following equation:

$$x(t) = \sin(2\pi f \cdot t + \phi) \text{ [2.1]}$$

Thus, a particle movement x is described by a sinusoidal function (sine or cosine). The particle speed with which it returns to its initial position is described in this equation by its frequency. The higher this parameter, the more times the particle motion will go back to zero in a given time (figure 2.2). Finally, the phase constant indicates at which point of the cycle the particle starts its motion. (1) (2)

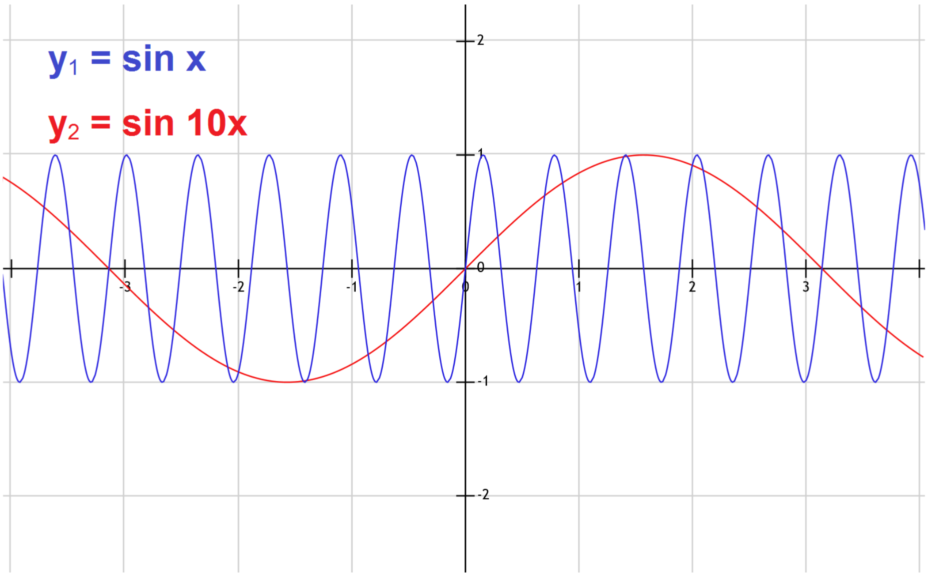

Figure 2.2. Frequency determines how fast the particle returns to the equilibrium position, in other words, how fast it vibrates. (Own production)

The human ear has evolved to be sensible to sound waves of frequencies between 20 Hz and 20 KHz. The frequency components that lie below the lower limit are called Infrasounds and those above the upper threshold are called ultrasound. Examples of the former include radio and infrared waves, which cross us by every day without us realizing it. An example of the later is the sound produced by bats and submarines to locate objects. (3) (4)

2.2.- Intensity of a sound wave

When a vibrating object, such as a violin string, forces surrounding air molecules to be compressed and expanded, it creates a pressure disturbance that travels from particle to particle through the medium, transporting energy as it moves. The energy that is carried by the disturbance was originally imparted to the medium by the vibrating string, and so the amount of energy that is transferred to the medium is dependent upon the amplitude of its vibrations. To illustrate this, let's use once again the analogy of a violin string. The harder a string is plucked, the more it will oscillate away from its rest position and thus the amplitude of the string vibration will be higher. In turn, the more it vibrates, the more energy it will impart to the medium. Subsequently, the amplitude of the vibration of the particles in the medium is increased, following the greater amount of energy being carried by the particles. (4)

The amount of energy that is transported past a given area in the medium per unit of time is known as the intensity of the sound wave (I). The greater the amplitude of vibrations of the particles in the medium, the more energy is transported through it, and the sound wave will therefore be more intense. Since the energy/time (E/t) ratio is equivalent to the quantity power (P_w), intensity is simply the power per unit area (A):

$$I = \frac{E}{t \cdot A} = \frac{P_W}{A} \quad [2.2]$$

Typical units for expressing the intensity of a sound wave are Watts/meter^2. (4)

As a sound wave carries its energy through a two-dimensional or three-dimensional medium, the intensity of the sound wave decreases with increasing distance from the source. The decrease in intensity with increasing distance is explained by the fact that the wave is spreading out over a spherical surface and thus the energy of the sound wave is being distributed over a greater surface area, as illustrated in figure 2.3. Since energy is conserved and the area through which this energy is transported is increasing, the intensity must decrease. (4) (1)

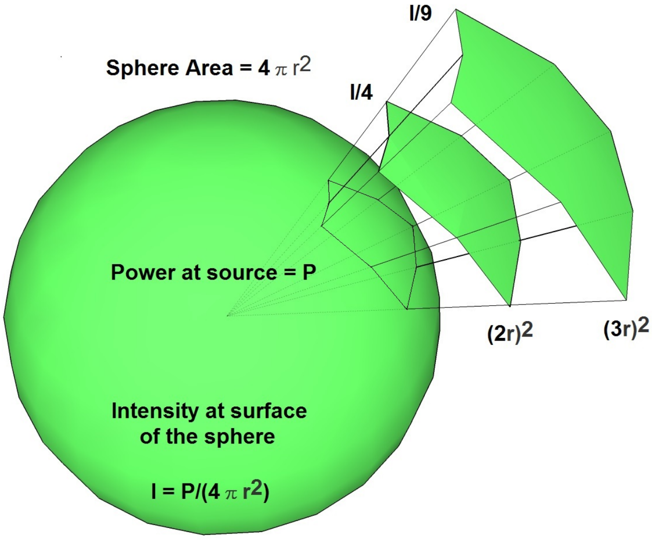

Figure 2.3. Sound intensity decreases as distance from the source increases because the same initial radiated energy is distributed through a larger area. (Own production)

The auditory system can cope with a wide range of sound intensities. Therefore, a logarithmic scale is best suited to describe the auditory frequency span, which expresses the ratio of two intensities. In this type of scale, Intensity I_0 is chosen as a reference and another intensity, I, is expressed in relation to it. The human ear is capable of hearing sounds at 10^{-12} W / m². This intensity is known as hearing threshold. When the intensity exceeds 1 W / m², the sensation becomes painful. By convention, the hearing threshold is used in this logarithmic scale as the reference level (I_0). The unit used is the decibel, which is represented as a function of sound intensity level (SIL), as:

$$SIL(dB) = 10 \log_{10} \frac{I}{I_0} \text{ [2.3]}$$

The factor of 10 multiplying the logarithm makes it decibels instead of Bels, because 1 decibel is around the minimum noticeable difference in sound intensity for the normal human ear. (5)

Another variable that can be used to describe sound intensity is the sound pressure level (SPL). Intensity and pressure can be related by the equation:

$$I = \frac{P^2}{\rho v} \text{ [2.4]}$$

Where P is sound pressure and is measure in Pascals (Pa), ρ is the density of air (1.2 kg/m^3 at 20^{o}C) and v is the speed of sound (331 m/s in air). Thus, SPL can be described as: (5) (6)

$$SPL(dB) = 20 \log_{10} \frac{P}{P_0} \text{ [2.5]}$$

While the intensity of a sound is a very objective quantity that can be measured with sensitive instrumentation, the loudness of a sound is a subjective response that will vary with a number of factors. The sound intensity must be factored by the ear's sensitivity to the particular frequencies contained in the sound and will vary slightly depending on the individual. But this will be discussed in more detail in further chapters. For now, it suffices to say that, since humans do not perceive low and high frequency sounds as well as they do perceive middle frequencies around 2 kHz, three weighing systems for SPL have been established: A, B and C. The graph in figure 2.4 describes the sensitivity of each weighing system as a function of frequency. Using the dB(A) measuring system, the sound level meter will be less sensitive to very high and very low frequencies. It approximates the ear when hearing a 1kHz sound at 40 dB. Measurements made with this scale are expressed as dB(A), which is very useful for eliminating inaudible low frequencies. The intermediate B-contour approximates the ear for medium loud sounds and it is rarely used. Finally, the dB (C) measuring system does not filter out the lows and highs to such an extent as the other contours. It approximates the ear at very high sound levels and has been used for traffic noise surveys in noisy areas. (6)

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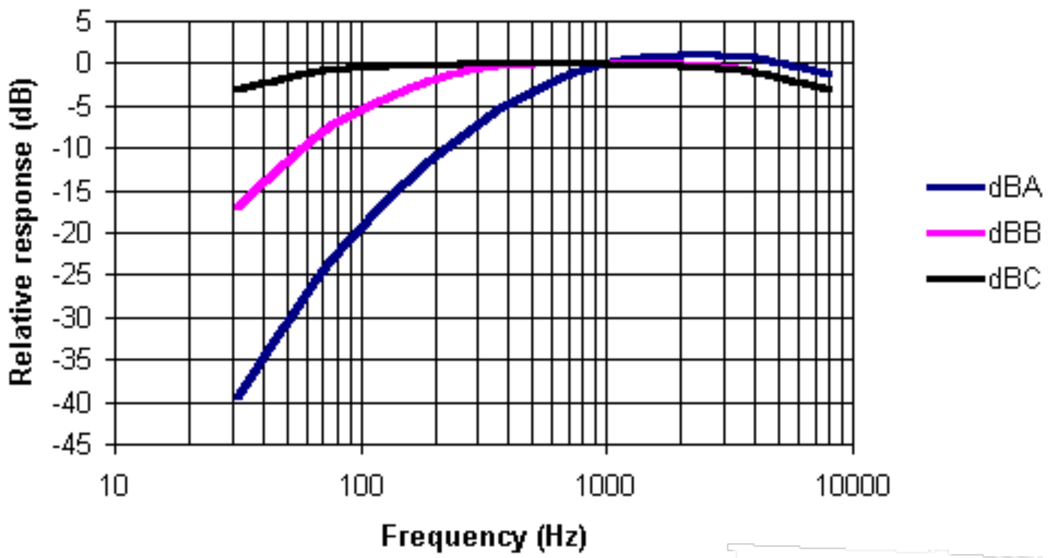

Figure 2.4. Relative response of each weighing SPL system as a function of frequency. It can be seen how dBA is most sensitive to frequencies up to 1000 Hz, while dBB and dBC tend to flatten out sooner. (Own production)

2.3.- Fourier Analysis of a sound wave

All sounds can be decomposed into superpositions of sinusoids of different frequencies, amplitudes and phases or, seen from the opposite perspective, simple sinusoids can combine to form any kind of complex sound wave. Figure 2.5 illustrates this principle. The decomposition is known as Fourier analysis and gives back the composition of the sound wave f(t) in term of its pure tones or frequency composition and the intensity with which they appear. This translates mathematically into a sum of cosines and sines of different frequencies:

$$f(t) = A_0 + \sum_{n=1}^{\infty} A_n \sin(2\pi \cdot n(2\pi f t)) + \sum_{n=1}^{\infty} B_n \cos(2\pi \cdot n(2\pi f t)) \quad [2.6]$$

Where the expressions in the sums are called the Fourier coefficients, A_0 is a constant that describes the average value of the function (i.e. the point around which it oscillates up and down) and A_n and B_n are the amplitudes of each frequency or tone component. (7) (8)

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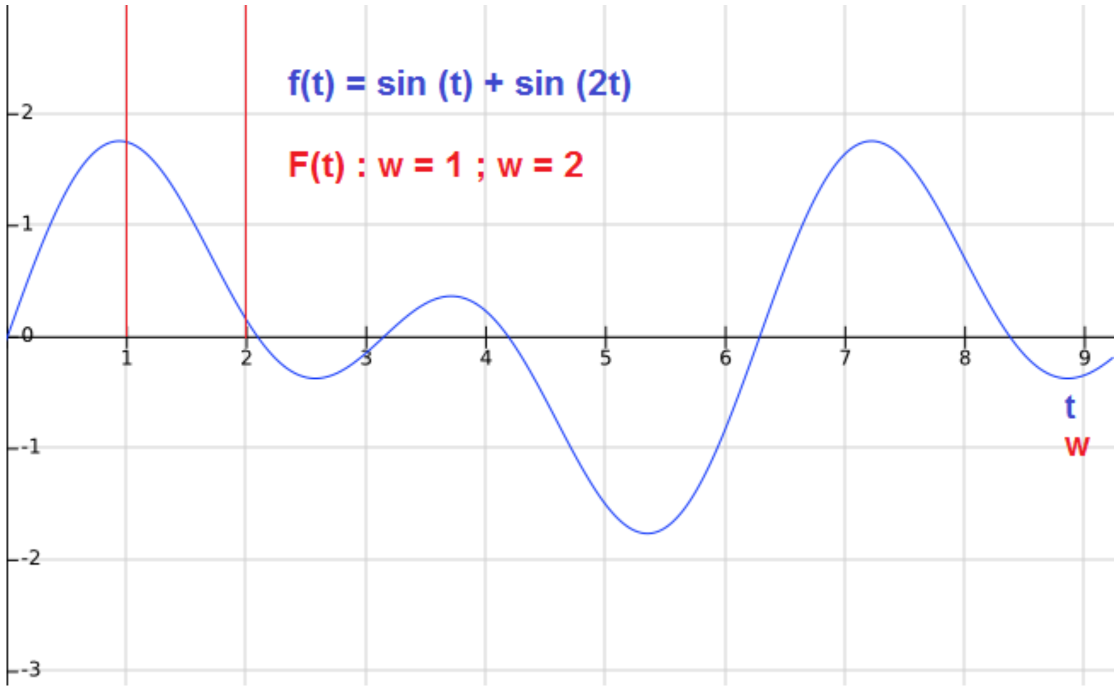

Figure 2.5. When two single frequency waves are combined, the result is a waveform that is the sum of them. (Own production)

The transformation of a complex wave pattern in time into its frequency component distribution is known as Fourier transformation and it constitutes a basic tool in the field of acoustics, because it allows the scientist to characterise sounds. A Fourier transform shows that any wave can be decomposed into a sum of sinusoids.

It is not the aim of this section to describe the mathematical background behind a Fourier transform, which is complex and rarely used analytically. Most real life waveforms are so complex that Fourier transforms are carried out by computers. These normally use an algorithm called the Fast Fourier Transform or FFT. Given a waveform function in the time domain (like that in figure 2.5, blue line) an FFT would give its profile in the frequency domain (figure 2.5 red lines). (8)

These illustrations show the essential nature of the FFT. For a sine wave with a single frequency, the FFT consists of a single peak. Combining two sound waves produces a complex pattern in the time domain, but the FFT clearly shows it as consisting almost entirely of two frequencies. Thus, Fourier

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analysis can greatly simplify acoustic problems and reveal aspects that are not clearly visible in the time domain. (8)

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3.- THE PHYSICS AND ANATOMY OF THE EAR

The ear is the most complex transducer ever created. It is able to convert energy from sound pressure into mechanical movement and this, in turn, into electrical nerve impulses that travel all the way to the brain. The ear's ability to carry out this process allows us to perceive the details of a complex sound. The ear detects the frequencies that make them up; each one of their individual intensities and, through the integration of both aspects, it can recognise features such as who is talking and what a person is saying even when noise is present.

The ear has three main parts: outer, middle and inner ear (figure 3.1). Each part has a specific function in the process of detection and interpretation of sound. The first serves to pick up sound and prepares it to transfer the energy to the middle ear in the most efficient way, by matching air and bone impedance. The second part is responsible for transforming the energy of the sound wave (pressure variations) into mechanical displacements of the ossicles, which creates a compressive wave within the inner ear. The inner ear is filled with fluid, which bends hair cells cilia as it moves and in turn transforms this mechanical displacement into electrical impulses that are identifiable by the brain.

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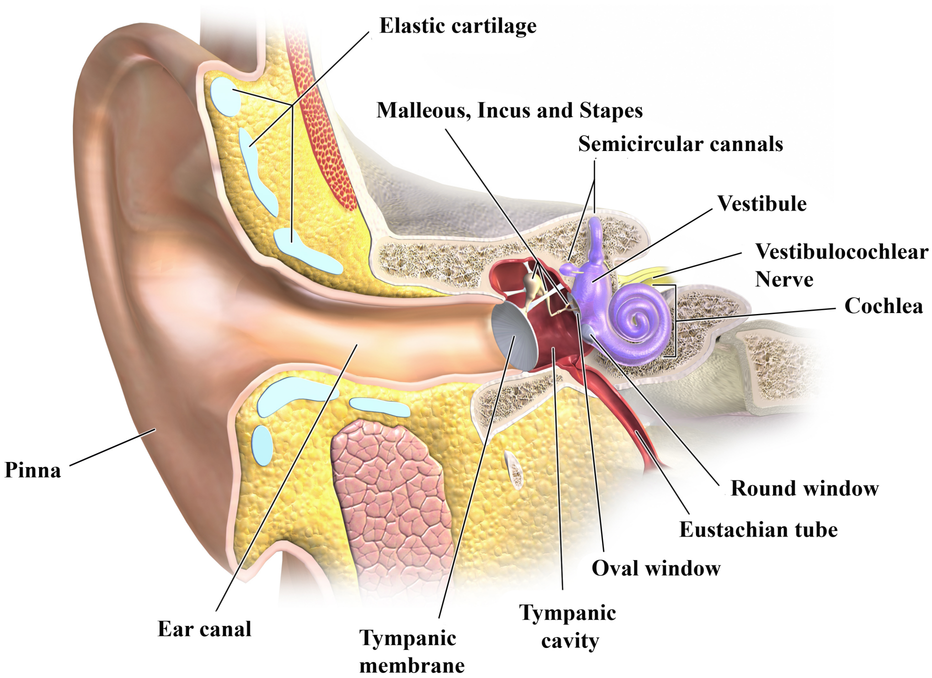

Figure 3.1. Anatomy of the ear. The pinna and external auditory canal form the outer ear, which is separated from the middle ear by the tympanic membrane. The middle ear houses three ossicles, the malleus, incus and stapes. Together they form the sound conducting mechanism. The inner ear consists of the cochlea, which transduces vibration to a nervous impulse and the vestibular labyrinth that houses the organ of balance. (Own production using a free license image from BruceBlaus)

3.1.- The outer ear

The outer ear consists of the ear and ear canal. Its overall function is to act as a pre-amplifier to improve the sensitivity of the system and its action mechanisms are a direct result of the physical characteristics of sound. (9)

The sound energy spreads spherically from the source. For a point source of sound, as the radii of the spheres of sound propagation get further away, they also become larger. Given that the surface of a sphere is proportional to the square of its radius ($A = 4\pi r^2$), the energy in the sound wave arriving at a small area (e.g. an ear drum) declines with the square of the distance from the source ($I \propto 1/r^2$)). This relationship between sound intensity or energy and distance from the source is known as the inverse square law. (9)

Therefore, for a given sound intensity, a larger pinna will have a broader area and thus will capture sound more efficiently. The ear canal, on the other hand, acts as a broadband resonator tube, closed at one end. It has an average diameter of 0.7 cm and is 2.7 cm long (L). By acting as a tube closed at one end as in figure 3.2, it gives rise to standing waves at a fundamental resonant frequency that is four times its length and at odd multiples of it, called harmonics. The closed end is restricted to be a wave node and the open end an antinode. As seen in figure 3.2, this condition yields a wavelength of the fundamental mode that is four times the length of the air column inside the canal (10.8 cm). The restriction of the closed end prevents it from producing even harmonics. Since ($\lambda = v/f$ and $\lambda = 4L$), its resonant frequencies are given by the equation:

$$f_n = \frac{nv}{4L}; n = 1,3,5 \ldots \quad [3.1]$$

So the first resonant frequency will be around 3 kHz. (9) (10) (11)

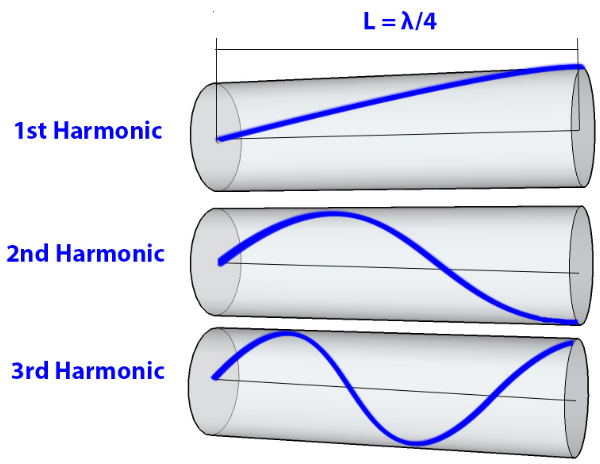

Figure 3.2. The ear canal acts as a tube open at one end, which forces the closed end to be a node and the open end and antinode. This causes the minimum resonant frequency (1st harmonic) to be four times the length of the ear canal as. The next harmonic that follows these boundary conditions is an odd multiple of this resonant frequency. (Own production)

3.2.- The middle ear

The main parts of the middle ear are the tympanic membrane and the three ossicles (malleus, incus and stapes). Together they amplify the sound wave

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and perform the important function of impedance matching. This improves the efficiency of the transfer of the mechanical vibrations of the tympanic membrane (~ low mechanical impedance) to the oval window (~ 20 times higher impedance). (11)

The tympanic membrane is a circular membrane of about 8-9 mm in diameter, 0.1 mm in thickness, with an area of 65 to 80 mm^2 and a weight of 14mg. It has a conical shape and is attached to the inner part of the stapes. It establishes the boundary between the outer and middle ear, acting just like the membrane of a microphone. It converts the air pressure waves into mechanical vibrations that are transmitted to the ossicles. The tympanic membrane has the peculiar property of being a damped resonator (the system goes back very fast to its resting position) due to the high tension force to which it is subjected. This broadens the number of frequencies that can cause the system to resonate and it therefore has a large range that goes from 20 to 20,000 Hz. Because it is 20 times the size of the oval window (the membrane that connects the middle ear with the inner ear fluid cavity), it amplifies the incoming sound by a factor of 20, compared with the direct reception of the sound by the oval window alone. (12)

The auditory receptors of the inner ear operate in a fluid environment, and the inner ear is really an "underwater" sound receiver. When sound in air strikes a fluid boundary (a boundary between media with different acoustic impedances) there is a theoretical loss of 99.9% of the energy in a sound wave in air due to reflection. This 99.9% loss is equivalent to 30 dB. In order to overcome this mismatch in the impedance between air and fluid, the middle ear is interposed between the tympanic membrane and the oval window. (13)

Ossicles work like a composite lever that achieves a multiplication of force between the tympanic membrane and the oval window. Although they are among the smallest bones in the body they are capable of spreading force with great pressure because of their physical structure and layout within the inner ear. When these three bones work together in combination with two small muscles, they form the ossicle chain. This chain transfers the force coming from the tympanic membrane (effective area of of 65 mm^2) to the oval window (effective area of 3.2 mm^2), thus increasing the pressure with which this force is applied. Since the pressure is the force per unit area ($P = F/A$), the relationship between the pressure in the tympanic membrane and the oval window will be:

$F = P_t \cdot A_t$ (Tympanic pressure)

$$P_o = \frac{P_t \cdot A_t}{A_o} \quad [3.2]$$

$F = P_o \cdot A_o$ (Oval window pressure)

The factor A_t/A_o is approximately 20. It also happens that the lever arm formed by the malleus in rotating about its pivot is somewhat longer than that of the incus, giving another factor of about 1.3 in pressure increase. These two factors multiplied, yield about a 26x increase in pressure, which is about 29 dB, thus just about overcoming the theoretical 30 dB loss due to the air/liquid interface. Therefore, the middle ear matches the acoustic impedance between air and fluid, maximizing the flow of energy from the air to the fluid of the inner ear. (9)

3.3.- The inner ear

The inner ear consists of a bonny labyrinth, divided into the vestibular system and the cochlea. It is a sealed cavity filled with a fluid called lymph. The cochlea is the organ in charge of sound transduction into electrical signals. The cochlea is spirally wounded up, forming two and a half turns, with a length of about 30 mm and a variable sectional area that ranges from 4 mm^2 at its basal end, to 1 mm^2 at its apical end. The cochlear cavity is divided by two membranes: the Basilar and Reissner membranes. Both divide it longitudinally into three channels. Figure 3.3 shows its main features. They connect at the apex of the cochlea in a region known as Helicotrema. The Reissner membrane separates the Scala Vestibuli from a smaller cavity known as the cochlear duct. This, in turn, is separated by the basilar membrane (BM) from a third cavity called Scala Timpani. The BM is composed of a great number of taut, radially parallel fibres sealed between a gelatinous material of very weak shear strength. These fibres are resonant at progressively lower frequencies as one progresses from the basal to the apical ends of the cochlea. Four rows of hair cells lie on top of the BM, together with supporting cells. A single inner row is medial, closest to the central core of the cochlea. It has an abundant nerve supply carrying messages to the brain. The three outer rows receive mainly an afferent nerve supply. The assembly formed by the BM, the tectorial membrane, the hair cells and the nerve terminals is called the organ of Corti. Any natural displacement of the cochlear partition results in a rocking motion of these structures and consequently a lateral displacement of the inner hair cells. This movement causes depolarization and hiperpolarization of the hair cells and transmits the nerve impulse to the neural ends that go along the centre of the cochlea or modiolus forming a structure known as the spiral ganglion. This complex is the heart of the ear, as it converts mechanical vibrations into electrical currents than can be understood by the brain. (10)

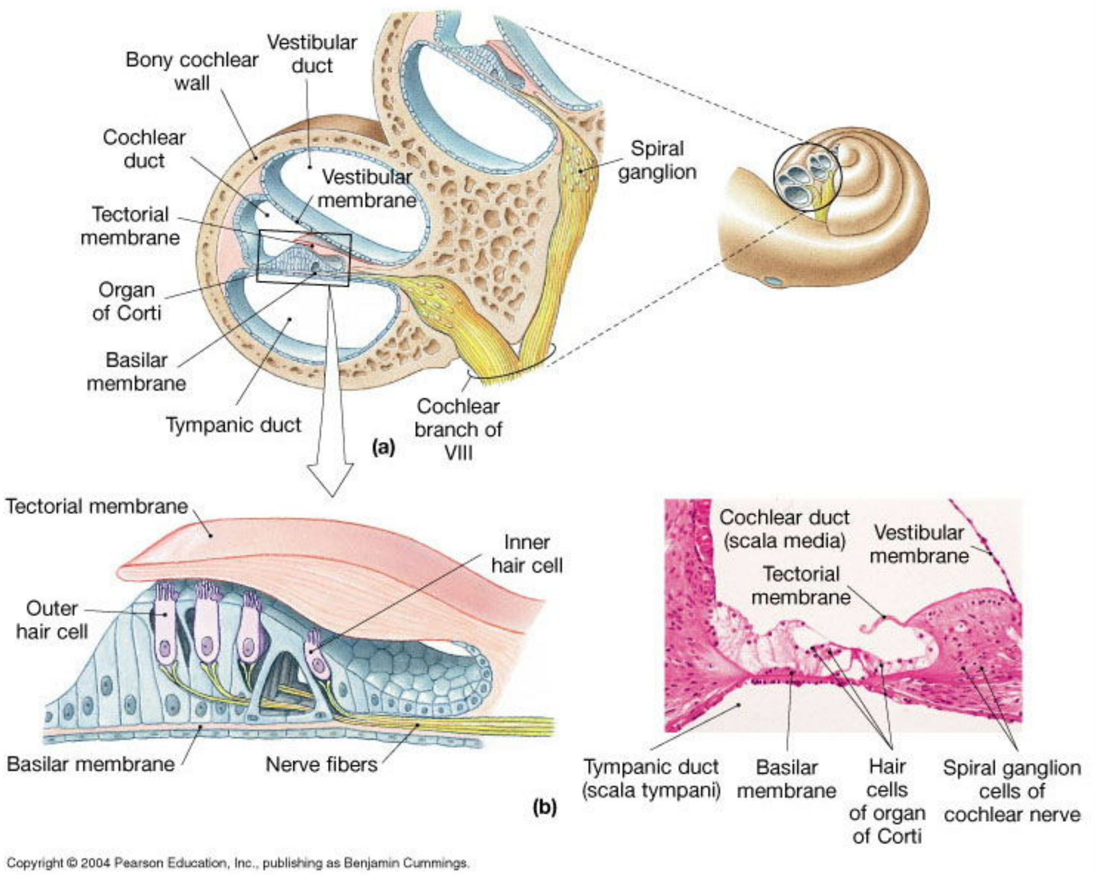

Figure 3.3. Anatomy of the cochlea with a close-up diagram (a) and a real histological image (b) of the organ of Corti, the three Scalae and the spiral ganglion (14).

Georg Von Békésy, Nobel Prize in Medicine in 1961, first described the mechanism by which the cochlea breaks down complex sounds into their frequency components. In his "Theory of Location", he discovered that pressing the stapes on the oval window causes an overpressure in the base of the cochlea that forces the lymph fluid to circulate to the lower cavity through the helicotrema. This causes the BM to deform downwards and an elastic membrane below the oval window known as the round window, to bulge outwards. If the stapes moves from left to right sinusoidally with a frequency n, the effect is the appearance of a traveling wave in the BM that moves from base to apex. The forward speed of the wave depends on its frequency and on the width and stiffness of the BM. Since the latter is narrower and more rigid in the base and wider and more flexible in the apical end, the result is that at some point in the cochlea the wave velocity for a given frequency becomes zero. At that point, the wave has its maximum amplitude of oscillation and consequently will cause maximum deflection of its hair cells (figure 3.4). Thus Bekesy came to the conclusion that each point of the BM responds to a certain frequency. Consequently, high frequencies are detected in the basal end and the low frequencies in the apical end, making the cochlea a remarkably efficient frequency analyser. (15)

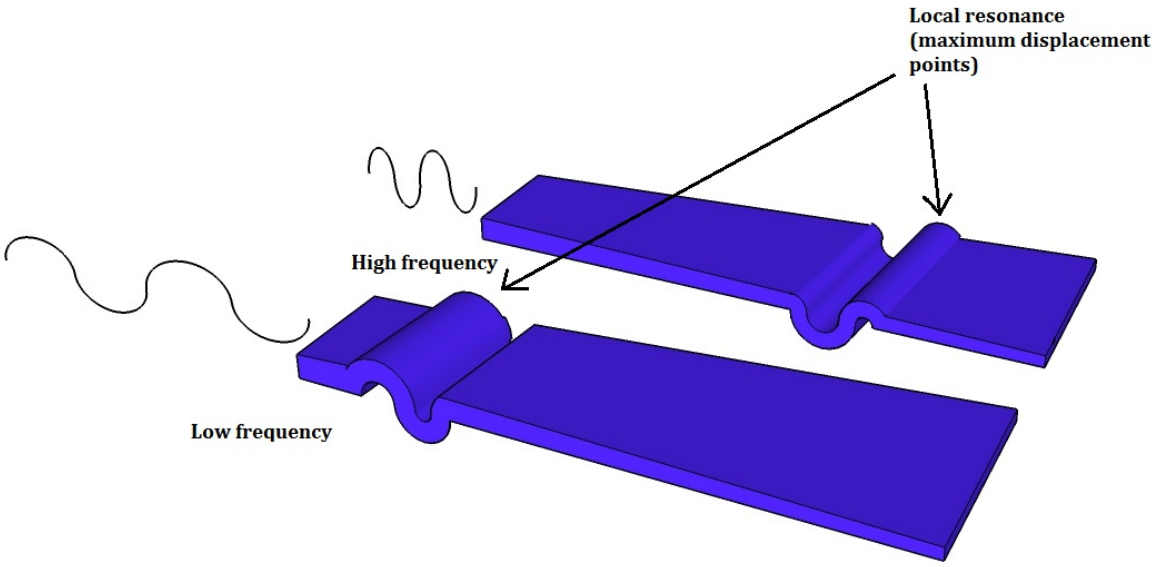

Figure 3.4. Basilar membrane vibration for high and low frequencies. The maximum displacement point along this membrane can be seen to be frequency dependent, thus making the cochlea a fine tuned frequency analyser. (own production)

When the BM moves relative to the tectorial membrane in which the hair cells are embedded, their stereocilia bend. The distortion of stereocilia causes potassium channels to either open or close. Because the concentration of K^+ in endolymph is higher than that in the cell, the opening of K^+ channels causes the cell to depolarize as K^+ rushes in. The closing of these channels causes hyperpolarization. The stereocilia are arranged in decreasing sizes and are connected by elastic protein filaments. When the hair cell is at rest, potassium channels are partially open and partial depolarization causes Ca^{++} channels to open. This results in the release of neurotransmitters by the hair cell, which cause neurons of the cochlear nerve to fire action potentials. When stereocilia bend in the direction of the Kinocillium the increased tension exerted on them opens the potassium channels more, augmenting depolarization. This increases Ca^{++} entry and neurotransmitter release and

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the frequency of action potentials in afferent neurons. When stereocilia bend away from the Kinocillium potassium channels close with opposite results. This process is illustrated in figure 3.5. (16)

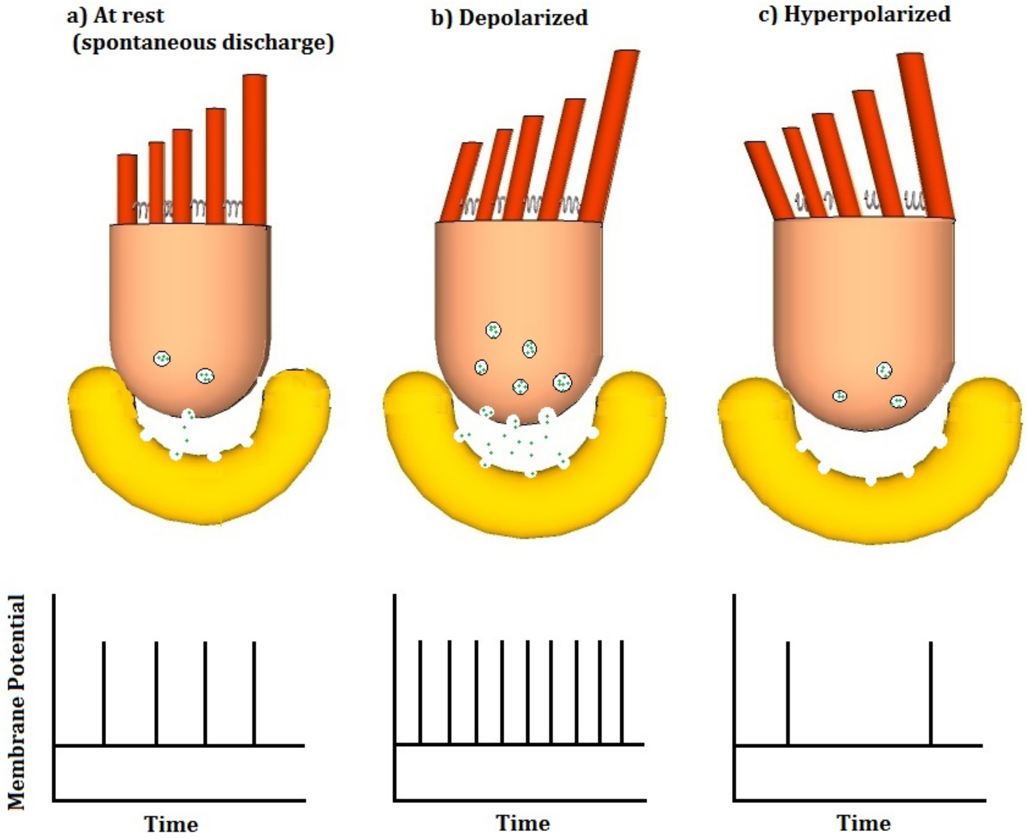

Figure 3.5. Hair cells have stereocilia arranged in a directional way, which causes hiperpolarization of the cell when bent in one direction, and depolarization when bent the opposite way. These changes release neurotransmitters that are received by the neural ends and sent to the brain as auditory input. (Own production)

The cochlea has an abundant nerve supply both of fibres taking impulses from the cochlea to the brain (afferent pathways) and of fibres bringing impulses from the brain to the cochlea (efferent pathways). As with virtually all neural-mechanisms, there is an active feedback loop. Nerve fibres can fire at a maximum rate of 200 times per second. Sound information is consequently conveyed to the brain at a maximum rate of 200 pulses per second (up to 200 Hz). They fire in locked phase (that is, at a fixed point every x wave cycles, see figure 3.6) with acoustic signals up to about 5 kHz. Both firing patterns are known as temporal hearing. Above this value, frequency information is conveyed to the brain based upon the place of stimulation on the BM, because the refractory period of neurons (the period when they are recharging), is slower than the firing rate. This is called spatial hearing. (16) (17)

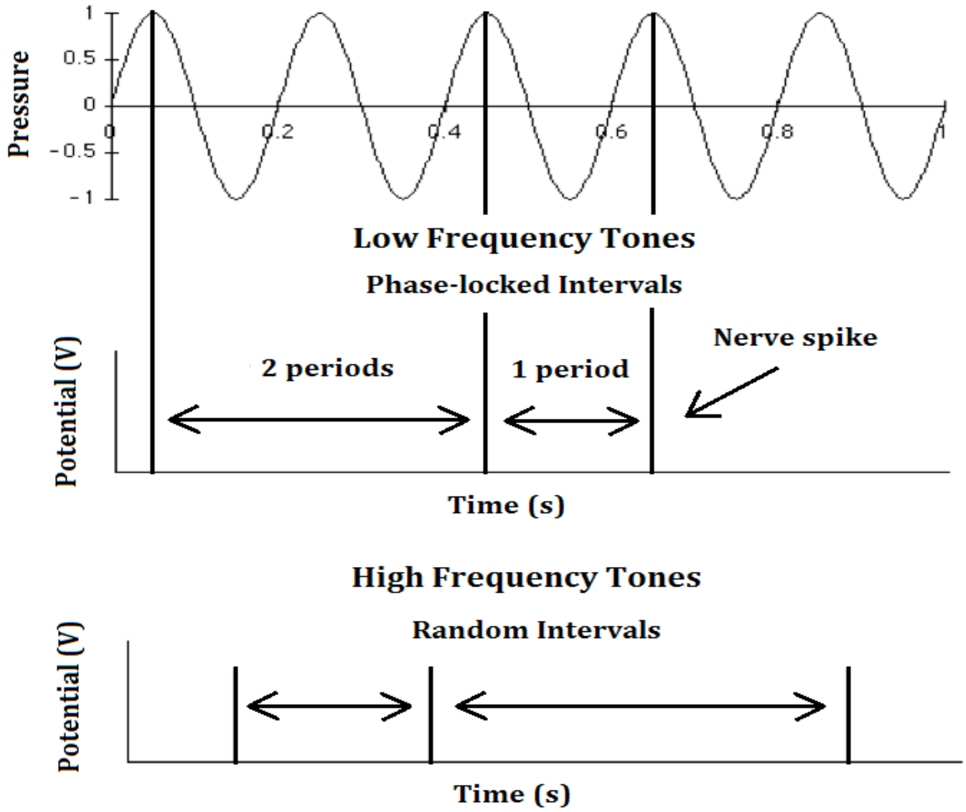

Figure 3.6. The auditory nerve will tend to fire at a particular phase of a stimulating low-frequency tone. So the inter-spike intervals tend to occur at integer multiples of the period of the tone. With high frequency tones (> 5kHz) phase locking is not possible, because the capacitance of inner hair cells prevents them from changing voltage sufficiently rapidly. (Own production)

Therefore, when the ear perceives a complex sound, each of its frequencies excites a point on the BM. As a result, the brain can interpret not only the intensity of the sound, but also its timbre based on the identification of which nerve ending was excited and how intensely. Nature has in fact developed the most exquisite frequency and intensity analyser that engineers could ever dream of.

4.- PSYCHOPHYSICS

The sensors that evolution has provided us with are highly specialized in increasing signals that have the potential to help us survive, but also in ignoring those that might distract us. The field of psychophysics investigates the relationship between physical stimuli and the perceptions or feelings that these evoke in us. In normal-hearing (NH) individuals, psychophysics mainly relates acoustics and mechanical vibrations to the psychology of perception. In the case of cochlear-implanted patients, psychophysics involves the study of perceptions and responses associated with different patterns and intensities of electrical stimulation. This allows researchers and clinicians to examine how electrical stimulation of neural fibres can be best used to convey auditory information, the main objective of cochlear implants. Psychophysical variables are useful for understanding the perceptions produced by different electrical stimuli and provide the researcher with factual information on how the device is working. This understanding is essential to further improve cochlear implants and speech processors technology.

There are several psychophysical variables that facilitate the description of the patient's ability to resolve different stimuli and even say whether this was done in either the temporal or spectral domain. The two most basic psychophysical variables measured in normal hearing and in cochlear-implanted individuals are: 1) threshold of hearing (T level in cochlear-implanted patients): 2) maximum level of comfort (C level), which is the intensity at which a stimulus is perceptible but not uncomfortable. Doctors use them to adjust the dynamic range for each electrode in the directory. In some cases, electrode stimulation may produce perceptions uncomfortable at all possible intensities, in which case the electrode is disabled.

There are more complex psychophysical variables that provide additional information on the perceptions associated with different stimuli. Several studies have measured forward masking, which is the degree to which a stimulus probe is masked by a previous stimulus. Another psychophysical variable in the time domain is the detection of amplitude modulation, which measures a patient's ability to discriminate between a constant amplitude stimulus and one where this is modulated. (18) (19)

Another crucial parameter in cochlear implants is electrode discrimination. The theory that each electrode provides a unique perception was evaluated using a reference electrode and searching for the first distinguishable electrode that can be perceived. (20) (21)

Two common psychophysical intensity variables in cochlear-implanted (CI) patients are: just noticeable difference (JNDs) and loudness increase. The dynamic range of each electrode can be divided into steps of discriminable intensity. Several studies have measured JNDs to quantify the intensity resolution for each electrode (22) (23). Loudness is a subjective term describing the strength of the ear's perception of a sound, so the interest lies in correlating increases in intensity with perceived increases in this parameter. (24) (19) (25)

4.1.- Pitch perception

The American National Standard Institute defines pitch as *"that attribute of auditory sensation in terms of which sounds may be ordered on a scale ranging from low to high. Pitch depends primarily on the frequency content of the sound stimulus, but it also depends on the sound pressure and the waveform of the stimulus "*(ANSI 1994). The inclusion a sound physical characteristics is too vaguely described and thus makes this property an attribute of sensation rather than the result of a single physical quantity. However, it is important to note that, with the years, the definitions of pitch have moved towards more physics oriented terms. The first definition did not include any physical properties of sound, describing it as: "*that attribute of auditory sensation in terms of which sounds may be ordered on a scale ranging from low to high"* (ANSI 1973). As our understanding of the mechanics of nature expands, so does the presence of physics in every field of science.

With the exception of pure tones, pitch is not a simple function of the spectral content of a sound. Rather, pitch is related more closely to the repetition rate, or envelop repetition rate, of the sound with a range from 30 Hz to 5 kHz. Sounds with the same repetition rate and very different spectra often have the same pitch (e.g. a pure tone with a frequency of 500 Hz and a complex tone with high harmonics with a fundamental or first resonant frequency, F0 of 500 Hz). On the contrary, sounds with similar spectra can have very different pitches (e.g white noise with amplitude modulated at 100 or 200 Hz). Thus, the auditory system combines information across the cochlear location (frequencies and intensity), to derive the pitch of the sound. Moreover, pitch is also represented in terms of the precise timing of neural impulses in the auditory nerve and at higher centres in the auditory systems.

Many sounds have acoustic waveforms that repeat themselves over time. These sounds are often perceived as having a pitch that corresponds to the repetition rate of the sound. This attribute of sound is one of the most relevant perceptual dimensions for speech communication, carrying prosodic information in western languages, but also semantic information in tonal languages such as Mandarin. Moreover, it enhances our ability to perceptually segregate sound sources, based on differences in fundamental frequency (F0). Conversely, it can be used to group together the individual sound components (harmonics) that arise from the same vibrating source. Pitch, therefore, defines and differentiates our acoustic environment.

Theories of pitch perception

There are currently two theories of pitch perception (see figure 4.1). The spatial theory is based on two main tenets. The first is that the stimulus is subjected to spectral analysis in the cochlea; so each frequency excites different locations along the basilar membrane and, consequently, neurons with different central frequencies. This process is called tonotopic organization. The second tenet states that the perception of a stimulus is related to the pattern of excitation that it produces. For a pure tone, it is assumed that this corresponds to the place of maximum excitation. The first of these principles has been confirmed in a series of independent experiments, including direct observation of the movement of the MB. The second is still in dispute. (19) (15)

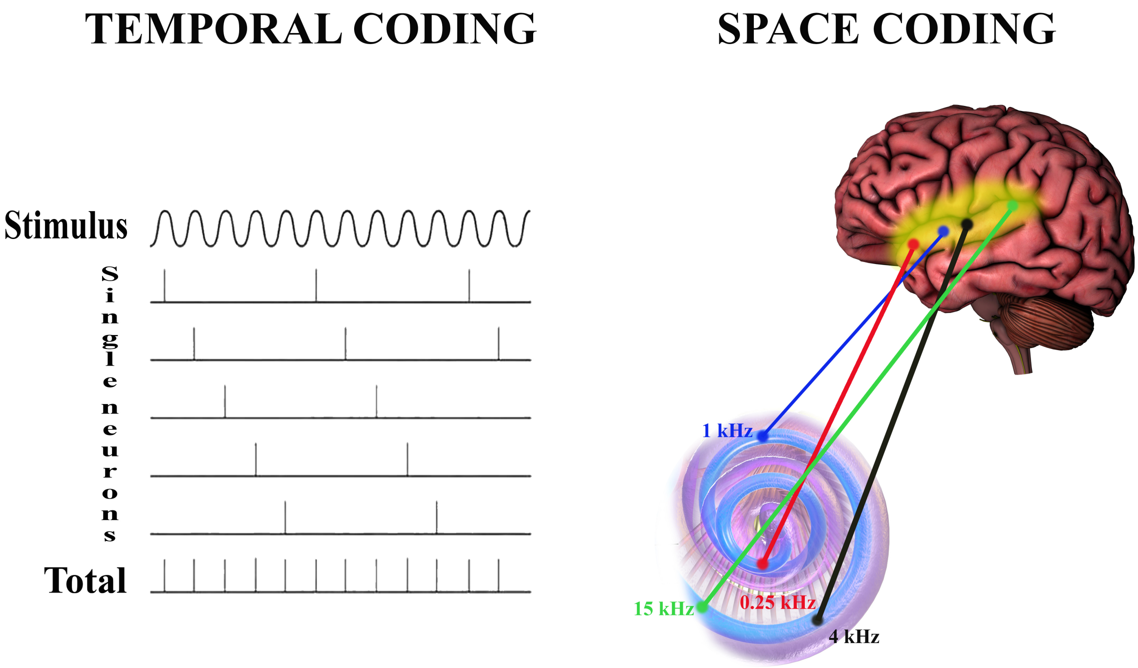

Figure 4.1. Temporal and place coding of sound. In temporal coding (left), neurons fire action potentials in phase with the sound waves. Place coding (right) refers to the perception of pitch depending on the site of stimulation. This is possible because the auditory cortex that responds to sound is arranged tonotopically. (Own production)

An alternative theory is the temporal theory, based on the assumption that the tone of a stimulus is related to its evoked neuronal firing time pattern. The peaks of neural firing tend to occur at a particular phase of the waveform in the MB (the cells phase-lock to stimuli). This causes the time intervals between successive peaks to approach integral multiples of the waveform period. Phase-locking becomes very weak for sinusoids with frequencies above approximately 5 kHz, as stated in the previous chapter, although the precise upper limit in humans is not known. However, the tones produced by musical instruments, the human voice, and the most common sources of sound have fundamental frequencies below 5 kHz. (Moore 2013)

Physiological measurements and simulations using computational models have demonstrated that the repetition rates of stimuli are very well represented by the pattern of phase locking in the auditory nerve. For many researchers, the time versus place argument has been replaced by the question of how and where in the auditory pathway the phase-locked activity is analysed. The maximum frequency to which a fibre will phase lock declines from the auditory nerve to the auditory cortex and it is thought that somewhere in the brainstem, synchrony representation is converted into a rate-place representation in which different neurons code for different pitches in terms of overall firing rate. (26) (27)

However, there remain some fundamental questions to be answered conclusively, such as how phase-locked neural activity is transformed into a rate-place representation of pitch; where this transformation takes place and how this information is used in object and pattern identification.

4.2.- Loudness perception

Loudness is the subjective intensity of a sound. It is subjective because it depends entirely on the individual. The term intensity is used here because the response indicates how strong the sound seems to the listener. This definition is vague, but in experiments on loudness it usually suffices to elicit consistent responses. These responses can then be studied to establish a relation between loudness and the experimentally manipulated physical variables. Sound intensity is the most important of them in determining loudness, but spectral variables such as signal frequency and bandwidth also have an effect on it. The duration and intermittency of a sound are among the significant temporal variables. Moreover, background sounds can also affect the perception of loudness. Therefore, sound intensity must be factored by the ear's sensitivity to the particular frequencies, and their durations, contained in the sound. As a complete review of all the physical parameters that affect loudness perception would require a level of detail that is beyond the scope of this dissertation, just the main variables will be described in detail here. For a more detailed description, the author refers the reader to. (19) (28)

Loudness level

Loudness level indicates how loud a 1 kHz tone must be in order to sound as loud as a test tone. To determine the level of a given sound, the subject is asked to adjust the level of a 1 kHz tone until it has the same loudness as the test sound. They are presented alternatively rather than simultaneously. The resultant level of the 1 kHz tone is the loudness level of that tone and is given in phons. If instead the 1 kHz tone is fixed in level, and test sounds at different frequencies are adjusted to give a loudness match, an equal-loudness contour is generated (figure 4.2). These standard equal loudness contours (ISO 226, 2003), are based on extensive measurements of several laboratories, as described by Suzuki and Takeshima (29) . This curve shows how strong each frequency sound must sound to equal the corresponding phon line. For example, if the line is at 30 dB SPL, a 125 Hz sound must have an SPL of 50 dB to sound as loud as a 1 kHz sound at 30 dB SPL. These contours tend to flatten at high loudness levels. This means that the rate of growth differs for tones of different frequency. The rate of growth is greater for low and very high frequencies than for middle ones. (30) (29)

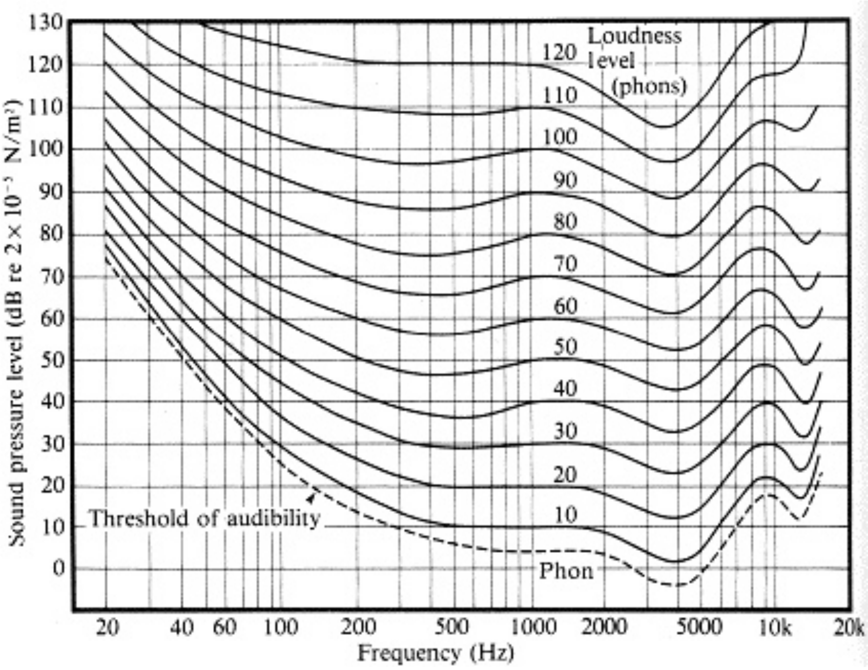

Figure 4.2. Equal-loudness contours for loudness levels from 10 to 120 phones for sounds presented binaurally from the frontal direction. The absolute threshold curve (MAF) is also shown. (19)

This is the basis behind the weighting systems seen in previous chapters. The A weighting is based roughly on the 30-phon equal-loudness contour. At high levels, the weighting becomes more linear, so the C weighting is used. The B weighting is used for intermediate levels, and it is based on the 70-phon equal-loudness contour. (19)

Loudness models

S.S Stevens was the pioneer in the development of scales of loudness and other sensory dimensions. He suggested that perceived loudness, L, was a power function of physical intensity I:

$$L = kI^{0.3} [4.1]$$

Where k is a constant that depends on the subject and the units used. According to this equation, given two sounds where one is 10 times more

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intense than the other in the physical domain, the perceived loudness is only two times higher ($10^{0.3} \approx 2$) which implies the existence of compression. (31)

The power law relationship between intensity and loudness has been confirmed in a large number of experiments using a variety of techniques. However, there have also been criticisms. The techniques are very susceptible to bias effects like: 1) the range of stimuli presented; 2) the first stimulus presented; 3) the instructions to the subject; 4) the range of permissible responses; 5) the symmetry of the response range and 6) other factors related to experience, motivation, training and attention. (32)

Unfortunately, the underlying mechanisms of loudness perception are not yet fully understood. A common assumption is that the intensity is in some way related to the overall neuronal activity evoked by a sound. In this scenario, the intensity of a sinusoidal tone would be determined not only by the activity of neurons with characteristic frequencies (CF) near the tone frequency, but also by the degree of activity of adjacent CF neurons. In other words, the volume may depend on a sum of neuronal activity through different frequency channels (known as critical bands).

Models incorporating this basic concept have been proposed by Fletcher and Munson, Zwicker and Scharf and, more recently, by Moore et al. The models proposed by Moore are illustrated in figure 4.3. The first stage is a fixed filter to simulate the transmission of sound through the outer and middle ear. The next step is to calculate a pattern of excitation for sound studio. Then, there is a transformation from the excitation level (in dB) to the specific loudness, which is a kind of "sound density" that represents the volume per critical band. This transformation involves a non-linear compression; for example, a 10-fold increase in the sound intensity, which corresponds to an increase of 10 dB, causes less than a 10-fold change in specific loudness. Although the models are based on psychoacoustic data, this transformation can be understood as representing the way the physical excitation is transformed into neural activity. The specific loudness is probably related to the amount of neural activity in

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the corresponding CF. The compression in the transformation partly reflects what occurs in the basilar membrane. The overall volume of a given sound, in sones, is assumed to be proportional to the total area under the specific pattern of loudness. One could think of this area as approximately proportional to the total neuronal activity evoked by a sound. Sound models of this type have been quite successful at explaining the experimental data on the sonority of simple and complex sounds. (30) (33) (34)

Figure 4.3. Basic structure of Moore et al. loudness model. (Own production)

5.- COCHLEAR IMPLANTS

Cochlear implants (CI) are devices designed to restore hearing to profoundly deaf people. Hearing aids are a common solution for people with moderate hearing thresholds; however, for higher thresholds (~ 90 dB), these do not work as successfully. In these situations, a cochlear implant is used to bypass the middle and inner ear and directly stimulate the auditory nerve from within the cochlea.

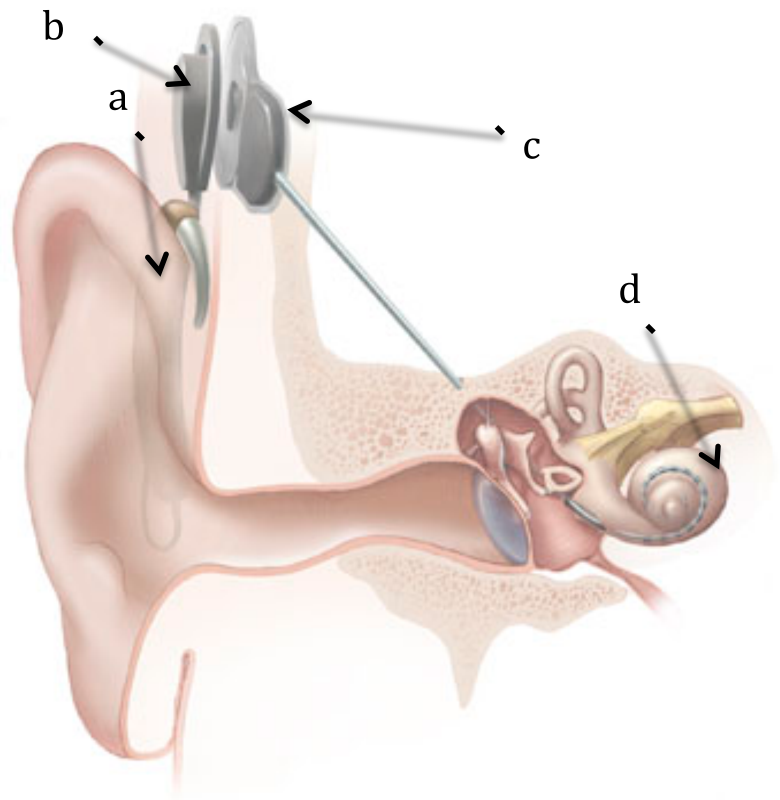

Figure 5.1. Main components of a cochlear implant: a) speech processor; b) transmitter; c) receiver/stimulator; d) electrode array. (Own production using free image from National Institute on Deafness and Other Communication Disorders at the National Institutes of Health)

Figure 5.1 shows the main parts of a cochlear implant. Most modern systems include four components: a portable external sound processor, a transmitter, an implanted receiver/stimulator and a set of intracochlear electrodes. The sound processor consists of a microphone that collects the sound signal; an electronic processor, generally in the form of a digital signal processor (DSP), that encodes it into electrical pulses. The signal then goes to the radio frequency transmitter, which sends it to the transcutaneous receiver / stimulator through a transmitter coil. This transmission is done through inductance (magnetic field variations), which induces alternating current pulses in the receiver coil. The receiver / stimulator is a surgically implanted electronic device. It receives and decodes signals from the sound processor and generates electrical signals to selectively activate the intracochlear electrodes. Small electric current pulses are sent to these electrodes to stimulate nerve cells in the peripheral auditory system, thereby causing auditory sensations. The electrodes are placed in the form of an array, which is made up of 16 to 24 contacts (depending on the device). They are 12.1 to 27 mm away from each other depending on the manufacturer. Each electrode stimulates a different site in the cochlea, which corresponds to a frequency band matching the cochlear tonotopy. Thus, mainly a limited population of nerve fibres perceives the stimulus. This allows different sections of the cochlea to be independently stimulated, mimicking and bypassing the basilar membrane frequency filtering function. (35) (36)

Sound is filtered in the speech processor by a bank of bandpass filters that do not overlap, creating a set of channels that contain information from different regions of the sound frequency spectrum. Each of these channels is able to send information to one electrode of the array. The resolution of a healthy human auditory system is 1 Hz in the 100 Hz region. This would be the equivalent of having 100 channels from 100 to 200 Hz. Instead, cochlear implants have 16 to 24 channels for the whole audible frequency range. This number depends on the processing algorithm of each specific cochlear implant manufacturer. (37)

5.1- Signal processing

The most important CI unit is the signal processing strategy used for the transformation of the electrical stimulation signal. The different signal processing techniques that have been developed can be classified according to the sound stimuli information that they use: 1) the waveform, 2) the envelope, and 3) the spectral characteristics (fine structure) of the signal.

The latest speech processors are based on the channel vocoder principle, which is used in telephone communication with much less bandwidth than that required to transmit the raw speech signal. A channel vocoder consists of an analyser and a voice synthesizer. CI speech processors use the signal analysing block of the vocoder. As shown in figure 5.2, a vocoder analyser filters the incoming speech signal into a number of contiguous frequency channels using a bank of bandpass filters. The output of each filter is then passed through an envelope detector, which consists of a full-wave rectifier and low-pass filter. This way, an estimation of the energy for each band is obtained as well as its evolution in time. In addition, the analyser makes the decision to allocate sound to that channel or not, and it also estimates its tone (F0). The dynamic range (DR) adaptation block transforms the acoustic DR for each channel into the electrical DR (defined as the span of intensities going from threshold to maximum comfortable level) necessary for each electrode. This transformation is specific for each patient and is different for each electrode. Finally, according to the stimulation rate (i.e. the number of pulses delivered to electrodes per second), the processor generates stimulation pulses representing the current level to be sent to each electrode at each time instant. In the majority of strategies, the stimulation pulses are generated in a way such that, at each moment, there is only one channel active, to avoid overlapping sensations caused by adjacent electrodes working at the same time. The stimulation pattern computed by the processor is transmitted to the CI receiver and the current pulses are then sent to the electrodes.

The performance of cochlear implants depends greatly on several factors such as the patient's age, duration and causes of deafness, the number of surviving spiral ganglion cells, the number of functional electrodes, the speech processing strategy and so on.

Coding strategies of speech processors for cochlear implants can be configured in different ways depending on the treatment of each stimulation parameter: the envelope detection method, the stimulation rate; the duration of the pulses, each of the phases and the rest time between them and between two consecutive pulses, etc. The main challenge of research nowadays is to determine these parameters optimally for each individual patient. To address this problem, psychophysical platforms that allow pulse parameter manipulations and that can collect consistent patient responses are crucial, which is one of the objectives of this thesis.

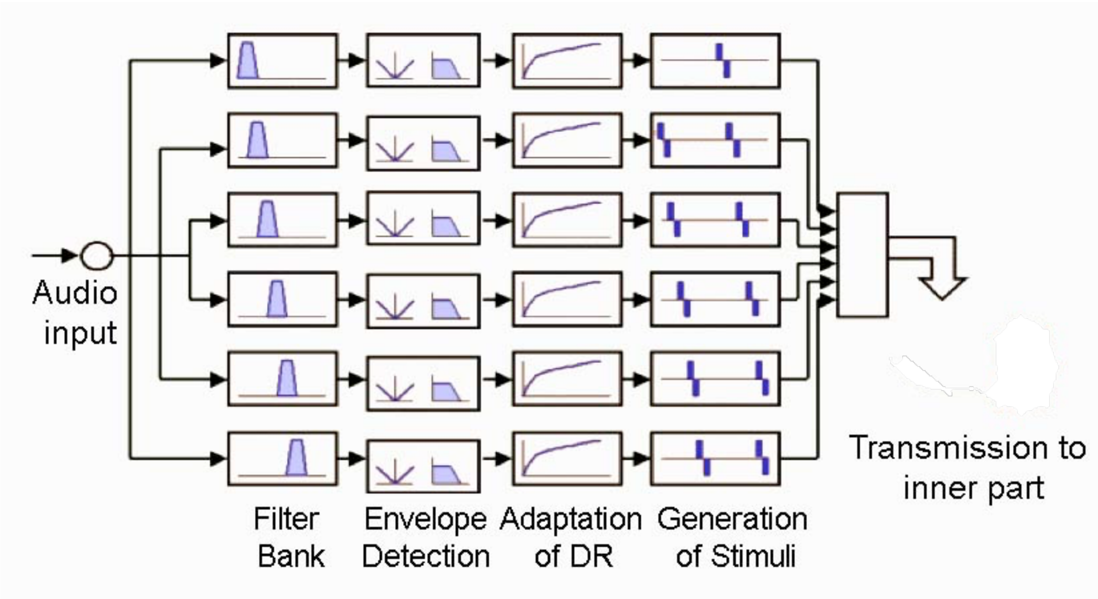

Figure 5.2. Block diagram of a CI system. Audio signal is acquired by the processor microphone and then amplified. Then it is passed to a filter bank in order to separate it into different frequency bands. The output of each filter is then passed through an envelope detector. The dynamic range adaptation block then transforms the acoustic dynamic range for each channel into the electrical dynamic range necessary for each electrode. Finally, the processor generates the stimulation pulses representing the current level to be presented at each electrode and at each time instant. (Own production)

5.2- Signal processing strategies

The signal processing strategy plays an extremely important role in generating the sounds heard by users. An ideal stimulating strategy is one

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that closely reproduces the original sound spectrum and allows a CI user to hear clear sounds. A complete stimulating strategy should address the following:

1. The number of channels selected to reproduce the original spectrum.
2. The number of electrodes activated to generate each channel.
3. The number of consecutive clock cycles required to deliver selected channels.
4. The scheduling of the activating sequence of electrodes.

Many stimulating strategies have been developed over the past two decades. The main strategies used today are the following:

Continuous Interleaved Sampling (CIS) strategy

Simultaneous stimulation in CI presents a major problem due to the interaction between channels caused by the sum of the electric fields of the individual electrodes (figure 5.3). Thus, neuronal responses to the stimuli of an electrode can be significantly distorted by stimuli from other electrodes. These interactions alter the spectral speech information and therefore degrade intelligibility. (38)

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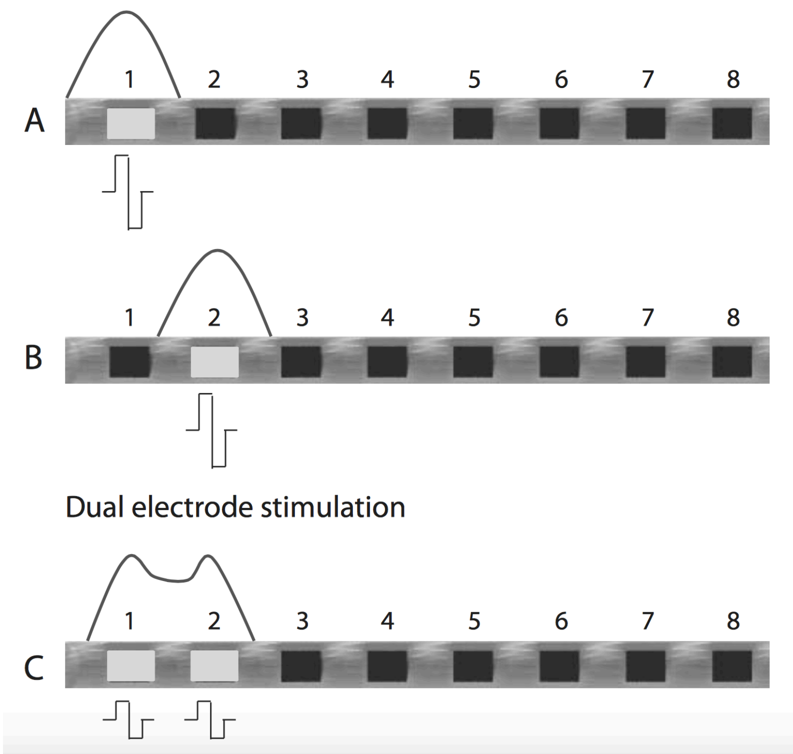

Figure 5.3. Simultaneous electrode stimulation effect. A and B represent single electrode stimulation patterns. C shows the overlap caused by simultaneous stimulation. (MEDEL)

The CIS strategy proposed by the Research Triangle Institute (RTI) addresses this problem using sequential, interleaved pulses. Figure 5.4 illustrates the signal processing procedure. Waveform modulated biphasic pulse trains are sent to the electrodes in an interleaved manner without overlap, so that at any time one electrode is stimulated. To generate the envelope of each channel, the speech coder method described above is used. The pre-amplified signal is then passed through a bank of bandpass filters. The envelopes of the outputs of these band-pass filters are then rectified and lowpass filtered (usually 200 or 400 Hz cutoff frequency). The envelopes of the outputs of the bandpass filters are compressed and used to modulate biphasic pulses. A nonlinear (e.g. logarithmic) compression is applied on the envelope according to the patient's dynamic range. Compressed envelopes are used to modulate a set of carrier signals of biphasic pulses at a constant speed and the biphasic pulses are delivered to the electrodes in non-overlapping way. (39)

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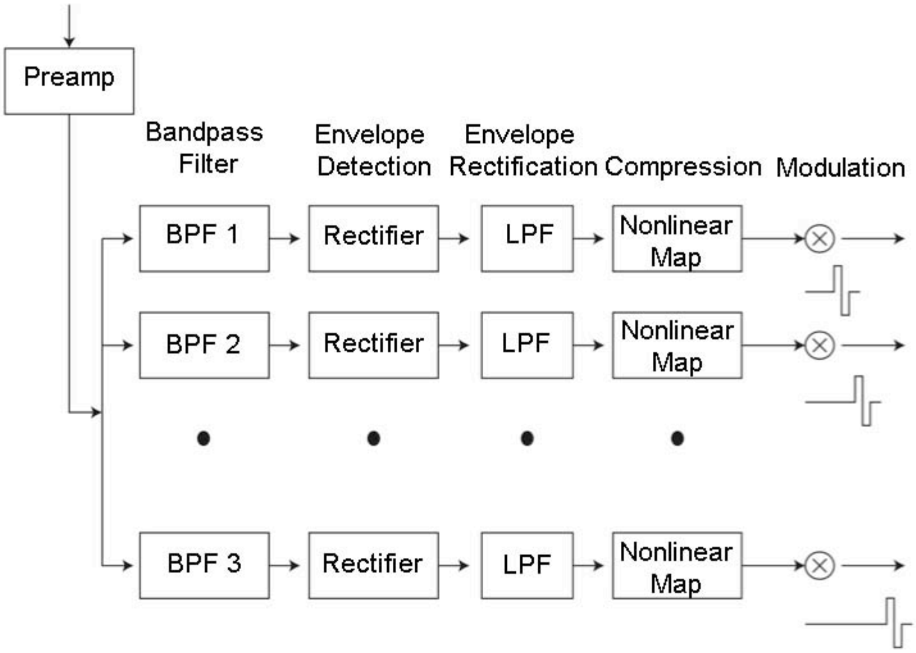

Figure 5.4. Block diagram of the CIS strategy. The signal is pre-amplified and filtered into frequency bands. Full-wave rectification and low-pass filtering then extract the envelopes of the filtered waveforms. Then the signal is compressed to fit the patient's dynamic range and then modulated with biphasic pulses. The biphasic pulses are transmitted to the electrodes in an interleaved way. (Own production)

N-of-m family of strategies (ACE, SPEAK)

It is believed that speech can be well understood when only peaks in the short-term speech synthesis are used. This principle is employed to activate only those channels with the most representative spectral maxima for each time bin. The strategy is similar to the CIS strategy, except that the number of stimulated electrodes is less than the total number of analysis channels. In this strategy, the signal is processed through m bandpass filters of which only a subset n (n<m) of the amplitudes of the envelope are selected for stimulation. More specifically, the maximum amplitudes of the curve n for stimulation are selected. The strategy is also known as "n-of-m". An advantage with the selection of fewer spectral bands is that the stimulation rate can be increased to provide more temporal information.

This strategy has been used in spectral peak coding (SPEAK) and advanced combination encoder (ACE) strategies. The SPEAK strategy selects the 6-8 largest peaks and has a fixed 250 Hz rate per channel. The ACE strategy has a larger range of peak selection and a higher rate than the SPEAK strategy. If n=m, then SPEAK and ACE strategies are essentially the same as the CIS strategy. (40)

Fine structure strategies

In contrast to coding strategies based on envelope detection that use a fixed stimulation rate (so that temporary stimulus information is lost), fine structure processing (FSP) uses the temporal structure of the sound signal, in particular in the range of low to medium frequencies. This is achieved by using specific sampling sequences for each channel. Pulses are initiated at each zero crossing of the positive sound signal slope, at the band-pass filter output for each channel. The length of these sequences is related to the higher frequency of the band-pass filter. Therefore, the rate of repetition of these sequences is equal to the instantaneous frequency of the fine structure of the signal in the respective frequency range. In the FSP strategy, this process is typically used in the most apical 2-3 electrode channels (low frequency), meaning that, depending on the arrangement of bandpass filters, it is used for frequencies up to 300-500 Hz. In the remaining channels stimulation remains tonotopic, as in previous strategies.

These strategies have evolved into FS4 and FS4-p. The first provides fine structure information on the apical four-electrode channels. In the second, these electrodes may be stimulated in a parallel manner. The operative hypothesis for this is that parallel stimulation enhances temporal information encoding as opposed to single-electrode stimulation. The hypothesis is supported by recent results from simulations of temporal processing in the cochlear nucleus which show that parallel stimulation is essential to encode temporal information (periodicity) successfully with variations in sound level. (41) (42)

5.3- Pitch perception with Cochlear Implants

Neurophysiological studies in animals have shown that pitch can be encoded by either the place of excitation as a function of stimulus frequency along the cochlea (place code) or the temporal structure of discharges that is phase-locked to the frequency (temporal code). However, their relative contributions have remained a subject of discussion for over 100 years and the issue still remains unsettled. As seen in the previous chapter, the place and temporal codes usually covary with stimulus frequency in acoustic hearing, but this characterization is not yet conclusive. (43) (44) (45) (46) (47)

In electric hearing, place pitch corresponds to the electrode position along the cochlea and the frequency band allocated to it. However, rate pitch has not been researched as extensively. Pitch estimate and rate discrimination in electric hearing have been mostly examined separately. Researchers have found that implant users could only detect differences in pitch for frequencies up to 300 Hz. These results suggest that this might be the upper boundary of the temporal code for pitch perception in CI users. (48) (49) (50)

However, their inability to discriminate a temporal pitch above 300-Hz does not inevitably mean that they cannot process temporal information at these frequencies. Physiological studies have found much higher synchronization in the auditory nerve's response to electric stimulation than acoustic stimulation. Psychophysical data show that implant users could detect temporal fluctuations at frequencies up to 4000 Hz. This use of higher than 300-Hz frequency boundary is in line with the phase-locking capability of the auditory nerve. However, the percept evoked by high-rate modulation or stimulation may reflect different degrees of 'roughness' in timbre rather than 'highness' in pitch. (51) (52) (48)

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5.4- Loudness perception with Cochlear Implants

In designing a speech processor for an implant listener, one of the most important factors lies in the proper transformation of acoustic amplitude into electric amplitude. Normal acoustic hearing can process sounds over a range of 120 dB. Speech sounds in a normal conversation can range from 40 to 60 dB. However, implant listeners typically have dynamic ranges of only 6 to 30 dB. Thus, the normal acoustic hearing range must be considerably compressed for electric stimulation, in order to preserve the acoustic amplitude variations that transmit important speech information. (53)

Loudness in electrically stimulated human listeners has been measured quantitatively using rating methods and magnitude estimation techniques. However, the existing data are not consistent across studies. Pfingst's group showed that loudness function depends on stimulus frequency for simple sinusoidal and pulsatile stimuli in cochlear implants. They found that loudness was an exponential function of stimulus amplitude for high frequencies (> 300 Hz) and a power function for low frequencies (< 300 Hz). Some investigators have found that log loudness grows linearly as a function of log current, although some data show two stages, with loudness that grows gradually at low stimulation levels and then grows steeply at higher stimulation levels (figure 5.5). Other groups found that loudness grows linearly as a function of charge. There has not yet been a widely accepted quantitative description in implant research of the relation between loudness and electric stimulus level. It is not clear whether the discrepancies among the above-mentioned loudness functions were due to procedural differences or due to individual implant listener differences. (54) (55) (56) (57) (58)

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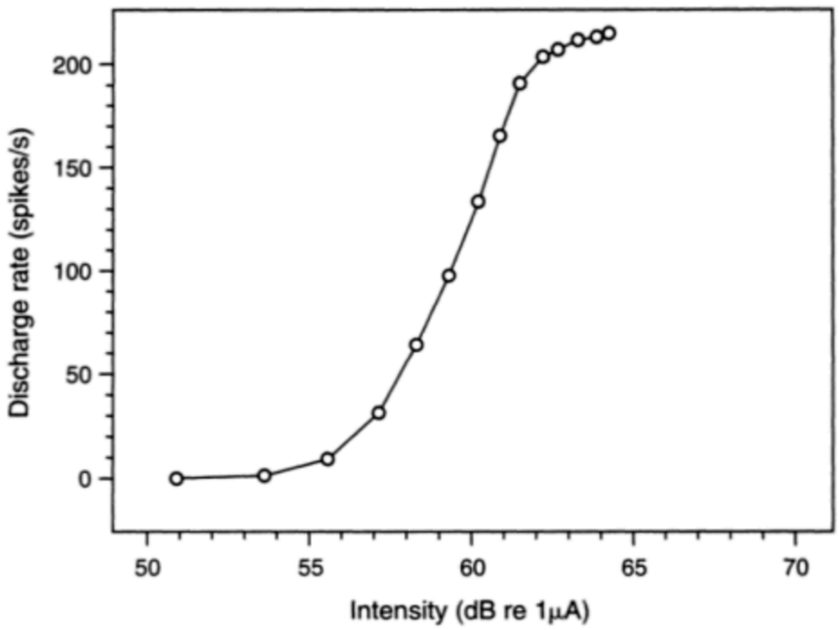

Figure 5.5. Growth function of discharge rate versus electrical pulse intensity. [19]

II. EXPERIMENTAL WORK

6.- PSYCHOACOPHYSICS: SPEECH PERCEPTION IN NOISE STUDIES

In everyday listening situations, accurate speech perception relies on the capacity of the auditory system to process complex sounds in the presence of background noise. There is ample evidence of auditory training resulting in perceptual enhancements. However, there have been surprisingly few investigations of how training impacts speech-in-noise perception. Even less has been done in the field of how high frequencies affect this ability.

In this chapter, two studies are described. The first focuses on the physical composition of sound, as a way to enhance speech-recognition in noise. More specifically, it investigates how speech's high frequency content affects its understanding in noisy environments. The second study deals with how training can enhance this ability even when sound has been band-frequency limited.

6.1.- Effects of High-Frequency Suppression for Speech Recognition in Noise in Spanish Normal-Hearing Subjects

(Published as: de Miguel, ÁR, Zaballos, MTP, Macías, ÁR, Barreiro, SAB, González, JCF, Plasencia, DP (2015). Effects of High-Frequency Suppression for Speech Recognition in Noise in Spanish Normal-Hearing Subjects. *Otol Neurotol*, *36*(4), 720-726.)

It is common for patients to ask why they hear but fail to understand or why they have trouble understanding voices. Part of the answer has to do with one's hearing and part with one's cognitive processing of the word. In this study, attention is focused on the first one. Hearing loss can affect the ability to understand the voices, especially when competing noise is present. With hearing loss, it isn't about not hearing sounds at all. It is about hearing just parts of the sounds, maybe even just the vowels, which have lower frequencies. To illustrate this, let's take the sentence: *high frequencies are important to understand speech in noise*. For someone with a moderate hearing loss in the higher frequencies, this might sound approximately like this: *_i _ __e_ue__ie_ a_e i_o_a__ _o u__e__a__ __ee_ i_ oi_e*. One would probably hear the words but not understand the meaning. Patients can sometimes complain about these problems but have an apparently normal audiometry. Thus, the aim of this study is to investigate whether frequencies above 8 kHz,-which are beyond the scope of a clinical audiometry and outside the bandwidth of operation of hearing prosthesis-, can have a significant effect on speech perception in noise.

6.1.1- Introduction

A healthy human ear can perceive frequencies from 20 to 20000 Hz. This range varies between individuals and degrades gradually with age, especially for high frequencies. Since the peak energy of fricatives spoken by female and child talkers tends to occur between 6.3-9 kHz, subjects with hearing loss (HL) in this range will have a limited audibility of such phonemes. Moreover, if there is important information contained in the high frequency end of the human hearing spectrum, then they will suffer from an important loss of information. In mild, moderate and high frequency hearing losses the only symptom may be subtle difficulty with word understanding, especially in situations where there is competing noise. (59) (60) (61) (62) (63)

A series of studies by Stelmachowicz et al. reported that when the high frequency components of a sound stimulus are removed, the perception of the phonemes /s/ and /z/ are seriously compromised in children with Normal Hearing (NH) and with hearing loss (HL). They investigated the effects of low-pass filtering on the perception of /s/ in four groups of listeners: children and adults with NH and children and adults with HL. Test stimuli were produced by an adult female, a male, and a child and were low-pass filtered at various frequencies from 2 to 9 kHz. In general, children with NH performed more poorly than the adults, and subjects with HL performed more poorly than their counterparts with NH. Importantly, in both cases mean performance for the female and child talkers improved up to a bandwidth of 9 kHz. These results suggest that the perception of fricatives may be difficult for listeners with HL, due to the limited bandwidth of current hearing prostheses. Nonetheless, the authors suggest that the effects of bandwidth may be task dependent. Since adults with post-lingual hearing loss are generally able to use semantic and syntactic cues in the perception of speech, the acoustic effects of removing high frequencies in speech may be masked in this age group. However, young children who are still learning speech and language may be more affected. Thus, it is important to establish how complex auditory conditions such as background noise can affect this ability of adults to compensate for a shorter audible frequency bandwidth. (63) (64) (65)

A number of authors have also studied the influences of high frequency on sound quality perception in various auditory situations. Gabrielsson et al. showed that the perceived sound quality of a female voice depended on the audible frequency bandwidth and on the smoothness in frequency response of the reproduction or amplification system. Moore and Tan also reported significant degradation in speech sound quality when the cut-off frequency decreased below 10.8 kHz. (66) (67)

These findings are also of vital importance in the area of Cochlear Implant (CI) development. Because such devices have a limited bandwidth of around 8kHz, their users cannot perceive auditory information from frequencies above this value. In spite of this, the majority of CI users have very good results in speech recognition tests in quiet. However, the same situation does not apply in more complex auditory conditions, such as in the presence of background noise. To get around this issue, CIs use a variety of strategies and noise filters, but even though some benefits have been reported, their performance is still not satisfactory. (68) (69) (70) (71) (72) (73) (74)

6.1.2- Objectives, hypothesis and rationale for the study

In order to understand why subjects with HL have difficulty in understanding speech in noise, it is first necessary to understand the processes allowing subjects with NH to direct attention to a desired source and comprehend it. However, the neural mechanisms that support speech discrimination in noise are not well understood. Therefore, **this study aims to evaluate how high frequency perception contributes to speech perception in noise**.

To this end, an experiment was designed to study speech understanding degradation as a function of maximum stimulus frequency in subjects with NH.

The hypothesis of this study is that high frequency filtering of the sound environment (words and noise) above 8 kHz causes a significant decrease in word recognition in noise.

The motivation for this study has come mainly from the realization that hearing prostheses only reach 8 kHz. Which means that any sound above it is not amplified (in the case of hearing aids) or heard at all (in the case of CI). Despite the fine spectral and even temporal resolution obtained for low frequencies in the field of CI, which correspond to the fundamental components of speech, patients still struggle with understanding in noise situations. Therefore, the cause might not be restricted to the low frequency, but also to the high frequency range. If NH subjects do make use of this information, then it would be interesting to alter CI design to try and find a way to broaden their frequency range.

6.1.3- Subjects and Methods

This observational cross-sectional study was undertaken at the Hearing Loss Unit in the Otolaryngology Head Neck Department, University Hospital of

Gran Canaria (Insular-Materno Infantil) (Universidad de Las Palmas de Gran Canaria). The corresponding Ethical Committee approved this study.

Subjects

A total of 29 subjects were selected, aged 19-55, with an average age of 30. All were native Spanish speakers. The subjects had normal hearing with pure tone thresholds better than 20 dB HL in all frequencies studied and with no other known pathology. None of them had prior experience with acoustic simulations or had participated in any hearing research experiment. All subjects came voluntarily and gave their informed consent.

Speech material and background noise

To study the effect of high frequency removal in speech recognition in noise, different amendments to the Spanish disyllabic speech lists were made. Frequency modifications and noise files were generated using *MATLAB and Statistics Toolbox Release 2012b* (The MathWorks, Inc., Natick, Massachusetts, United States).

Six validated lists of 25 words each and spoken by a female speaker with a Spanish accent were used (Appendix II). They belong to the standard Spanish Disyllabic test, commonly used in hearing-loss assessments. Two groups were made of three lists each. In the first group, frequency components were unaltered and in the second group the words were band-pass filtered. For future reference, we will refer to them as groups A and B respectively. The audio files were saved in Waveform Audio File Format (.Wav) format with a sampling frequency (Fs) of 44 kHz. Fs determines how many times per second the analogue signal is digitally sampled by the recording system. According to the Sampling Theorem, for a band-limited signal to be reconstructed fully, this must be done at twice the rate of the highest frequency that it is of interest to record. Therefore, the maximum sound frequency that could be reproduced with this sampling rate was 22 kHz. (75)

To generate group B, the unfiltered lists were passed through a Butterworth filter of order 4 with a band pass from 70 Hz to 8 kHz, available in the signal processing toolbox in MATLAB. This type of filter is designed to have a maximally flat frequency response in the band of interest, with a gain given by the transfer function:

$$G^2(\omega) = \frac{G_0^2}{1+\left(\frac{\omega}{\omega_c}\right)^8} \text{ [6.1]}$$

Where ω_c is the cut-off frequency and G_0 is the gain for frequencies below ω_c. So from this function it can be seen that frequencies above 8 kHz will be supressed, while those below will be passed with a gain G_0. The filter has a pass tolerance of 0 dB SPL and an elimination tolerance of 40 dB SPL. The procedure is illustrated in Figure 6.1. The outcomes were three sound files with a maximum frequency of 8 kHz, as desired. (76)

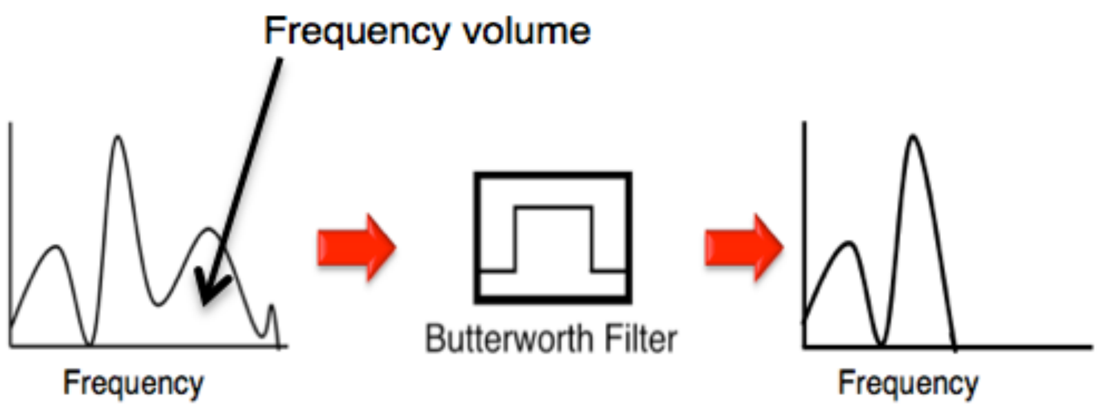

Figure 6.1. Butterworth filtered FFT of a single word. The original signal is passed through a filter that removes the high frequency part of the spectrum, leaving the lower components unaltered. (Own production)

Two white noise files were also generated. The first was sampled at 44 kHz and the second was created from the first, but passed through the Butterworth filter. Three sets of SPLs were used for the two noise files: 60 dB, 65 dB and 70 dB.

Therefore, each one of the three lists in groups A and B were assigned a noise file at a particular sound level. This way, three sets of SNR conditions were created for each group: SNR +5dB, 0dB and -5dB. Figure 6.2 further illustrates the procedure for the preparation of sound stimuli. Thus the subjects listened to six lists at 65 dB SPL in total: 3 lists from group A, each assigned a noise file (60, 65 and 70 dB SPL), and similarly for group B.

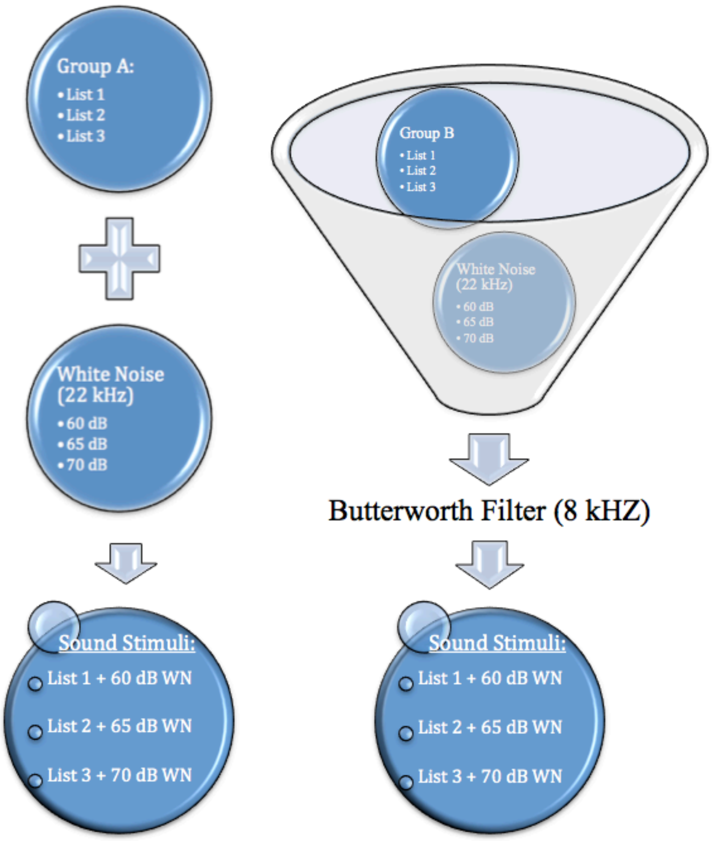

Figure 6.2. Preparation of sound stimuli: 6 sets of lists were divided into two groups (A and B) and three noise files were generated at 60, 65 and 70 dB. Each of the lists in group A were played along one of the three noise conditions. Group B lists and three noise files of the same characteristics as before were first filtered using a Butterworth filter. Similarly, each list was reproduced together with one of the three noise files. (Own production)

Procedures

-Experimental set up: The experimental set up is depicted in figure 6.3. During the experiment the listener was seated 1 m away from two loudspeakers *Samson Resolv A5* (Samson Technologies, Haupauge, New

York, United States). The loudspeakers had two voids, a frequency response of 50 Hz-30 kHz and were placed in azimuth 0º position to the subject. The experiment was done in a sound treated room, with an attenuation of 32 dB. Sound was produced in the acoustic card of a *MacBook Pro, Core i5 Intel double core* 2,4 GHz computer with sample frequency 44 kHz (Apple Inc., Cupertino, California, United States). The software used was the *Audacity v2.0.5* software (Audacity Computer software audio editor, freely available for all computer platforms from: http://audacity.sourceforge.net/).

Each listener was presented the lists without prior practice time. All subjects listened to six lists plus noise from -5 to 5 dB SNR, but the order of presentation of groups A and B was random to avoid adaptation effects. The sound pressure level of the lists was fixed at 65 dB SPL. The calibration was done using a continuous pure tone at 1 kHz as reference.

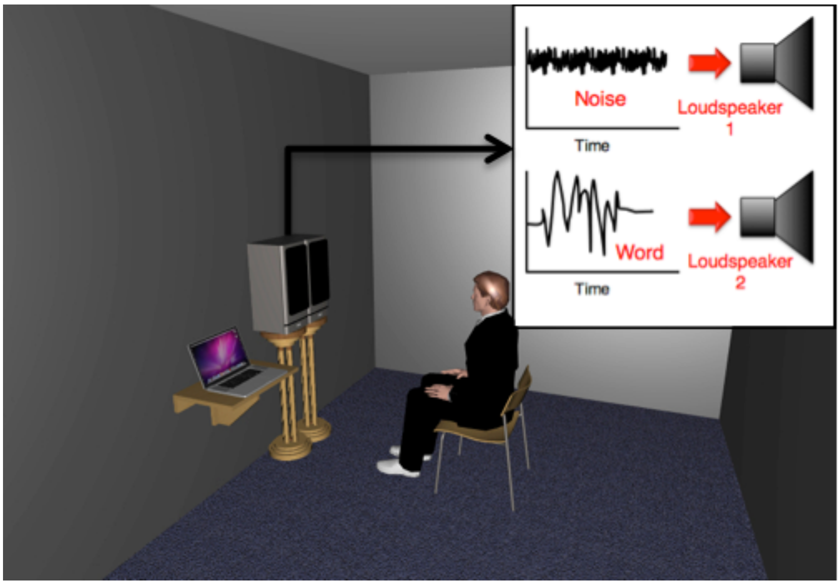

Figure 6.3. Experimental set up. Subjects were seated in a sound treated room, 1m away from two adjacent loudspeakers. The word lists and noise files were played separately from each of the loudspeakers. (Own production)

- Statistical analysis: The results of individual word success rate per list per patient were collected using *Microsoft Excel 2011* (Microsoft. Microsoft Excel. Redmond, Washington, United States). The statistical analysis was done in

IBM SPSS Statistics for Macintosh, V. 21.0. (Armonk, NY: IBM Corp). A two-tailed t-test was used to compare performance between groups A and B for each SNR condition.

Then, how the amount of change in the frequency spectrum of a word affected its success rate was examined. To this end, the spectrogram of each word from group B was compared to its unfiltered version. The hypothesis tested was whether words whose spectrograms differed significantly after filtering yielded poorer performance. The correlation between word spectrograms was done using the MATLAB function *mscohere(x,y)*. This function finds the magnitude squared coherence estimate of the input signals x and y (e.g. words 1 and 2 of list A1) using Welch's averaged modified periodogram method. The result is a function of frequency with values between 0 (no correlation) and 1 (full correlation), that indicates how well x corresponds to y for each frequency. It is a function of the power spectral densities of x and y and the cross power spectral density. It returns a frequency-based cross-correlation graph of each frequency component, like that shown in figure 6.4. For a complete description of the function please refer to the MATLAB documentation section. Then, an average of the frequency spectrum correlation was calculated and a Pearson correlation test was done between this value and the word score. (77)

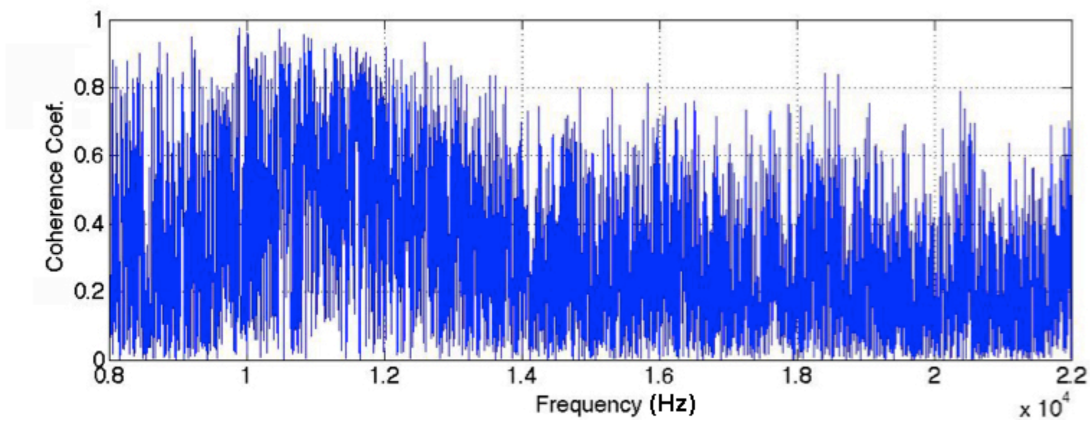

Figure 6.4. Example of cross-correlation results of words 1 and 2 of list A1. The x axis represents sound frequency and the y axis represents the degree of correlation of each frequency value. (Own production)

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6.1.4- Results

The goal of this study was to investigate the effects of stimulus bandwidth on speech recognition in noise. To this end, the overall percentage-correct scores and correlations between individual-word scores and high frequency content were statistically analyzed.

Overall Performance

The complete individual word score for each list is disclosed in Appendix III. As expected, overall word identification decreases as noise increases. To quantify the extent to which subjects were able to make use of high-frequency information, scores for speech in noise for broadband and filtered speech at 8 kHz were compared. Figure 6.5 shows the percentage average scores on the disyllabic lists. They are plotted as a function of SNR for filtered (blue) and unfiltered (red) cases. Error bars indicate one standard deviation. The mean scores for 5 dB SNR were 90.86 and 73.71% in the filtered and unfiltered case respectively; for the case 0 dB SNR, mean scores were 66.29 and 67.43% and finally for -5 dB SNR, these were 29.71 and 46.86%.

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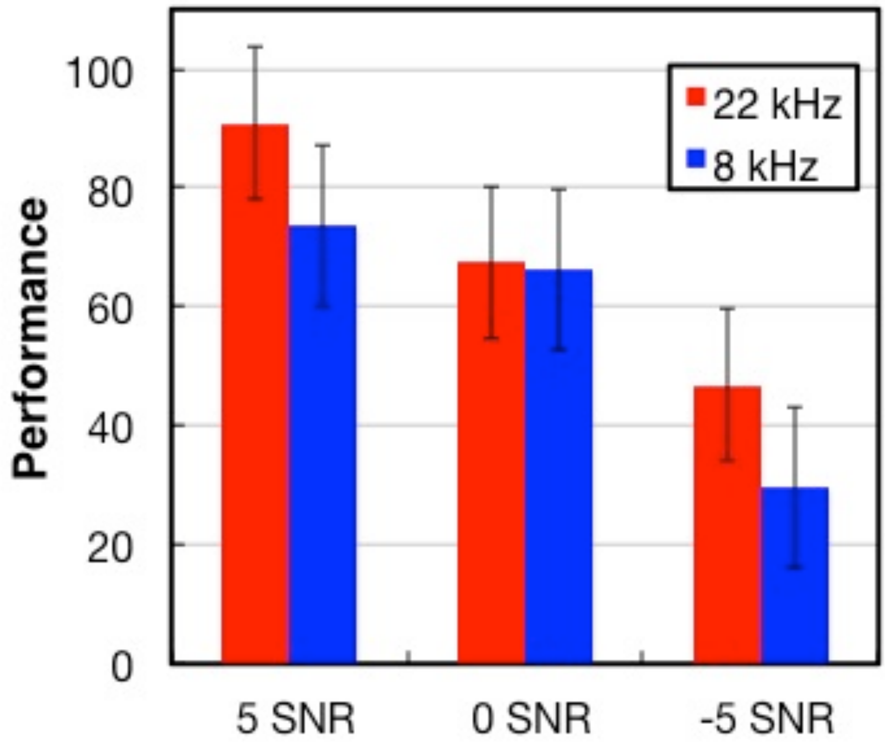

Figure 6.5. Success Rate for each SNR condition in both filtered and unfiltered situations. Statistically better performance is observed when high frequencies are preserved for ±5 dB SNR ($p < 0.05$). However, for SNR= 0 no statistical difference is observed. (Own production)

A tendency towards better performance is observed in the cases where high frequency components of words are not suppressed, for the conditions ±5 dB SNR. In these cases, 17% more of the words were correctly identified in the lists where frequency components were not filtered. The t-student test for the conditions 5 and -5 dB SNR revealed significant effects for frequency filtering: $t = 9.354$; $p = 0.003$ for 5 dB SNR and $t = 23.480$; $p = 0.000$ for -5 dB SNR. However, for the intermediate case, where signal and noise were at the same level, both conditions yield statistically similar results: $t = 0.010$; $p = 0.919$.

Spectrogram Analysis

Given the above results, it is of interest to determine which phonetic features are affected most by changes in bandwidth and whether the effect of bandwidth transmission of phonetic features is associated with scores for individual words. To examine these issues, correlations between the amount of alteration in a word and its mean score were analysed for group B lists (the filtered ones). The following hypothesis was evaluated: words whose high frequency spectrum had suffered the highest changes gave lower scores.

Table 6.1 summarizes the results of the Pearson correlation for each list. The only statistically significant correlation was found for SNR = 5 (p = 0.037). In the remaining cases, no correlation was found between these two variables (p = 0.487 for 0 dB SNR and p = 0.171 for -5 dB SNR).

PEARSON CORRELATIONS		
5 SNR	Pearson Coefficient	0.363
	Sig	0.037
0 SNR	Pearson Coefficient	-0.007
	Sig	0.487
-5 SNR	Pearson Coefficient	-0.198
	Sig	0.171

Table 6.1. Pearson correlation results for the comparison of autocorrelation of words pre and post filtering and their total score. The only significant correlation was found for the case 5 dB SNR. (Own production)

6.2.- Effects of speech-in-noise training and high frequency suppression in Air Traffic Controllers. A control group study

(Published as: Zaballos, MTP, de Miguel, ÁR, Plasencia, DP, González, MLZ, Macías, ÁR (2015). Effects of long-term speech-in-noise training in air traffic controllers and high frequency suppression. A control group study. *J Int Adv Otol*, *11*(3), 212-7.)

Learning electrically stimulated speech patterns is a new and difficult experience for many CI users. It does not suffice to have been given the tools to hear. These patients require extensive rehabilitation and auditory training to be able to make use of the information provided by the device, identify differences in sound parameters and to make sense of them. Specifically, speech-in-noise-understanding still presents a great challenge to most CI users. Given that CI and hearing aids alone do not restore these abilities, efforts are currently focused on the role of the brain in such situations and whether training can be an efficient therapy. In this study, air traffic controllers have been studied because they seem to have developed a greater ability to understand speech in noise than the average individual. This ability is a result of their continuous and long-term auditory training in noise under restricted frequency bands. This is an encouraging discovery for the field of cognitive sciences and auditory rehabilitation. Moreover, the effect of high frequency suppression for speech-in-noise understanding observed in the previous study has been further assessed for this job sector to evaluate possible learning effects.

6.2.1- Introduction

The ability to understand speech in noise requires both sensory and cognitive skills (78). The sensory part consists of the auditory system locking on to the target speech signal while excluding ambient noise and competing voices. This is achieved by organising auditory inputs into different groups by identifying shared characteristics, such as location and acoustical similarity. The identification of these groups is given by the relative stability of voice pitch and helps in grouping it separately from other voices. In addition, other signal-based cues like harmonics, location and timing, aid in group formation of speech. On the cognitive side, attention and working memory skills are key to a good speech-in-noise (SIN) ability. (79) (80) (81)

There is evidence that such skills can be improved during an adult's life, causing perceptual enhancements, plasticity in single neurons and in neural populations (82) (83). Studies of auditory perceptual learning reveal long-term neural changes in the adult auditory cortex (AC) of both animals and humans after intensive auditory training (84) (85). Current models propose that perceptual learning in adults depends strongly on top-down influences such as attention, reward, and task relevance. This demonstrates that the adult AC is a dynamic and adaptive processing centre. (86) (87) (88) (89) (90) (91) (92) (93) (94) (95)

Natural auditory training occurs in job sectors with frequent daily radio communications. This is the case of Air Traffic Controllers (ATC). People that work in this field are constantly exposed to SIN, which is extremely hard to understand to an outsider. During their working life, ATC learn to identify, extract and comprehend conversations embedded in white noise. These conversations are normally restricted to the aeronautics field, which makes them an ideal population to study the effects and transferability of long-term auditory training.

Unfortunately, the majority of studies on how training affects SIN perception used small stimulus sets and revealed that, while SIN perception can improve in artificial listening conditions, the benefit can be extended to untrained material is not clear. (96) (97) (98) (99)

6.2.2- Objectives, hypotheses and rationale for the study

As opposed to previous work where the patients are trained in situ and then tested, this study looks at how someone's job based listening experience can lead to enhanced performance on a laboratory-based test. Moreover, the role of high frequencies has also been evaluated because radio communications are limited, just like hearing prosthesis, to 8 kHz.

Thus, this study has two main objectives. The first is **to evaluate whether Air traffic Controllers enhanced ability to understand speech in noise could be tested in a speech-in-noise laboratory test.** To this end, a control group of normal hearing listeners and a target group of ATC were assessed. The second is **to compare their speech in noise performance in high frequency (>8kHz) filtered and non-filtered conditions to that of normal hearing individuals from the previous study (control group).**

The hypotheses are: 1) that **ATCs understand speech in noise significantly better than an ordinary normal hearing individual**; 2) **the differences between filtered and non-filtered word recognition are lower than those for the control group.** The first hypothesis is proposed on the basis that these individuals have undergone speech-in-noise training during adulthood. The second is constructed on the fact that their training material involved high frequency filtered radio communications (>8 kHz).

The reason behind this study is that our aging society is bound to suffer speech-in-noise-understanding impairments. As individuals grow older, this ability degrades even before auditory losses become clinically relevant. Therefore, it is capital to come up with strategies to try and slow this process down as much as possible. Given the recent evidence that learning during adulthood is far greater than previously thought, the demonstration that auditory training under degraded sound quality can lead to better speech in noise understanding can have important implications in the way we treat hearing losses today. An effective rehabilitation program can improve people's quality of life by easing their integration back into the sound complexity of society.

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6.2.3- Subjects and Methods

This study was done at the Air Base of Gando Airport of Gran Canaria and the Psychoacoustics Laboratory of the Hearing Loss Unit, Otolaryngology Head and Neck Department, Complejo Hospitalario Universitario Insular Materno Infantil de Gran Canaria (CHUIMI). The Ethical Committee approved this study.

The methods of this study are analogous to those for the previous study. So only the adds-on will be described.

Subjects

A total of 29 normal-hearing subjects (aged 19-55) from the previous study were used as control group and 48 ATC (aged 34-56), as the target group. All subjects were native Spanish speakers. The subjects had NH with pure tone thresholds better than 20 dB HL as measured by a validated clinical audiometry (from 0.5 to 8 kHz in steps of 0.5 kHz) and with no other known pathology. None of them had previously taken part in any auditory experiments before and they all signed an informed consent before commencement of the study.

Age histograms for the ATC and control groups were illustrated in figures 6.6 and 6.7, and years working in the sector for ATC in figure 6.8.

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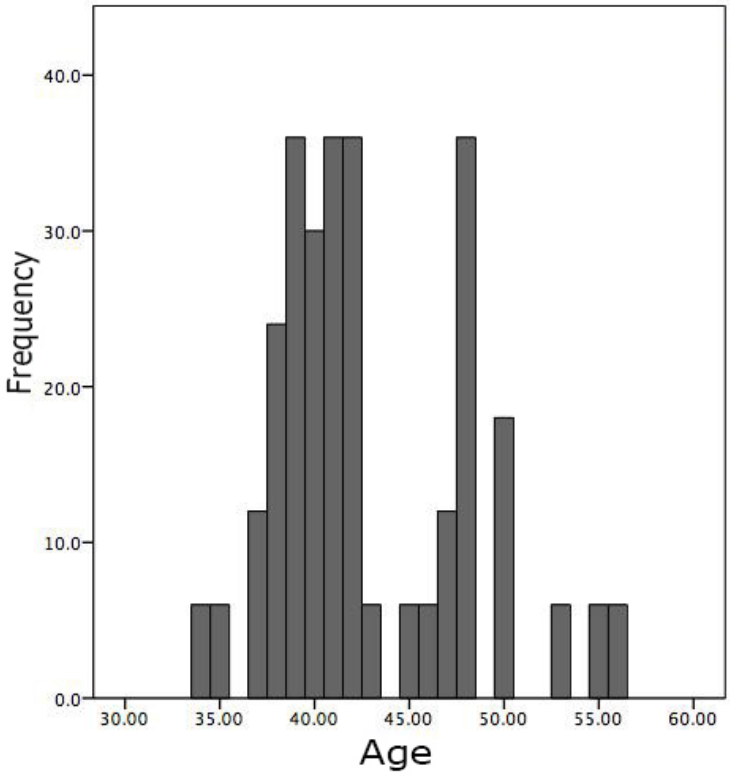

Figure 6.6. ATC age histogram.

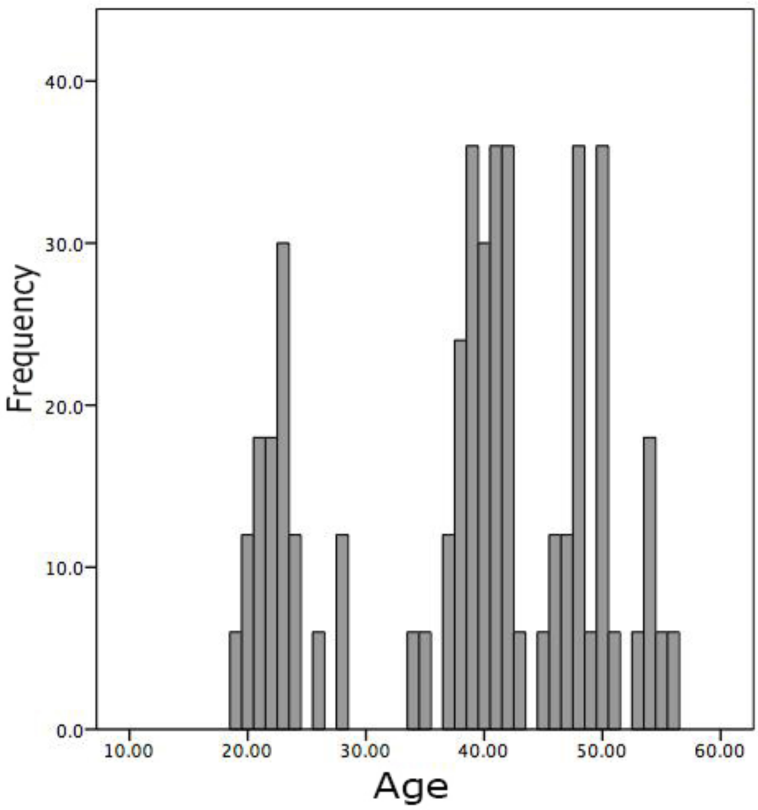

Figure 6.7. Control group age histogram.

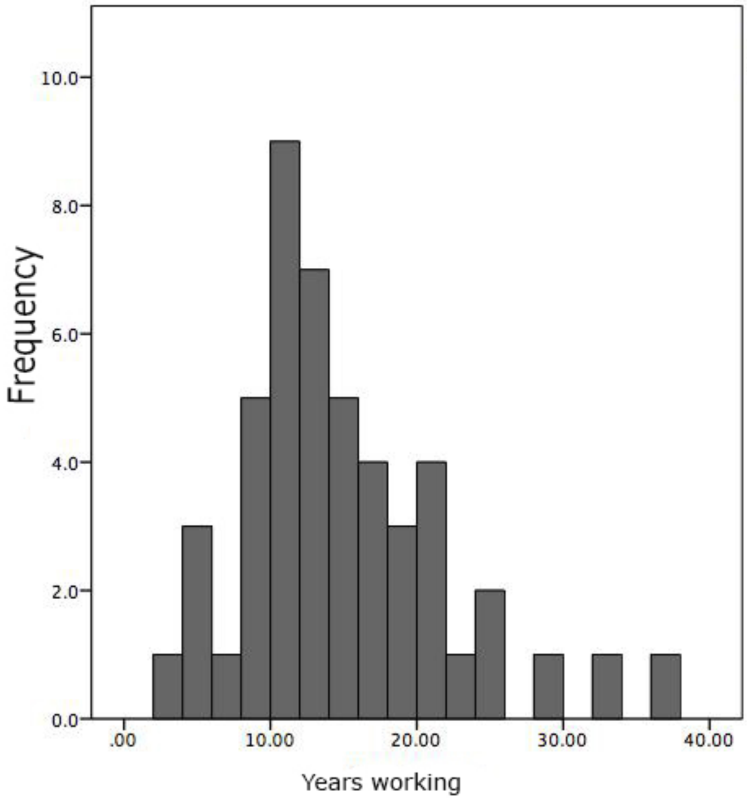

Figure 6.8. Years working as ATC histogram.

Speech material and background noise

The speech material was the same as for the previous experiment.

Procedures

The experiment procedure is analogous to that of the previous chapter.

The data was recorded in an .xls file, *Microsoft Excel 2011* (Microsoft. Microsoft Excel (Redmond, Washington, United States) and *IBM SPSS Statistics for Macintosh, V. 21.0*. (IBM Corporation, New York, United States) was used for the statistical treatment. The statistical analysis starts with a normality check for each specific condition using the Kolmogorov-Smirnov test for one sample. According to the result, a non parametric or a two tailed t-student test was performed. Finally, correlations between age, years of experience in the Aeronautic field and test scores were analysed using a Pearson correlation test. The level of significance chosen was 5%.

6.2.4- Results

First, overall performance difference between the two groups was analysed. Figure 6.9 shows how the ATC group outperforms the control group in both conditions, -filtered and non-filtered-, and for all SNR conditions. Moreover, the difference between them becomes larger as noise increases. Normality check gave Gaussian distributions. The evaluation of significance returned p-values < 0.01 for all cases, thus revealing significant mean differences between the two groups. For SNR 5 dB, this difference was 7.27% for list A and 13.25% for list B. For SNR 0 dB, these values rise to 16.95% and 11.45%. The most interesting difference is observed at SNR -5 dB. In this case, ATC outperform the control group by 17.03% and 22.52%.

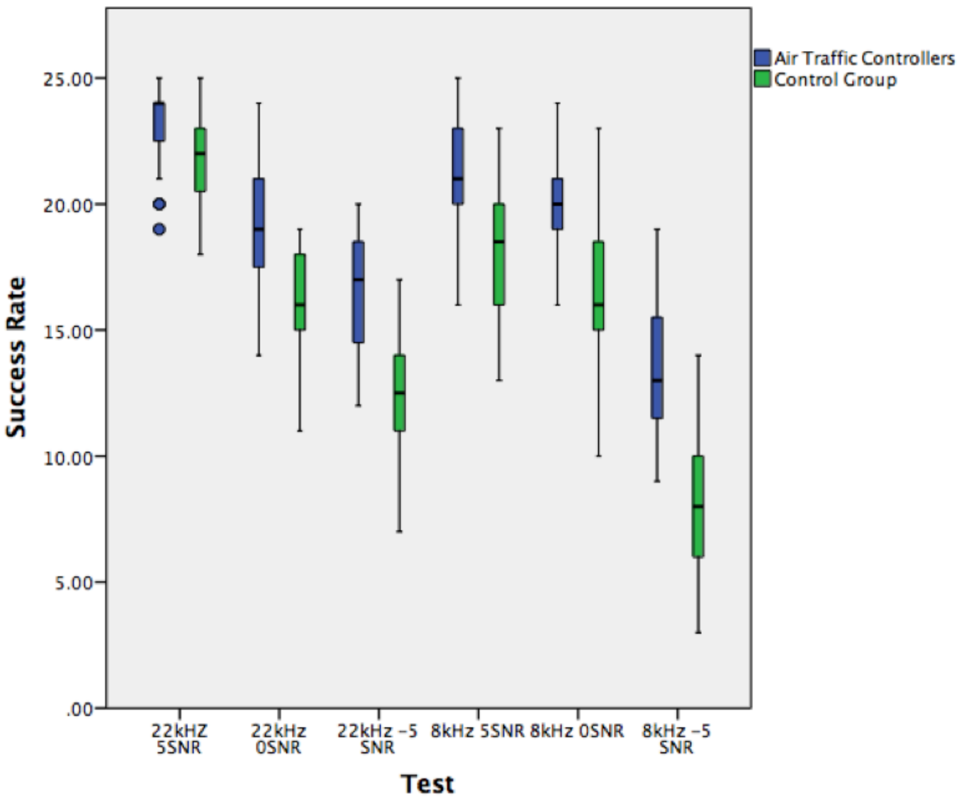

Figure 6.9. Success of ATC and control groups as a function of the test. The ATC group outperforms the control group in all SNR conditions and both in non-filtered (A=22 kHz) and filtered (B=8 kHz) conditions.

Figures 6.10 and 6.11 illustrate the relation between years working as ATC and test performance and between age and test performance. The correlation study between age, years working in this profession and overall test results for each condition is shown in Table 6.2. No significant correlation was found between working experience and overall test results. A detailed analysis of each case revealed significant ($p<0.05$) inverse correlations between 8 kHz 5 SNR test results and both variables. 22 kHz 5 SNR also gave a significant, inverse link between age and test results (Table 6.3). However, these results do not have a physical justification.

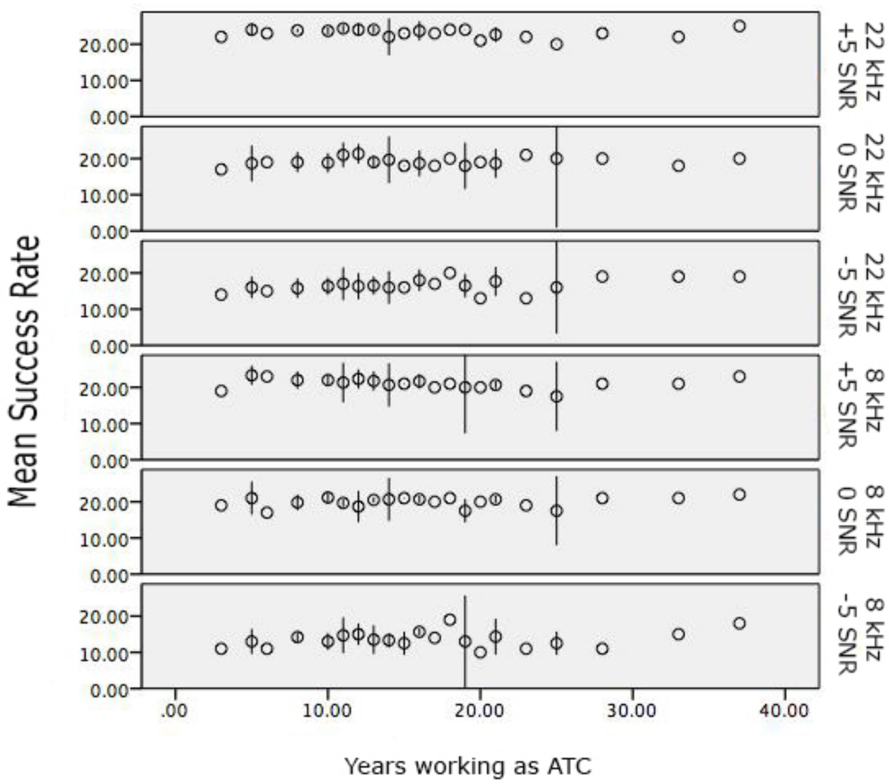

Figure 6.10. Success rate as a function of years working as ATC.

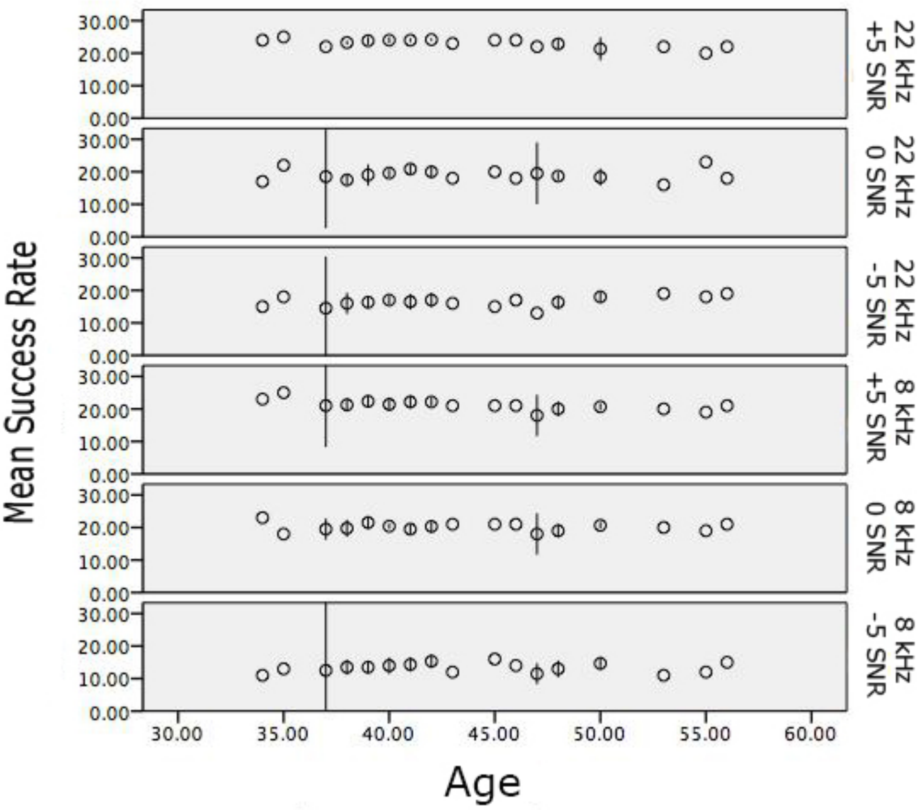

Figure 6.11. Success rate as a function of age for the ATC group.

	Age	Years working as an ATC	Test results
Age			
Pearson Correlation		0.759	-0.068
Sigma (bilateral)	-	0.000	0.249
N		288	288
Years working as an ATC			
Pearson Correlation			0.002
Sigma (bilateral)		-	0.973
N			288
Test results			
Pearson Correlation			
Sigma (bilateral)			-
N			

Table 6.2. Pearson correlation of age, years working as ATC and overall test results. No correlation was found between tests results and either age or experience. Obviously, in the vast majority of cases, the older the ATC, the longer they have worked, so there is a correlation between these two variables.

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		Years working as an ATC	Age
Years Working as an ATC	Pearson Correlation	1	0.759
	Sig. (bilateral)		0.000
	N	48	48
Age	Pearson Correlation	0.759	1
	Sig. (bilateral)	0.000	
	N	48	48
8kHz 5SNR	Pearson Correlation	-0.288	-0.411
	Sig. (bilateral)	0.047	0.004
	N	48	48
8kHz 0SNR	Pearson	0.012	-0.141
	Sig. (bilateral)	0.937	0.340
	N	48	48
8kHz -5SNR	Pearson Correlation	0.134	-0.021
	Sig. (bilateral)	0.363	0.886
	N	48	48
22kHz 5SNR	Pearson Correlation	-0.279	-0.503
	Sig. (bilateral)	0.055	0.000
	N	48	48
22kHz 0SNR	Pearson Correlation	0.049	-0.040
	Sig. (bilateral)	.741	.787
	N	48	48
22kHz -5SNR	Pearson Correlation	.264	.188
	Sig. (bilateral)	.069	.200
	N	48	48

Table 6.3. Correlation between years working, age and individual test results. Significant inversed correlations were found for: a) 8 kHz 5 SNR and years working; b) 8 kHz 5 SNR and age; c) for 22 kHz 5 SNR and age.

ATC performance was also evaluated independently to study the effect of high frequency filtering on speech perception in noise. Statistical differences between A and B lists were searched for to evaluate if filtering affects ATC as much as the control group from the previous study, were high frequency filtering significantly affected speech recognition in noise, suggesting that NH subjects use high frequencies above 8 kHz to identify speech in noise.

The ATC group performs better when words are not filtered with $p<0.01$ in all SNR conditions. For 5 dB SNR, 7.95% more words were correctly identified in list A; t(83)= 1.99, p= 5.20E-07. For 0 dB SNR, 5.08% more words were guessed correctly. Finally, for -5 SNR, a difference of 11.2% is observed between A and B lists. The results indicate that ATC are also affected by high frequency removal in speech. In this case, the effect is observed for all SNR conditions.

7.- ELECTRICALLY EVOKED PSYCHOACOUSTICS IN COCHLEAR IMPLANT RECIPIENTS

The level of hearing rehabilitation obtained through cochlear implants has increased dramatically since their introduction. This is the result of developments that include advanced signal processing, higher stimulation rates, greater number of channels, and the development of more efficient electrode arrays. The earliest cochlear implant recipients could only obtain benefits when combined with lip-reading, and were only able to recognize environmental sounds, but little or no speech using only the auditory information provided by the implant. As multichannel cochlear implants were introduced, several studies demonstrated that their users were capable of discriminating speech without visual assistance, thus succeeding in successfully restoring a sense for the first time in human history. (35)

Multichannel electrode array design has evolved into two major categories: straight and perimodiolar electrodes. When implanted, the first category of arrays lies along the outer wall of the scala tympani. However, animal experiments and computer modelling studies have indicated that the preferable position of an electrode array is close to the modiolus. A perimodiolar position of the electrode array could be expected to result in reduced stimulus thresholds and stimulating currents, an increased dynamic range, and more localized stimulation of the neural elements. However, their advantage for pitch discrimination has not been conclusively stated. (100) (101) (102) (103) (103) (104)

This chapter looks at how physical variables, especially electrode distance to the neural ends, affect electrode discrimination. To this end, a psychophysical platform was purpose-design and then a three-stage study was conducted.

7.1.- PsychoLAB: A Psychophysics experimental setup to measure electrode discrimination in Nucleus cochlear implant recipients

(Published as: Zaballos, MP, de Miguel, AR, Killian, M, Macías, AR (2016, February). A Psychophysics experimental software to evaluate electrical pitch discrimination in Nucleus cochlear implanted patients. In *Journal of Physics: Conference Series* (Vol. 689, No. 1, p. 012030). IOP Publishing.)

In order to study electrode independence, a psychophysical software has been developed, making use of the *Nucleus Implant Communicator* (NIC) Python library provided by Cochlear company under a research agreement. The application comprises a graphical interface to facilitate its use, since previous software has always required some type of computer language skills. It allows for customization of electrical pulse parameters, measurement of threshold and comfort levels, loudness balancing and alternative forced choice experiments to determine electrode discrimination in Nucleus© users.

7.1.1- Introduction

The results of several experiments in which electric field patterns for different electrode configurations have been investigated within the cochlea, reveal that the spatial gradient increases as the distance between the stimulating electrodes and neuronal tissue decreases. Therefore, an electrode that is closer to the neuronal endings would be expected to reach the hearing threshold current at lower levels due to the steeper slope of the electric field (105) (106). The result would be higher for sets of electrodes placed laterally. In addition, electric fields steeper slopes are less spread to adjacent neuronal populations, which decreases the probability of interaction between channels. This can result in a better discrimination between the electrodes (107) (108) (109).

However, the selectivity of the electric fields about the neural stimulation depends on a number of factors apart from the electrode-nerve proximity. These include: patterns of neuronal survival population, stochastic variability characteristics of individual nerve fibres, cochlear anatomy, signal characteristics and design features of the electrode (110). Many of these factors may be interdependent and may affect the threshold as well as the selectivity. For example, an individual with a poor nerve survival will have poorer selectivity and higher thresholds than one with good nerve survival, regardless of the type of electrode array, as the subject requires higher values to achieve optimal neural excitation.

Several research groups have designed psychophysical platforms to reduce the amount of programming needed to build experiments. Although these are very comprehensive and adaptable tools that allow the researcher to design virtually any experiment on psychoacoustics, they lack the necessary tools to perform electrical stimulation experiments. (111) (112) (113) (114)

An exception to the limitations of these platforms is APEX, a software application build and distributed under contract by the Experimental Audiology Department in Leuven (115). This platform supports psychoacoustic experiment design both for acoustical and electrical stimuli by means of a cochlear implant, and even the combination of both. In APEX, the actual experiments are designed in XML format. Even though this is a great advantage and requires much less learning time, it still requires some degree of computer language knowledge, since XML is not as straightforward as a visual interface.

This chapter describes the visual interface-based psychophysical software that was developed to conduct the electrode discrimination experiment described in the next chapter. This tool has helped to work towards better understanding of the psychophysics of electrical hearing. The platform is controlled exclusively through a graphical interface, where pulse parameters are entered in purpose-designed cells and loudness balance and alternative-forced-choice experiments are already built in.

7.1.2- Platform design

The present Psycoachoustic Research Platform was designed using *Python* (Python Software Foundation, v2.3) to define the stimulus characteristics with the help of the Nucleus Interface Communication (NIC) library and *Visual Studio* (Microsoft Corp. Visual Studio Community 2013) to design the researcher and patient interfaces. NIC is a research tool developed by Cochlear LTD and allows researchers to build applications to control the electrical stimuli delivered by the intracochlear electrodes of Nucleus Cochlear implants. Cochlear LTD. Also provided us with a special processor that can be connected to a computer via USB, so that the orders to the internal part of the implant come directly from Python commands. The application supports the three steps required to perform a psychoacoustic experiment: 1) stimulus definition, 2) experimental procedure and 3) analysis of the results.

The Case of Uses diagram shown in figure 7.1 illustrates the researcher and patient interactions with the system. The researcher can insert, edit or delete patient data and can choose to create or load a previous session to work with. The researcher interface also allows the definition of stimulus parameters. Once patient data and stimulus have been created, the researcher can interact with the patient through a series of tests. First the C and T levels need to be determined for the defined stimulus. Here the patient provides oral feedback. Then the electrodes are loudness balanced by asking the patient when the level of a test electrode matches a reference one, which the patient should indicate by pressing the equal-sign button. Finally, the experiment in itself is done by letting the patient hear a series of three sounds and indicate which of the three sounds is different, pressing one of the three buttons displayed in the patient screen. The program will register the answers and calculate electrode discrimination performance as percentage correct scores.

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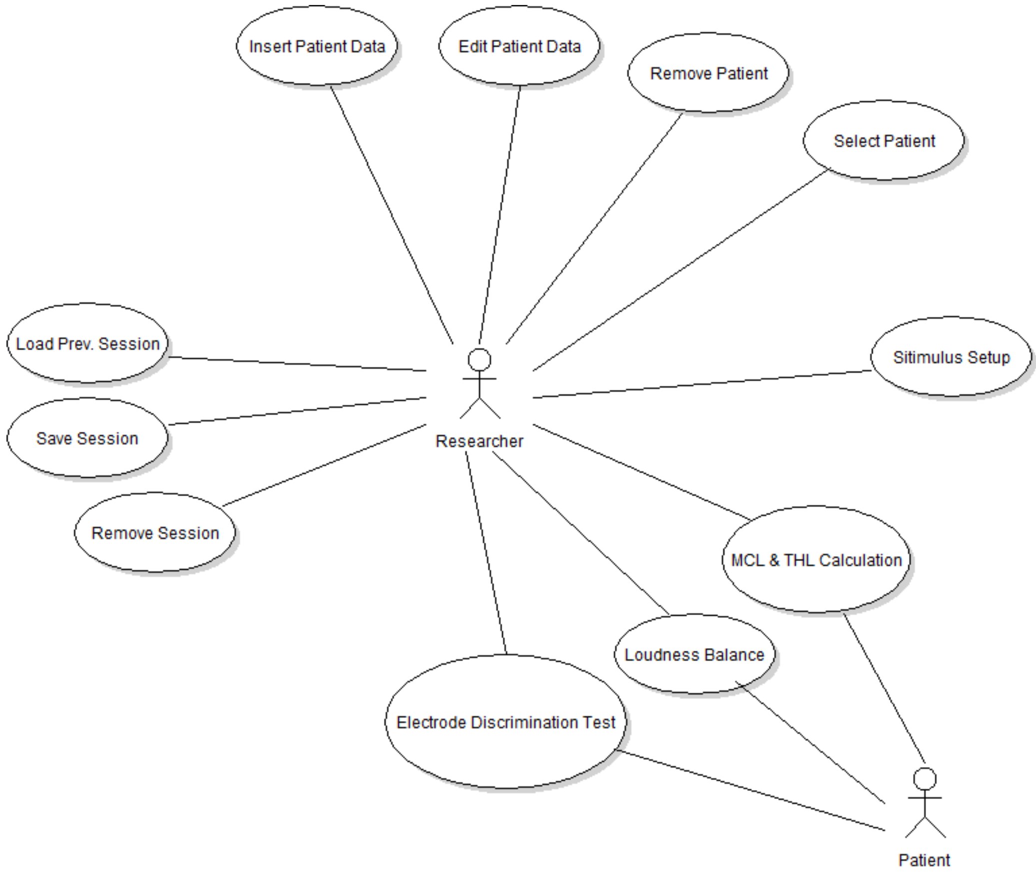

Figure 7.1. Case of Use Diagram. The patient interacts with the researcher by providing answers for each of the test steps: T and C level calculation (MCL and THL, respectively), loudness balance and electrode discrimination test.

Communication between the user interface in Visual Studio and the NIC Python library

Because the NIC library was designed for Python, but the user interface was built in Visual Studio, it is necessary to develop a communication and depuration method between these two platforms. To this end, a file-based communication system was deployed. The data flow from Visual Studio to Python indicates the stimulus parameters: pulse gap, pulse width, rate, phase gap, intensity and duration of stimulus, and the test in progress. The flow from Visual Studio to Python gives information about the connection state of the device to the computer, when it is stimulating, etc.

The (NIC) library for Python, developed by Cochlear LTD. was used to create and send electrical stimuli to the intracochlear electrodes. This is a specific library built to allow researchers to design customized stimuli by changing its parameters: pulse width, pulse rate, phase gap and inter-pulse gap.

To eliminate the need for programming, a user interface was built using Visual Studio, which communicates with Python. In turn, Python scripts control the implant receiver/transmitter by sending instructions to the supplied processor. Thus, the platform development has two main parts: 1) Python functions and 2) user interface development and its communication with Python.

Data Base

The data base has been arranged in such a way that a patient can be assigned to a range of psychoacoustic tests that are controlled by the investigator and test setup and outcomes allocated to an individual patient. That is, the tests are nested within each patient. In figure 7.2, this structure is depicted. Testing generally starts with obtaining Threshold and Comfort levels and loudness balance testing and ends with alternative forced choice (AFC) experiments. It can be seen how for each patient, the user can define as many stimuli as desired, each of which can be associated with different C and T levels. In turn, each C and T level configuration can be matched with many loudness balance tests. Finally, for each of the loudness balance configurations, an unlimited number of alternative forced choice (AFC) experiments can be set up.

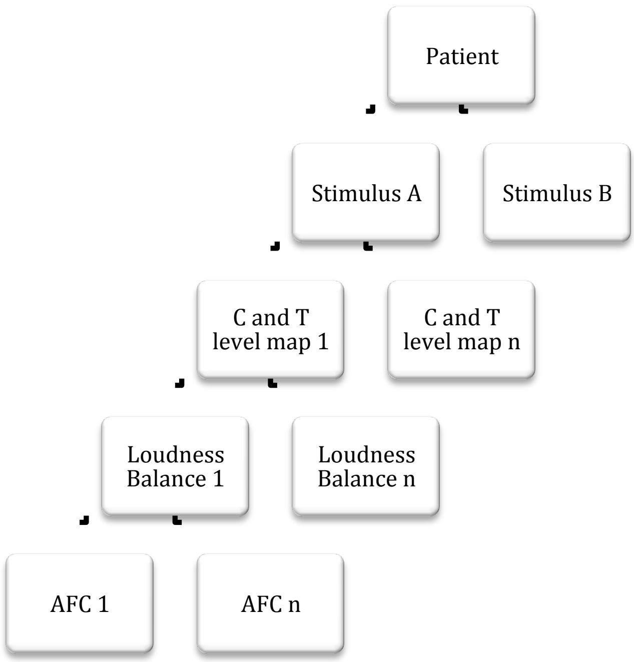

Figure 7.2. Data Base schematics. Tests are nested within their previous required step, the variable patient being the main sorting variable.

User Interface

The user interface was developed using Visual Studio and is thus compatible with Windows 7 and Vista. The computer requirements are: 800 MHz 32/64 bits processor, 1GB of RAM memory, Microsoft. NET Framework 4.5 and Microsoft SQL Server 2008 R2. The application allows the design of psychoacoustic tests where various types of stimulation modes and electrodes can be evaluated without the need to script order commands in any programming language. The files can be exported to an excel file for later analysis.

How it works

Insert Patient

The option *Insert Patient* on the Start page opens the window in figure 7.3, where demographic and clinical data can be inserted. The Comfort (C) and Threshold (T) level values from the patient's standard map can also be inserted. Finally, the location of the electrodes with respect to the modiolus can also be catalogued as perimodiolar, mid-modiolar and lateral wall.

On this window, and all subsequent ones, there is an indicator of the communication state between the application and the processor (connected/not connected) for safety reasons. It will show whether the computer correctly detects the processor or not.

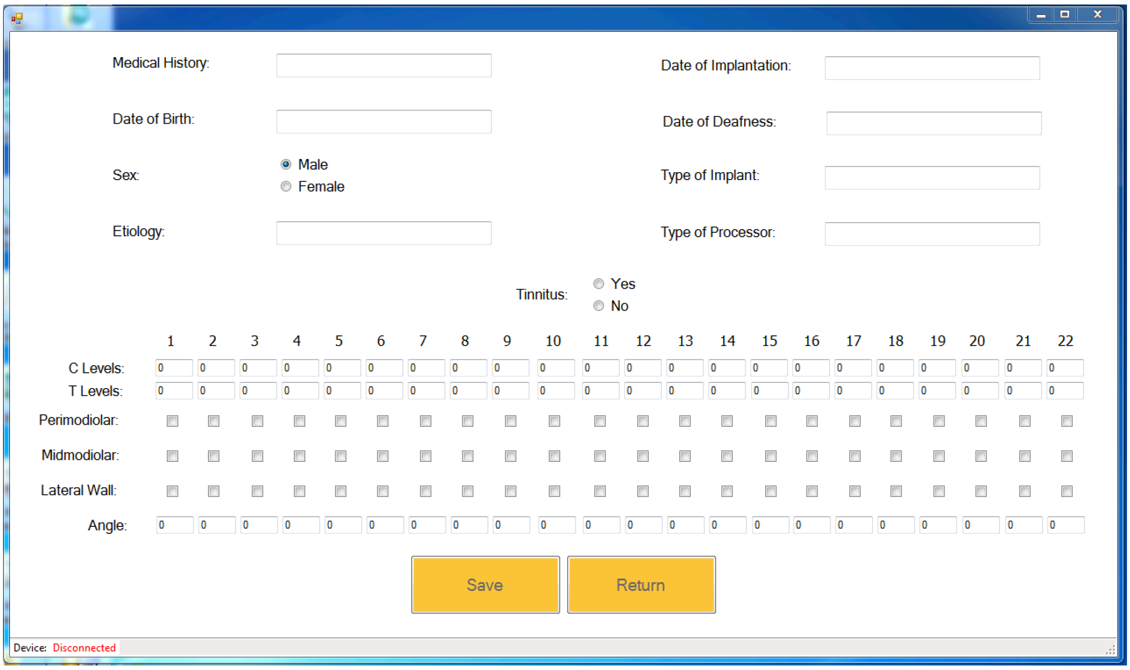

Figure 7.3. Insert Patient. Here demographic, clinical data and electrode position with respect to the modiolus can be registered.

Stimulus Definition

Once the patient has been saved, the Existent Patient window replaces the New Patient, where the same information is displayed, but now the user can select the different tests to perform the study.

A stimulus must be defined before the experiment can proceed. The Define Stimulus button opens a window, figure 7.4, that allows customization of the following parameters: stimulation mode, stimulus duration, inter-stimulus interval, pulse phase and gap durations and pulse frequency. The stimulus consists of a train of electrical biphasic pulses at a frequency specified by the pulse rate option. Stimulation mode refers to the type of ground electrode used: the reference electrode located in the mastoid region (1), the internal processor case (2) or a combination of the two (1+2). The inter-stimulus interval refers to the resting time between two stimuli. These settings will remain fixed for all tests of the study.

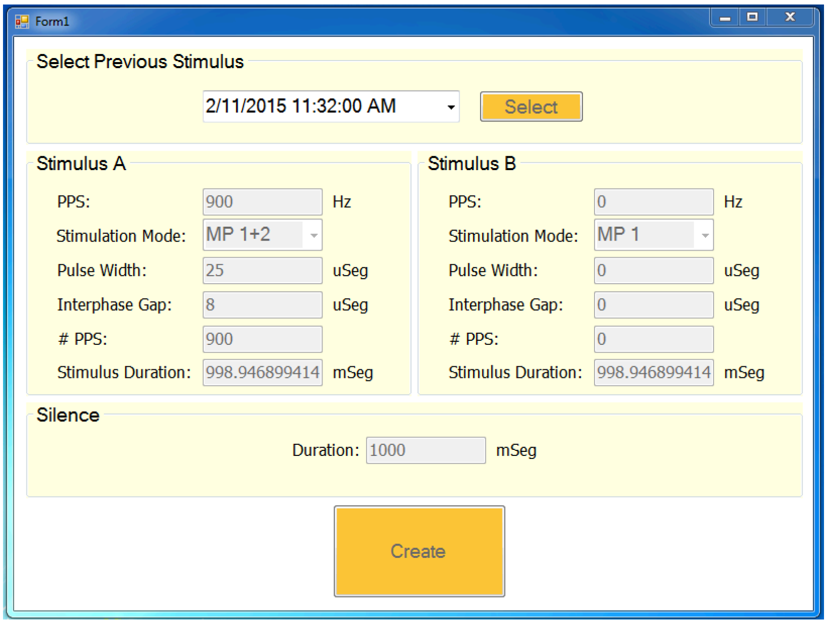

Figure 7.4. Stimulus definition window. PPS: pulses per second; Stimulation mode: ground electrode location (1 for ground electrode in the mastoid, 2 for internal processor case and 1+2 for a combination of both); # PPS: number of pulses per second (defines stimulus duration); silence: defines time lapse between pulse trains.

Dynamic Range Map

The Dynamic Range (DR) of an electrode defines the current intensity span that produces auditory sensation and is determined by the T and C levels of each electrode. The window can be seen in Figure 7.5. The T level is calculated using the classical *ascending and descending method of limits* (116): a stimulus is presented at progressively increasing intensity until the patient reports hearing sensation and then it is lowered down again using smaller intensity steps until it is no longer heard. To find the C level, the stimulus is presented in progressively smaller steps as the intensity rises until the patient reports that the sound is no longer comfortable because it is too loud, but never painful.

In this part of the test, a new map can be created or a previous session can be loaded by selecting one from the *Load Session* scroll. When two different stimuli are used in an experiment, two independent maps must be created using one stimulus at a time.

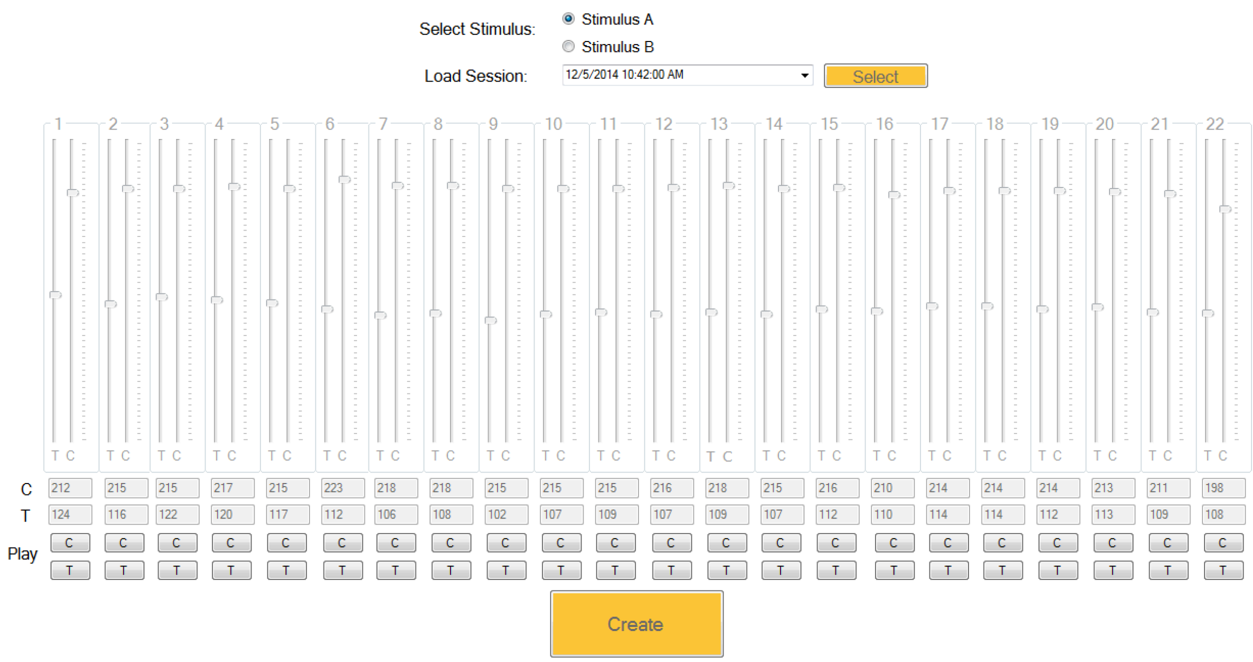

Figure 7.5. Dynamic Range Map. C and T levels for each electrode are determined using the up-down method.

Loudness Balance

The second step is to loudness balance all electrodes so that the stimuli delivered to all electrodes are perceived as equally loud. In the *Loudness Balance* window shown in figure 7.6, the user can compare an electrode with 21 other electrodes. In addition, one can select the stimulus to be used (A or B). The user must select the percentage of DR intensity that the initial reference electrode will have. This will be the electrode that will be objectively fixed at the desired percentage, and all others will be perceptually balanced with respect to it. The number of times one electrode will be compared to another (repetitions) can also be selected.

The method used is the confluence method. The patient is presented with two stimuli, the first corresponding to a reference electrode, and the second to a test electrode. The latter begins randomly above or below 10% of the DR percentage selected prior to test commencement. The patient selects which one of the two stimuli sounds louder using the window in figure 7.7 and the intensity of the test electrode will be lowered or raised until confluence is achieved. Then the patient has to state that they sound equal using the appropriate button. The stimuli will be repeated with the same settings to confirm that the patient perceives them as equal and then the system will automatically start again but with the test electrode being 10% different in the opposite sense. The test electrode final value is the average of the two approaches.

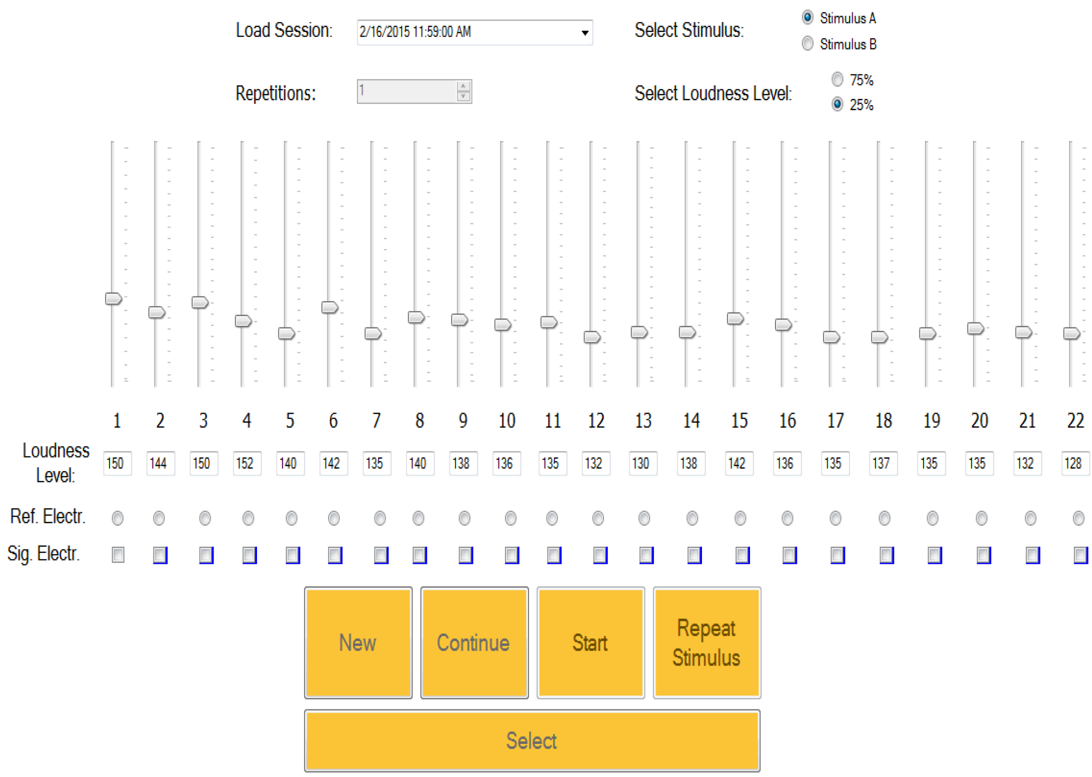

Figure 7.6. Loudness Balance window. Adjacent electrodes are loudness balanced using the confluence method.

Figure 7.7. Patient view for electrode loudness balance. The patient has to state which stimulus sounds louder.

Alternative Forced Choice Experiment for Electrode Discrimination

The chosen method for electrode discrimination is a three-interval, forced-choice procedure (3FC) (figure 7.8). Two of the stimuli come from a selected reference electrode and a third one from a signal electrode. They are presented in random order and each test electrode is presented the number of times stated in the *Number of Repetitions* scroll. The patient has a user interface with three buttons (stimuli 1 to 3) and has to select the one that sounds different. The researcher will select a reference electrode and a number of electrodes to compare it with. Results are given as percentage correct scores.

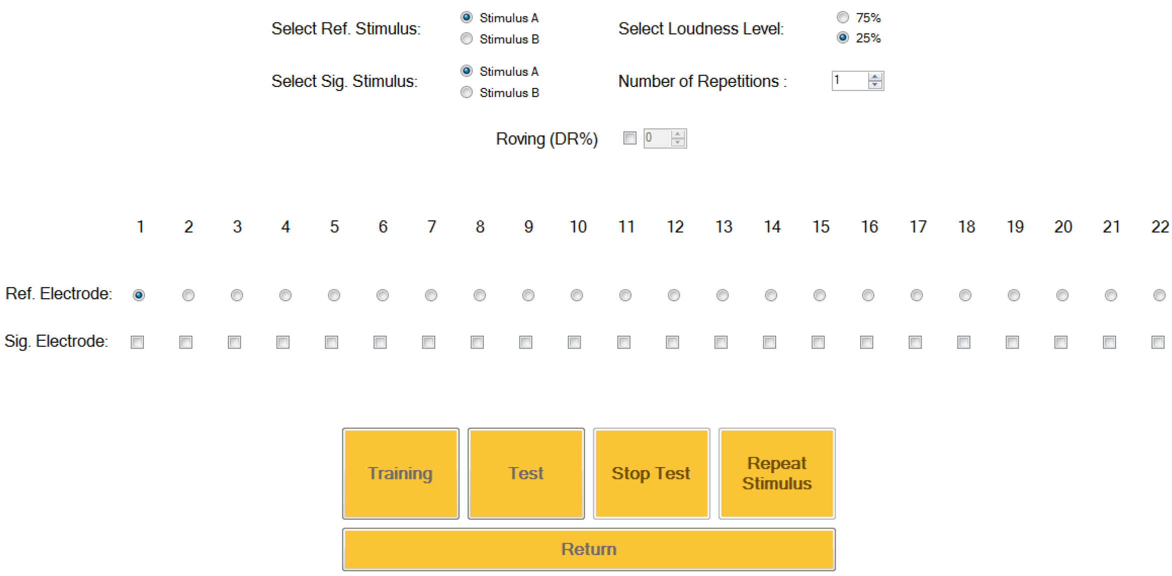

Figure 7.8. Alternative forced choice experiment for electrode discrimination.

7.2.- Pitch discrimination comparison in CI users using perimodiolar vs straight electrode arrays

There is great interest in placing the electrode arrays as close as possible to the modiolus, in an attempt to achieve lower thresholds and hence reduce excess dissemination of current that can result in interaction between channels. Physiological studies and mathematical models suggest that such placement reduces the stimulation current required to activate a neuronal population, increases the dynamic range and results in more localized regions of neuronal excitation. (117) (104) (118) (119)

For this reason, the aim of this study was to investigate the difference in electrical pitch perception between these two types of electrode and test it on subjects carrying both types of arrays. Using the software application described in the previous chapter, patients fitted with the two types of electrode arrays were tested for their ability to discriminate electrical pulses from adjacent and near-adjacent electrodes at low intensity level (matching 25% of the DR of electrode 11).

7.2.1- Introduction

Pitch perception is known to vary regularly and essentially monotonically with the longitudinal position of the electrode stimulated in most patients fitted with straight arrays. However, the predictive model developed by Frijns suggests that in some circumstances, there is a "cross-arousal", where axons from more apical areas are also excited. Thus, besides the local target population, they also stimulated its upper and lower spiral regions. This would result in abnormally low pitch perceptions, especially at high intensity levels. The study suggests that this effect would be more likely in a set of electrodes medially positioned and beyond the basal turn. It is therefore important to investigate how electrical pitch perceptions vary with transverse and longitudinal stimulation sites. (120) (121) (107)

The results of several experiments in which electric field patterns for different electrode configurations have been investigated within the cochlea, reveal that the spatial gradient increases as the distance between the stimulating electrodes and neuronal tissue decreases. Therefore, due to the steeper slope of the electric field, an electrode that is closer to the neuronal endings requires less current injection to reach the threshold current discharge. The result would be lower for sets of electrodes placed proximally thresholds. In addition, electric fields with steeper slopes spread less to adjacent neuronal populations, which decreases the probability of interaction between channels. This can result in a better discrimination between the electrodes. (122) (123) (118) (124) (119) (108) (125)

A number of studies have examined the effects of the proximity of the electrode pacing thresholds. Recent studies comparing the position of perimodiolar and lateral wall electrode arrays have shown lower thresholds in the former. Cohen, Saunders and Clark also measured the ability to discriminate in three subjects with perimodiolar arrays. In two of the three subjects, a portion of the electrode was located near the modiolus. They showed better pitch discrimination for electrodes in this region. These results suggest a better spatial selectivity and lower thresholds for electrodes located closer to the neuronal tissue. (126) (127) (128) (129)

Shepherd et al. recorded lower electrical auditory brainstem responses (EABR) and a smaller slope of the growth function's amplitude for electrodes placed close to the modiolus and to the spiral ganglion, as opposed to electrodes placed near the lateral wall or the middle portion of the scala tympani. Liang et al. showed that lower thresholds were associated with better spatial selectivity of the electrically stimulated auditory nerve fibers in cats. They also found great variability in the response of auditory neurons to electrical excitation and suggested that the nerve fibers closer to the stimulating electrode are the most selective. Finally, Cohen et al. described narrower width excitation profiles for implanted subjects with perimodiolar arrays than for the subjects implanted with straight arrays, as measured by evoked compound action potentials (ECAPS). The width of the ECAP measure was significantly correlated with the distance of the electrode band from the modiolus, though the width of the forward masking profile was not. They also found that the more intense the stimulus the wider would the spread of excitation be and the less relevant the distance of the electrode from the modiolus. In short, these studies suggest that a closer distance to the modiolus results in lower excitation thresholds and less overlap. (104) (130) (131)

However, the selectivity of the electric fields regarding neural stimulation depends on a number of factors apart from the electrode-nerve proximity. These factors include patterns of neuronal survival population, stochastic variability characteristics of individual nerve fibers, cochlear anatomy, signal characteristics and design features of the electrode. Many of these factors may be interdependent and may affect the threshold as well as the selectivity. For example, an individual with poor nerve survival will have poorer selectivity and higher thresholds than one with good nerve survival, regardless of its electrode array, as he requires higher values to achieve optimal neural excitation. (110)

It is believed that speech perception with a cochlear implant requires good resolution in the spectral and temporal domains. However, the relationship between speech perception and spectral resolution is not entirely clear. Several studies have shown that there are significant correlations between processing capabilities of spectral patterns and speech perception in CI users. However, the intensity of the electrical stimulus has been found to correlate directly with performance. Pfingst et al. studied the effects of stimulus level on electrode spatial discrimination. They found that there was a weak trend towards better discrimination for the highest levels of stimulation. McCay et al. also obtained similar results using only patients with Nucleus 22 (Cochlear LTD.). These results seem contradictory; since a higher power level also means a wider excitation region, potentially, this produces an overlap between two adjacent stimulation sites, reducing spatial discrimination. (132) (125) (108)

McKay *et al.* found only small improvements in electrode discrimination between 40% and 70% of the DR (133). This is consistent with later findings indicating that the largest changes in electrode discrimination occur below 40% of the DR (134). Pfingst *et al.* also showed moderate improvements in electrode discrimination with increasing level in some instances (135). Therefore, it is of capital interest to study how perception happens at low DR percentages and the importance of electrode distance to the modiolus at low intensity levels below 40%, which have been much less studied.

7.2.2- Objective, Hypotheses and rationale for the study

The objective is **to give evidence that perimodiolar electrodes give better electrode discrimination at low intensities because they are closer to the modiolus**. To this end, electrode discrimination abilities for users of both straight (422 cochlear) and perimodiolar (Cochlear CA (RE)) arrays have been studied using a platform developed specifically for this task.

The hypothesis addressed in this study are: 1) **patients implanted with perimodiolar electrodes have better electrode discrimination for low (25% of the DR) intensity stimuli than carriers of straight, lateral wall electrodes** and 2) **patients implanted with perimodiolar electrodes have lower impedances**.

This study has been conducted to investigate whether the results observed in previous studies are the cause of a compromise between the stimulus intensity needed to trigger a significant neuronal population and the subsequent spread of excitation along the region. It aims to clarify if for low intensities perimodiolar electrodes, which yield more focused stimulation without incurring into spatial overlap, provide its users with better electrode discrimination.

7.2.3.- Subjects and Methods

Subjects:

The total sample comprised 10 cochlear implant carriers: 5 of them using a Cochlear Nucleus CI 422 slim straight electrode array and 5 using Cochlear Nucleus CI 24RE (CA) perimodiolar electrode array. The age range was 36-77, with 2 women and 8 men. They are patients of the Hearing Loss Unit of the Department of Otolaryngology of the Complejo Hospitalario Universitario Insular Materno Infantil of Las Palmas de Gran Canaria, between December 2014 and December 2015. The following demographical data were collected: age of implantation, date of profound hearing loss diagnosis at the clinic, type of electrode array and speech processor, aetiology of deafness, date of birth and sex (table 7.1). The C and T values of their stable map, impedance and neural response telemetry (NRT) recordings were also collected.

Patient	Sex	Year of Birth	Implantation Date	Diagnosis	Electrode Array
1	Male	25-Sep-1958	21-May-2004	1989	Nucleus CI 24RE (CA)
2	Male	17-Sep-1966	30-Dec-2011	2010	Nucleus CI 24RE (CA)
3	Male	28-Mar-1939	21-Sep-2011	2010	Nucleus CI 24RE (CA)
4	Male	13-Dec-1957	17-Feb-2009	2008	Nucleus CI 24RE (CA)
5	Female	14-Dec-1977	31-Jan-2012	2007	Nucleus CI 422 slim straight
6	Male	08-Dec-1957	24-Feb-2015	2014	Nucleus CI 422 slim straight
7	Female	09-Nov-1949	05-May-2015	2010	Nucleus CI 422 slim straight

8	Male	12-Mar-1980	24-Mar-2015	2012	Nucleus CI 422 slim straight
9	Male	23-Jun-1976	17-Jun-2014	2013	Nucleus CI 24RE (CA)
10	Male	12-May-1960	16-Dec-2014	2010	Nucleus CI 422 slim straight

Table 7.1. Demographic data

The inclusion criteria were:

- Adults over 18.
- Postlingual bilateral hearing loss from severe to profound. (Average Tonal audiometry thresholds: greater than 70 dB).
- Absence of retrocochlear pathology and without central auditory processing disorders.
- Stimulation rate of 900 pps and pulse width of 25 us in their standard fitting map, with a minimum of 6 months of use of their sound processor.
- SPEAK or ACE strategy.
- Stable fitting map.
- A minimum of 18 operational channels.
- 50% of speech understanding for sentence tests in silence without lip-reading using the CI at 65 dB HL.
- Full insertion without Scala Vestibuli dislocations.
- Informed consent to participate in the study.

Methods

The subjects were instructed on what to do before commencement of each stage of the experiment, each of which took one session to complete. In turn, each session took place once a week during a one-month period. The procedure follows the same order described in the platform design chapter. First, C and T levels were determined for the stimuli created for this experiment. Then the electrodes were loudness balanced and, finally, the alternative forced choice experiment was completed.

Since electrode discrimination can be affected by ossifications and neural survival, both impedance and NRT variables were also collected from each patient's medical history.

Stimulus Characteristics

In order to isolate the effect of electrode positioning within the cochlea as a contributing factor in electrical pitch discrimination, the stimulus parameters were set so that they were those used by patients in their standard, daily maps. Accordingly, the stimulation rate used for this test was 900 pps, the pulse width 25 us, the phase gap 8 us, the pulse gap 1s and the stimulation mode MP1+2 (meaning that both the reference electrode in the mastoid and the one embedded in the internal receiver/transmissor were used). If such parameters had been changed, the speed at which each individual adapts to a new way of stimulation could have played a major role in electrode discrimination, thus hindering the effects sought out in this experiment.

Step 0: selection of electrodes to be used

All active electrodes were used to get the data from the largest possible number of electrodes and hence correlate the position of the electrode with its psychoacoustics results. Despite belonging to a perimodiolar guide, some electrodes may actually be as far removed from the modiolus as perimodiolar ones. Those at the basal end of the cochlea might be far away because of the insertion region (cochleostomy) being at a certain distance from the modiolus. At the other end, the diametre of the Scala Vestibuli is so much reduced that it might cause both perimodiolar and lateral wall electrodes to be positioned at very similar distances to the modiolus. It is therefore interesting to contrast psychoacoustic results with the position of each electrode by means CT imaging.

Step 1: Determination of C and T levels

Threshold levels for each electrode were determined using the specific stimulus created for this study. The method of obtaining these levels was the classical *ascending and descending method of limits* described in the previous chapter. (116)

This procedure took one session split into two or three sub-session, each lasting around 40 minutes with 10-15 minute breaks in between.

The test started at the apical electrodes and moved towards more basal ones. Sometimes, when middle electrodes were reached, the first electrodes were retested, as patients tend to be scared by intense high pitch sounds at first but grow more confident after a few tests, when they realize that it is not an uncomfortable procedure.

Step 2: Loudness balance of all electrodes

The loudness of the electrodes must be balanced to minimize interference effects. It has been shown that for a given electrode, a difference in volume can cause the illusion of a difference in electrical pitch. To avoid this, each electrode must be adjusted to sound as intense as their neighbours. There are numerous techniques to establish optimum loudness balance. There is always a compromise between speed of testing (to prevent fatigue and saturation) and accuracy. Since the length of the experiment causes great inconvenience to patients who have to put off work to come to the laboratory three times, an adapted staircase method was used. (136)

The process starts in the central electrode (electrode 11). This first electrode acts as the starting point. The software sets the reference electrode to a loudness level of 25%, and the intensity of its adjacent neighbor (towards the apical end) is initially assigned a value 10% above or below 25% of its dynamic range at random. The patient listens to the two sounds sequentially and must answer the question: which sound is louder? ", The answer being either 1st or 2nd. The response is recorded using the buttons on the interface to lower or raise the signal electrode by 1 CL until the patient perceives them as equally strong. At this point, the same sound is replayed to verify that this is the case and then the test is reset but starting with the opposite situation. That is, if the target electrode started 10% above 25%, this time it will begin below 10%. The equal loudness point is set as the average of both approaches. Subsequently, the target electrode will serve as the reference electrode for the next adjacent electrode. Hence, the same procedure is repeated until the apical end is reached and then the process starts again from electrode 11 towards the basal end of the array.

The balance is set between the closest electrode pair for two reasons: 1) Discrimination is most difficult at this spatial distance than any other, so their stimuli need to be as similarly intense as possible and 2) because the patients reported great difficulties and even inability to establish equal loudness between two very different electrical pitches.

Step 3: Test electrode discrimination

Electrode discrimination was measured using the three-alternative-forced-choice method. After selecting the central (reference) electrode and its two nearest neighbors on either side (signals), the number of times that each electrode will be tested with the central electrode was set to three. Thus, three electrical stimuli were sent to the implant, two of which came from the same electrode (reference) and one from the signal electrode. They were presented in random order. The patient had to indicate the stimulus that sounded different from the other two.

After all electrodes were tested with the reference, the test finalizes and the researcher had to select the apical, most adjacent electrode as reference and its two nearest neighbours on either side as signals. The process was repeated until all electrodes had acted as references. This way all electrodes were tested against 4 nearest neighbours, except for the apical and basal most electrodes, which could only be tested to one side. A similar situation holds true for the ones before these, because they have two electrodes on one side but only one on the other. The results were given as percentage of correct answers.

Imaging

Plain X-ray Stenvers projections have been used in the past as a standard for electrode array location within the cochlea, but nowadays it has been abandoned (137). Many recent studies consider the benefits of high-resolution Computer Tomography (CT) scanning as a more exact method to visualize the structures of the temporal bone. The advantages of CT scanning include the elimination of respiratory and other motion artefacts and an obvious improvement in the diversity of tissue that can be observed (138). The main disadvantage of CT in the postoperative assessment of a cochlear implant is image degradation, due to partial volumes and the metallic artefacts that may interfere with the visibility of individual electrodes (139). In this study, a temporal bone CT scan with a z resolution of 0.062 mm was used.

To overcome this limitation, a computer-based system was developed to obtain the distance between each electrode in the array and the inner wall of the cochlea. A MATLAB script was made to get an estimate using two images of a CT scan. High-density (i.e metallic) objects were highlighted in the first image, so that the contours of the implant became visible. In the second one, an average density with a high contrast was used to see the inner wall.

Once the images were imported, the first image was used to mark a series of points (as many as necessary are created) that follow either: 1) the inner contour of the perimodiolar array, because the contour advance implant has half-banded electrodes that are only located in the inner side of the array; or 2) the middle portion of the straight array, because in this array the electrodes are a full ring. Once the points were set, the system generated a parametric, interpolating function to calculate the curved shape of the implant. Then the same procedure was repeated for the second image, but marking the inner wall instead.

Once generated, the inter-electrode distance specification of the manufacturer is used to determine individual electrode location. Then the distance to the inner wall was calculated as the perpendicular distance to the tangent of the curve at the electrode location. The final measures are displayed on the second image (Appendix IV).

Data analysis

Analysis of the variables under study was done using *IBM SPSS Statistics for Macintosh, V. 21.0*. (IBM Corporation, New York, United States) and eViews . First, descriptive statistics for the endogenous and all exogenous variables were done. Then a linear regression model, a logit and a binomial logit model were calculated to study the correlation between the variable and its predictors.

7.2.4.- Results

Four perimodiolar electrode array carriers (patients 2-4 and 9) and 4 straight array carriers (patients 5-8) completed the test. Patient 1 did not attend the CT scan appointment and patient 10 had an inappropriate MRI technique for an unrelated health issue and his array was displaced so he had to undergo a repairing surgery and so could not complete the test.

Consequently, a total of 1968 trials were used for the analysis. Of all these, 1169 were correct answers and 799 were not. Trials were grouped into units of observations. A unit is the result of first averaging the success of a given reference electrode with each of its nearest and second nearest neighbours (4 for electrodes 3 to 20; 3 for electrodes 2 and 21 and 2 for electrodes 1 and

22) and then further averaging these into a single value. The success rate is thus the quotient between the number of successes by the number of trials, for a given reference electrode (6 trials for electrodes 1 and 22, 9 trials for electrodes 2 and 21 and 12 trials for the rest of electrodes). Its descriptive statistics (Table 7.2), the Jarque-Bera normality test, together with the histogram of frequencies are shown in figure 7.9. From the latter it is concluded that normality cannot be rejected at 4.60% significance (p=0.046) and hence the variable behaves has a near Gaussian distribution, with a success rate of 59.70%.

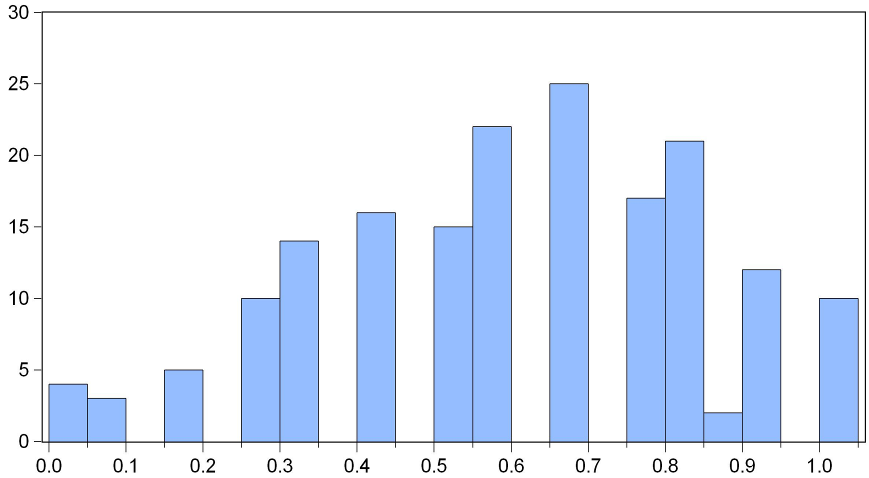

Figure 7.9. Success rate histogram representation

Descriptive Statistics	
Mean	0.597
Median	0.583
Maximum	1
Minimum	0
Standard Deviation	0.247
Asymmetry	-0.373
Kurtosis	2.470
Jarque-Bera Test	6.141
Jarque-Bera associated probability	0.046

Table 7.2. Descriptive statistics.

-

To identify which factors have the capacity to influence and modify the success rate, we started by contrasting whether the electrode number could be one of them. A Levene's test for equality of variances of the mean success rate indicates that the variance is the same for all electrodes (p=0.35, the null hypothesis which states that the variances are different cannot be rejected). Consequently, an ANOVA contrast was also done, concluding that the equality of the means of the successes for all electrodes cannot be rejected (The F statistic value had an associated p= 0.82).

A similar analysis of the variance was conducted for the patient factor. The result suggests the existence of statistically different results depending on the patient, both for mean and variance values (probability associated with the Levene statistic of equality of variances p=0.0001, and probability associated with the F statistic of Welch of equality of variances p=0.000). Accordingly, the empirical evidence supports the hypothesis that there is a patient effect that significantly affects success rate. Figure 7.10 shows the standard deviation plot of the mean success rate for each patient.

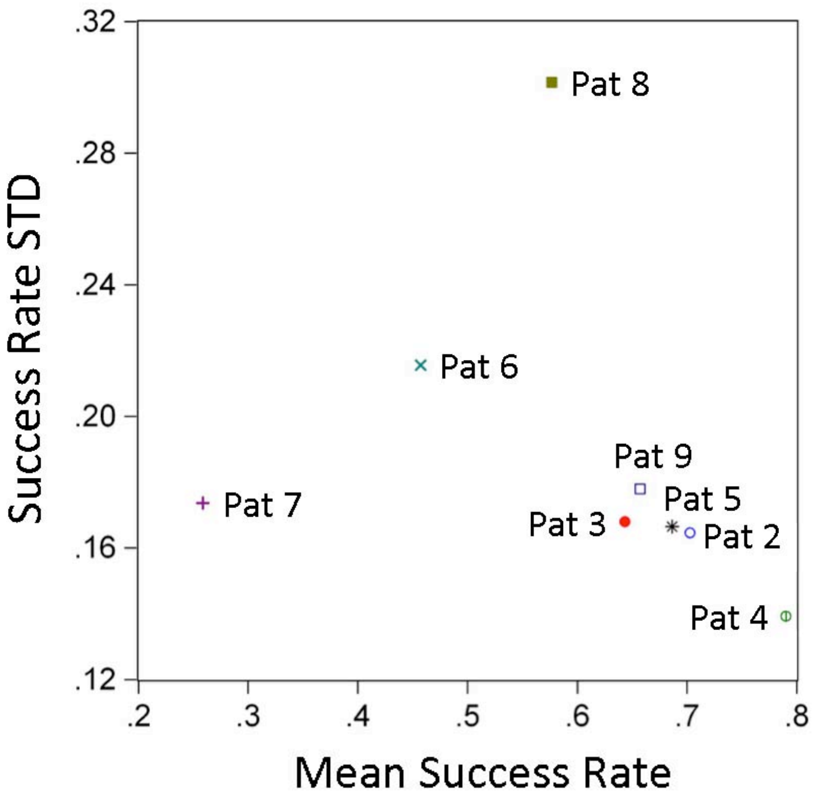

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Figure 7.10. Mean success rate values and standard deviations per patient (pat).

As can be seen from the chart, there are four patients that stand out from the rest. Patients 6 and, most prominently, 8 have a large dispersion. In addition, patient 6 also has a low mean success rate, although not as low as patient 7, which is the third patient that stands out from the sample. On the opposite end is patient 4, with the highest success rate and the smallest dispersion between electrodes.

Of these patients that have a differential behaviour, three of them, patients 6, 7 and 8, have straight electrode arrays (although so does patient 5). In fact, the data shows a statistically significant difference in success rates between patients implanted with straight and perimodiolar arrays. The probability associated with the Welch's F statistic (because variances between groups were different) for the equality of the probability of success in the presence of different variances also gave a value of 0.000, being the probability of success 53% for straight electrode array carriers and 70% for perimodiolar electrode array carriers. This result provides empirical evidence in favour of the hypothesis that perimodiolar electrode arrays provide better electrode discrimination.

The mean effect of distance on success rate in average terms is evident when these two variables are plot against each other (figure 7.11). Patients whose electrodes are further away from the inner wall are those with a lower success rate, on average.

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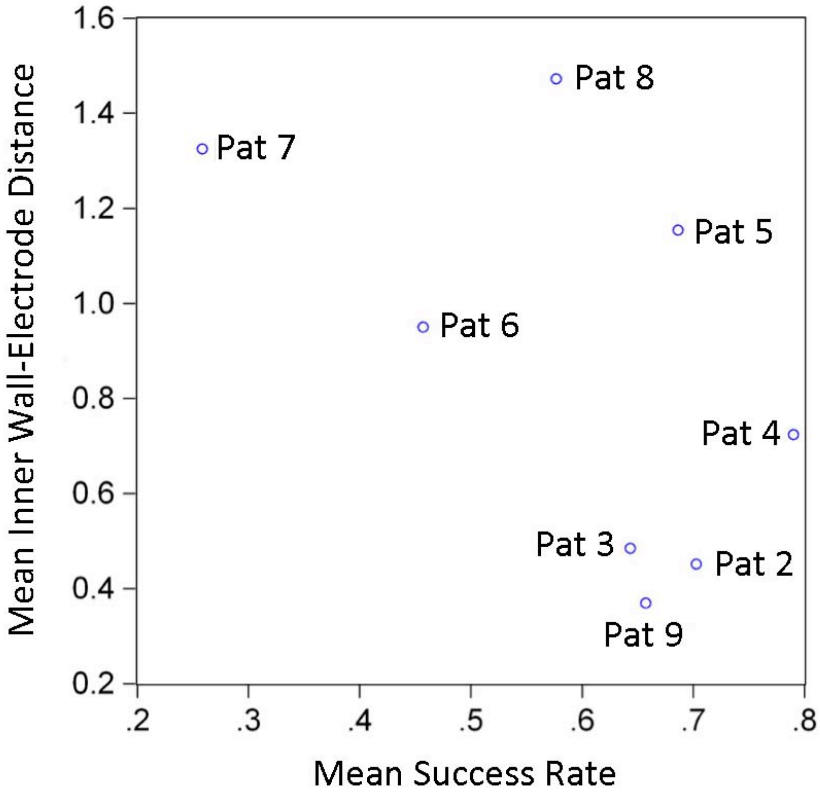

Figure 7.11. Relationship between the mean electrode-inner wall distance per patient and mean success rate.

In any case, the above results must be taken as indicative,-not conclusive-, for several reasons. On the one hand, one cannot differentiate the patient effect from the distance effect on success rate. On the other hand, the results are based on this variable having normal behaviour. However, success is a binomial variable. The raw output data consisted on a matrix of ones (success) and ceros (failure). In fact, even though the Jarque-Bera contrast does not reject the null hypothesis of normality of the success rate, other contrasts reject such behaviour for significance levels of less than 1% (Kolmogorov-Smirnov, Shapiro-Wilk).

A correlation and regression analysis was done to overcome these limitations and to search for greater precision in the measurement of the relationship between the distance of the electrode to the inner wall of the cochlea and the success rate. To this end, the following additional variables were also incorporated:

1. **NRT**: quality of the nerve measured in CL. A higher value indicates a worse state of the nerve, as more current needs to be delivered to the nerve end to elicit a measurable response. Thus, it is expected that the increase in the value of this variable results in a decrease in the probability of correctly discriminating between electrodes.
2. **T level**: minimum current needed to elicit hearing sensation, measured in CL. According to the literature, this variable is expected to have a positive relationship with success rate.
3. **Impedance**: measured in kΩ. Because larger impedances result in higher currents required to reach the neural ends, this variable is expected to have a negative effect on the success rate.

To do this analysis, as indicated earlier, the sampling units are all the different electrode-patient combinations, i.e. electrode 1 of patient 1, electrode 1 of patient 2, etc. The descriptive statistics of the variables are shown in table 7.3. On average patients have been successful in 60% of the tests, have an electrode distance of 0.87 mm, a mean NRT of 183.61 CL, a T level of 126.29 CL and an impedance of 9.66 kΩ. The relative dispersion is moderate in all variables, being the variable distance the one with greater relative dispersion. Because the variable deterioration of the nerve (NRT) was not measured in 12 electrodes, the sample size had to be reduced to 158 elements.

Table 7.4 contains the correlation matrix of the variables and their level of significance. It can be concluded that the variable of interest, the success rate, is significantly correlated with all explicative variables, and with the expected signs as well. There are also significant correlations between the different explanatory variables, which make it difficult to quantify their individual effect on the mean success rate. However, their values are not high enough to produce serious problems of multicollinearity.

	Success Rate	Electrode-inner wall distance	NRT	T level	Impedance
Mean	0.60	0.87	183.61	126,29	9.66
Maximum	1.00	2.09	237.00	176.00	20.33
Minimum	0.00	0.11	109.00	95,00	4.23
Standard Deviation	0.25	0.50	22.06	18.56	3.24
Relative dispersion	0.41	0.58	0.12	0.15	0.34
No. of Observations	176	176	158	176	176

Table 7.3. Descriptive statistics of the variables: electrode-inner wall distance, NRT, T level and impedance.

	Success Rate	Electrode-inner wall distance	NRT	T level	Variance Inflation Factor
Electrode-inner wall distance	-0.207***				1.23
NRT	-0.294***	0.421***			1.87
T level	0.169**	0.260***	0.521***		1.64
Impedance	-0.318***	-0.015	0.217***	-0.222***	1.27

* Significant at 10%, ** significant at 5%, *** significant at 1% for bilateral contrasts

Table 7.4. Pearson correlation of the variables: success rate, electrode-inner wall distance, NRT, T level and impedance.

The determining factor when modelling the endogenous variable is the very nature of the variable itself. Success rate is bounded between 0 and 1 and its theoretical behaviour corresponds to a Bernouilli distribution. Let us remember that the success rate has been obtained as the quotient between the number of right answers divided by the number of trials per electrode pairs for a given reference electrode (6, 9 or 12). When calculated in this way the variable has a binomial behaviour. Therefore, the assumption of linear relationship with the explanatory variables, which implies a constant linear effect, is not strictly valid because linear models give values below zero and above one. Moreover, the variance of the success rate is not constant and so there is heterocedasticity. These properties limit the use of the standard method of multiple linear regression and its estimation by ordinary least squares and thus we must resort to alternative methods before giving final conclusions.

Three predictive methods were used, all of them using the explanatory variables described above. First, success rate was modelled using the method of minimum chi-squares. To adapt the model to the nature of the variable of interest, two more models are proposed, both taking the logit transformation of the success rate as the endogenous variable.

Linear probability model estimated by the method of least chi-squares (MCSM)

The first model describes the probability of success as a linear function of the predictive variables and the weighted least square estimation of the following form (140)**:**

$$p_i = \frac{n_i}{N_i} = \beta_0 + \beta_1 x_{1i} + \beta_2 x_{2i} + \cdots + \beta_k x_{ki} + u_i \text{ [7.1]}$$

This is the simplest method and the easiest to interpret, but is not without serious limitations. *pi* is the success rate (it is neither the number of successes, *nor* the number of tests); $\{x_1,\ldots,x_k\}$ are the set of predicting variables; u_i is the random perturbation; β_i are the parameters of the model that quantify the effect of each one of the predictive variables on the success rate and *i* relates to each of the sampling units (patient-electrode). Given the nature of the endogenous variable, the model has a problem of heteroscedasticity. Therefore, the data was weighted using a new variable, *Pond1*, defined as:

$$Pond1 = \sqrt{\frac{N_i}{p_i(1-p_i)}} \text{ [7.2]}$$

-

In order to do this estimation, the variable success rate (p_i) was re-scaled to take values in the range (0-1) instead of in the interval [0-1]. Following Maddala, the expression $p'_i = p_i + (2N_i)^{-1}$ was used instead of p_i. (140)

Predictive variables used were distance (of the electrode to the inner wall), NRT, impedance and T level. Table 7.5 summarizes the model parameters.

Predictive Variables	**MCSM Estimated Coefficients**	**MCSM Typified Coefficients**	**MLCSM Estimated Coefficients**	**LBGLM Estimated coefficients**
Constant	1.454 (0.000)	1.138(0.082)	1,884(0.08)	1.184
NRT	-0.008(0.000)	-0.010(0,037)	-,018(0.001)	-1.846
Impedance	-0.042(0.005)	-0.049(0.044)	-,053(0.041)	- 0.499
T level	0.009(0.000)	0.014(0.003)	,021(0.000)	1.797
Distance	-0.149(0.006)	-0.314(0.034)	-,336(0.03)	- 0.323
			-2.385	

Table 7.5. Estimation using the MCSM, MLCSM and for the variable success rate.

The estimators are shown in the second column along with the probability associated with the contrast whose null hypothesis is that the corresponding variable has no ability to explain the success rate. As can be seen from this table, each of them explains the success rate in a statistically significant fashion. In addition, all estimators had the expected signs. Focusing on the variable Distance, the model indicates that, on average and ceteris paribus, an increment of 0.1 mm in the distance reduces the rate of success in 0.0149 percentage units. Table 7.5 also shows, by means of the typified coefficients, that this has the lowest relative influence of all variables. In relative terms, a low NRT, a low impedance and a low T level are more determinant for success rate than distance to the inner wall. The correlation between the predicted and the real success rate was also calculated and showed that, as a whole, the four explicative variables explain 25.1% of the original variable.

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Minimum Logit Chi-Square Method (MLCSM)

The previous model is easy to interpret, but has some shortcomings that limit its use. The main one is that its marginal effect is constant. This means that, whatever the distance of the electrode to the inner wall, a unitary increase of this distance always produces the same effect on the success rate. Another problem is that physically unrealistic results above 1 or below 0 are obtained with the linear method.

To solve these problems, Berkson proposed in 1953 the MLCSM (141). This method takes the logit transformation of the endogenous variable and models it as a linear function of the predicting variable. The estimation is done using weighted least squares. It has the form:

$$\ln\left(\frac{p_i}{1-p_i}\right) = \beta_0 + \beta_1 x_{1i} + \beta_2 x_{2i} + \cdots + \beta_k x_{ki} + u_i \quad [7.3]$$

This transformation of the variable of interest ensures that the estimated values of the success rate are always in the interval [0.1]. As in the previous method, the variable success rate has been re-scaled to the interval (0.1). However it can be demonstrated that it still has heteroscedasticity problems (140). This is again solved using weighted least squares. In this case the weighting used was *Pond2*.

$$Pond2 = \sqrt{N_i \cdot p_i(1-p_i)} \quad [7.4]$$

Therefore, the estimation method will be to apply Ordinary Least Squares to model 7.3 previously weighted by equation 7.4. The estimators obtained are optimal unbiased linear estimators (142).

The first thing to note with this model is that the marginal effects of a predictor variable are not constant. They will depend on their starting values. What is true is that if the sign of the parameter of a predictor variable is positive, this implies that if the variable is incremented, the probability of success is also going to increase and viceversa (in terms of mean, ceteris paribus).

To interpret these models, the concepts of ODD and ODD ratio are often used. The ODD is the ratio between the probability of guessing right and the probability of failing for a combination of predictors. That is to say, if for a certain combination of values of the variables the predicted success rate is 0.8, the ODD for that individual is equal to 0.8/(1-0,8)=4. Therefore, for that individual the probability of guess right is four times the probability of failing. For two individuals, 0 and 1, the ratio of their ODDs for a variable x_i can be demonstrated to be:

$$\frac{ODD_1}{ODD_0} = e^{\beta} \quad [7.5]$$

β being the parameter of the predictor x_i in equation 7.3. This quotient is called the ODD ratio. The conclusion is that the ODD ratio calculated between two situations that only differ in a unit of one variable is a constant value.

From the previous result, the expression:

$$\left(e^{\beta} - 1\right) \cdot 100 \quad [7.6]$$

is a constant value and measures the percentage change of the ODD, when x_i is increased one unit. It can also be demonstrated that, to measure the effect of the increase of $1/n$ of x_i, the appropriate expression is

$$\left(e^{\beta/n} - 1\right) \cdot 100 \quad [7.7]$$

Table 7.5 shows the weighted least squares estimation model parameters using the weighting *Pond2*. The variables are individually significant at 5%, but this is not always the case at 1%. With the purpose of comparing this model with the previous one, a correlation was done between the observed and estimated success rate. The result was that this model explains 25.3% of the original variable.

The interpretation of the estimators in table 7.5 for MLCSM is more complex than the one for the previous model. What is straightforward is that the signs are the same in both. The interpretation of the estimated coefficient of the variable Distance is done as follows. The estimated parameter of the distance is -0,314. Therefore, ceteris paribus and on average, if the distance is increased by 0.1 mm, the expression 7.7 takes the value -3.1. This means that due to that 0.1 mm increase in the distance, the probability to guess right against guessing wrong is reduced by 3.1%.

It should not be interpreted as a change in the success rate though. If the success rate at distance d_0 is p_0, then the ODD is $ODD_0 = p_0 / (1 - p_0)$. If the distance is increased 0.1 mm then β becomes β /10. Solving for p_1 gives the result:

$$ODD_1 = \frac{p_1}{1-p_1} = e^{\beta} ODD_0 \quad [7.8]$$

$$\boldsymbol{p_i} = \frac{e^{\beta/10} ODD_0}{1+e^{\beta/10} ODD_0} = \frac{e^{\beta/10} \frac{p_0}{1-p_0}}{1+e^{\beta/10} \frac{p_0}{1-p_0}} \quad [7.9]$$

β being the parameter of the variable electrode-inner wall distance in equation (7.3).

The last equation shows the relationship between the success rate at distance d_0 and at distance $d_1 = d_0 + 0.1$ mm. For example, if for d_0 the success rate is 0.8, a distance increase of 0.1 mm would cause the success rate to fall to 0,795, ceteris paribus and on average.

Logit Binomial Generalized Linear Model, estimated by robust maximum likelihood estimation and with over dispersion correction (LBGLM)

The difference between this model and the previous is that the former takes into account that the variable of interest is binomial and keeps the logit transformation (so the interpretation of the parameters of the model is similar to the earlier model) (143). When working with a binomial distribution, the existence of heteroscedasticity is already considered in the inference process, which in this case is done using maximum likelihood estimation using iterative procedures. In spite of this, it incorporates the robust estimation of the variance-covariance matrix of disturbance and correction for overdispersion.

The parameters estimated using this method are shown in table 7.5. All variables are statistically significant at 5%, and so is the model in global terms (omnibus contrasts yielded an associated probability equal to zero). As for previous models, to obtain the goodness of the fit, the probability of success with this model and then its correlation with the original variable were calculated. The model explains 26,52% of the observed results. Complementary analysis (normality of residuals, absence of outliers, etc.) confirms that these estimators presented better properties than those obtained previous models.

Numerically, the results do not differ from those obtained in the Logit model. On average, ceteris paribus, an increment of 0.1 mm between the electrode and the inner wall of the cochlea reduces the chance of success with respect to failure 3.3%.

Comparative evaluation of model results

Figure 7.12 shows the probability of success estimated for each one of the three methods. Two cases have been depicted: a patient with a good success rate, such as patient 5 (in blue) and the patient with the lowest success rate, patient 7 (in red). It can be seen how the LBGLM model has the most appropriate shape, showing ceiling and floor effects for perfect performance (value one) and lowest performance (value 0). This model also shows greatest variability in performance as distance decreases in the case of patient 7. Thus, this patient could potentially get a 25% improvement in electrode discrimination if an electrode located at 1mm from the inner wall would be placed in contact with it.

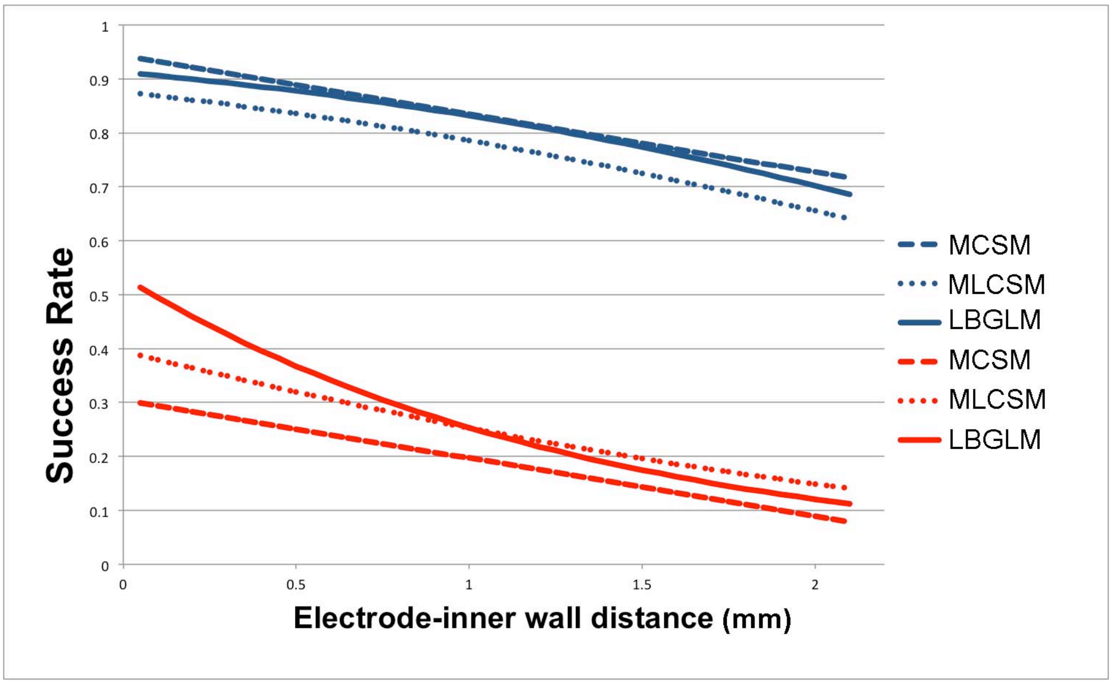

Figure 7.12. Comparison of performance between an average and a low score patient using the three models calculated. The LBGLM model has the most appropriate shape, showing ceiling and floor effects for perfect performance (value one) and lowest performance (value 0).

8.- ELECTROMAGNETISM

8.1.- The effect of reference electrode position on power consumption in Cochlear Implants

(Published as: Ramos-Miguel, A, Ramos-Macías, A, Artiles, JV, Zaballos, MTP (2015). The Effect of Reference Electrode Position in Cochlear Implants. *J Int Adv Otol*, *11*(3), 222-8.)

Despite the use of different electrode designs and coding strategies across Cochlear Implant (CI) companies, the variability in auditory abilities of CI recipients seems to be more similar across devices than for the same device across individuals. This suggests that significant recipient-dependent factors limit overall auditory ability at an individual level. Aspects such as neural survival are known to have a significant effect. Unfortunately, little can be done to improve patient performance in such cases. However, there might be other reasons related to the physical characteristics of the device, rather than the patient's etiology. (144)

In this study, the effect of Reference Electrode (RE) position on power consumption and stimulation intensity was investigated using two different approaches: 1) 2D numerical simulations and 2) real temporal bone measurements using a test implant developed by the *Laboratory of Psychoacoustics of the Otoneurology Division, ENT Department of the Complejo Hospitalario Universitario Insular Materno Infantil (CHUIMI)*.

8.1.1- Introduction

The RE position in CI has been surprisingly understudied. There are no articles investigating a correlation between battery life and position of the RE. Most centers do not even use this electrode, with just the body of the implant being used. From a physical point of view, RE position has a bigger influence regarding power consumption than intracochlear electrode position.

Despite using different electrode designs and coding strategies across CI companies, the variability in auditory abilities of recipients appears to be more similar across devices than for the same device across individuals. This suggests that at an individual level, significant recipient-dependent factors limit the overall auditory ability. Aspects, such as ossifications, are known to have a significant effect. Unfortunately, to improve patient performance in such cases split electrodes have to be used, and they require exquisite optimization of parameters to obtain good results (144). Goehring et al. discovered that 12.4% of implantations intraoperatively demonstrate high impedances. In 8.2% of cases, impedances postoperatively remain high. (145)

There is extensive literature wherein many variables concerning intracochlear electrodes are evaluated. In contrast, very few studies have explored the effects of position, number, or shape of RE, without which CI would not properly function (146). A field that has been explored in more depth is the comparison between mono- and bipolar stimulation on the still controversial hypothesis that more restricted current fields can provide better speech perception. However, there are problems with bipolar stimulation: 1) Bipolar electrodes require higher currents for supra-threshold stimulation because of a current shunt from electrode to electrode and 2) high potentials and high current densities form around each of the two electrodes and supra-threshold excitation is possible near both the source and sink electrodes (147) (148) (149) (150) (151) (152). Therefore, power consumption is still a major drawback in this stimulation mode, thereby making monopolar the preferred choice of most audiologists. Therefore, it is important to evaluate whether RE position in monopolar stimulation can decrease power consumption and improve electrical field sharpness.

In monopolar stimulation, the current is injected through a contact in the scala tympani and returned to a far-field electrode contact. Briaire and Frijns studied the current distribution in multiple electrode contacts along the scala tympani using this mode (106). The result demonstrated that electrical potential field patterns caused by monopolar stimulation are broad in nature and that potential relatively and slowly drop off as one moves away from the stimulating contact. High impedances in the tissues can cause losses in the electrical impulse quality, thereby requiring high power for the patient to efficiently detect this signal. It can even lead to the deactivation of some electrodes because their power consumption is so high that they hinder the global stimulation strategy. When impedance is too high, particularly at high current levels, then it is possible that the implant voltage will not be sufficient to generate the programmed electric current level. In this case, an electrode is said to be “out of compliance.” This leads to the situation where perceptual loudness does not increase with increases in the current level. To increase loudness, the width of stimulation pulses can be increased. In some cases, impedance is too high that this approach is not feasible because of too large pulse duration required. Hence, high impedance results in a decrease in battery life but also in a wider spread of excitation because of the large current intensity required.

However, decreasing impedance is feasible and can solve the issue of battery consumption. How this might be done depends on the source of impedance. It could be achieved by reducing the space between the RE and signal electrode. The RE can be the extracochlear electrode of some types of implants, which can be placed in different parts of the mastoid, but also the body of the implant, used as ground electrode as well. By reducing the amount of tissue between the ground and intracochlear electrodes, impedance could be lowered, and the pulse intensity required could be obtained with less power. (153) (154)

In CI surgery, not much attention is given to the specific position of the RE. The surgical technique historically suggests positioning the RE as far as possible from the electronic part of the implant to avoid interferences during stimulation. This part is currently isolated from this type of interference by Faraday cages. Therefore, RE can be currently positioned closer to the stimulation regions. (155) (156)

This study aims to examine the variability in extracochlear electrode position as a contributing factor to power consumption in CI. This simple change can be of great advantage regarding the battery life for patients with cochlear ossifications. Because they require high currents to perceive electrical stimulation, a closer distance between the intra- and extracochlear electrodes may enable the generation of larger current intensities without the need to upturn the pulse duration to increase auditory sensation.

8.1.2- Objective, Hypotheses and rationale for the study

The aim of this study is to examine variability in extracochlear electrode position as a contributing factor to variability in auditory ability in CI patients, in the context of two questions, 1) **whether RE placement has a measurable effect on consumption,** and 2) **if RE position has a functional effect on stimulation intensity.**

The hypothesis is that **the closer the RE is placed from the stimulation region, the lesser power consumption will be required to reach a point at a set intensity.**

The reason behind this study has been to try and increase the battery life of the cochlear implant and to improve implant functioning in patients with ossifications in the cochlea. Because they require high currents to perceive electrical stimulation, a closer distance between intra and extracochlear electrodes may allow the generation of more intense currents without the need to increase the pulse bandwidth to augment the auditory sensation.

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8.1.3- Material and methods

Ethics committee approval was not required for this study because it did not involve human subjects.

Classical Circuit Analysis

To characterize the theoretical physical system a much-simplified electrical model was used.

Three electrical formulas were used. The first is Ohm's Law, which was postulated by Geor Simon Ohm in 1826. It states that a current I through a conductor between two points is directly proportional to the potential difference V across the two points, with resistance R being the constant of proportionality between them (157):

$$V = I \cdot R \text{ [8.1]}$$

In this case, *V* is the potential difference produced by CI between the signal and the RE, *R* is the impedance caused by the tissue between the electrodes, and *I* is the electrical current passing through it. Solving for I, it becomes clear that to raise this value, V needs to be increased, or R has to be lowered. Thus, the only way to do this without supplying extra energy is by reducing the resistance between the RE and the intracochlear electrode.

The second equation is the electrical power formula:

$$P = V \cdot I = R \cdot I^2 \text{ [8.2]}$$

It represents the rate of energy that is converted from the electrical energy of the moving charges to some other form, e.g., heat, mechanical energy, or energy stored in electric or magnetic fields. Power is measured in Watts (W) and is used as a measure of the efficiency of a device. It is directly proportional to the square of the electrical current and resistance of a circuit. (157)

Thus, for a fixed *I*, the higher the *R*, the more the power required to be supplied by the circuit to keep I constant. Electrical hearing threshold occurs when the current *I* delivered to the nerve fibers reaches a minimum value. This value can be taken as the fixed value of *I*. Then, it becomes clear that the further away the RE is from the active electrode, i.e., the higher the resistance between them, the more power will be required to reach that threshold.

The last formula used is the mathematical definition of the electrical resistivity of a material, which is the resistance opposed by a material to an electrical current flow across it. This formula gives the characteristic *R* of each biological tissue between the RE and an intra-cochlear electrode.

$$R = \rho \frac{l}{S} \quad [8.3]$$

Where l is the length of the conductor, *S* is its cross-section area, and *ρ* is the electrical resistivity coefficient of the material, which is measured in Ohms per meter (Ω·m). In this study, the values of *ρ* for endolymph (*ρ*=0.75 Ω·m), bone (*ρ*=20 Ω·m), and muscle (*ρ*=3.5 Ω·m) were obtained from the database that was created by IT'IS Foundation. (IT'IS Database Low-Frequency Conductivity)

2-D Numerical Simulations

The first technique used was numerical simulation. The advantage of using this approach is that the physical phenomenon can be recreated in a completely controlled environment where all the variables and geometries can be defined, measured, and modified to match the real-life system.

The simulation software used was Mathematica (Wolfram Research, Inc., Mathematica, Version 10.0, Champaign, IL (2014)). A 2D finite element

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simulation of electromagnetic phenomena for low frequency currents was build.

A useful approach to the calculation of electric potentials (Voltage) is to relate that potential to the charge density. Since electric charge is the source of electric field, the electric field at any point in space can be mathematically related to the charges present. For multiple point charges, a vector sum of each of their electrical fields is required. For a continuous distribution of charge, vector sums are better handled through calculus. This is done through the divergence relationship. This states the divergence $\nabla \cdot$ (the scalar derivative) of the electric field vector *E* is proportional to the charge density ρ (158):

$$\nabla \cdot \mathbf{E} = -\frac{\rho_e}{\varepsilon_0} \text{ [8.4]}$$

Where $\varepsilon_0$is the permittivity and ρ_e is the charge density.

Because the electric field is also the derivative of the potential, we obtain Poisson's equation:

$$\mathbf{E} = \nabla \cdot \nabla \mathrm{V} = \nabla^2 V = -\frac{\rho}{\varepsilon_0} \text{ [8.5]}$$

In a charge-free region of space or when the frequency is low, the governing equation for the electromagnetic field is the static equation, which is Laplace's equation. This is a differential equation with a smooth solution, without singularities. It is used to define the potential energy field caused by a given charge distribution:

$$\nabla^2 V = 0 \text{ [8.6]}$$

The model's task will then be to determine the solution to this equation for a biological model, which involves various materials and geometries. This equation gives the second derivative of V; however, because it is V itself that is needed, integration is required. However, while integrating, constants appear that cannot be numerically determined unless the boundary or initial conditions of the system are known. These are the values at the borders of the system's space or the initial value of V before it starts moving through space and time. (158)

Two types of conditions were used: Dirichlet and null Neumann. The first specifies the conditions that the electrical potential requires to take along the boundary. The second sets the value of its derivative along this boundary. Dirichlet conditions of 0 and 5 V were imposed on the RE and intracochlear electrode, respectively. The null Neumann condition was applied to the edges that limit the geometry of the system. (159)

Another important step in designing the computer model is the geometry, which is illustrated in Figure 8.1. The model represents three electrode positions. They all share the same elements; however, the distance between the electrodes decreases by a factor of 2 between one case and the next.

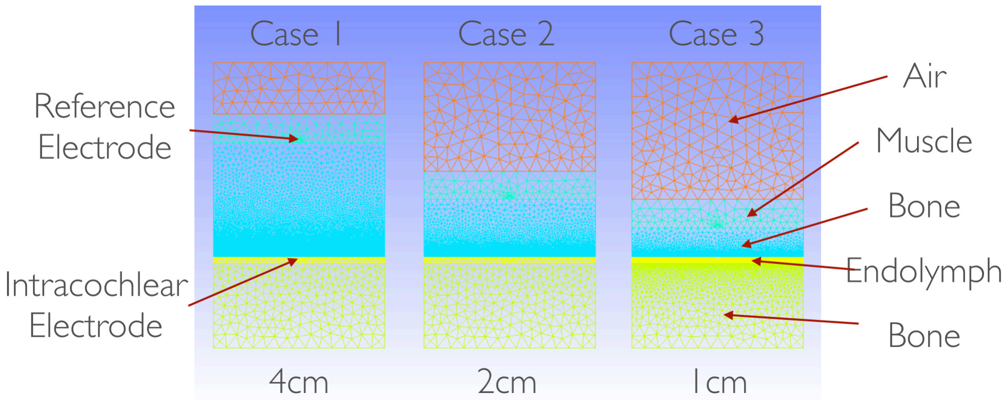

Figure 8.1 Geometries used in the numerical model. The inter-electrode bone layer thickness was modified to study its effect on impedance, and thus, electrical consumption.

Each element was assigned the electrical resistivity values described in the circuit analysis section. Air was included in this model, with ρ=1.3×1012 Ω·m. (160)

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Temporal Bone Measurements

A device was designed to stimulate and monitor electrical impulses that were delivered to a temporal bone while measuring impedance, current, and voltage.

The purpose-built implant comprises one intracochlear and six extracochlear electrodes. Because only the effects of relative position between the RE and intracochlear electrode are of interest in this study, CI was simplified and comprised a single intracochlear electrode. In contrast, to assess extracochlear electrode currents along different regions at the same time, six of them where located along the temporal bone. They all have spherical ends with a mean diameter of 0.9 mm. All extracochlear electrodes can be set to be RE, one at a time, thus providing six different measurement positions. The remaining electrodes act as probes throughout a given stimulation trial.

The system comprises a microcontroller (Arduino UNO, Somerville, Massachusetts, USA) that is connected to a computer via USB. It was programmed to transmit electrical signals, sample each of the probes, and store data. It also calculates impedances, currents, and power that are generated during stimulation. The board is connected to a breadboard with the electrodes and probes to be inserted into the temporal bone.

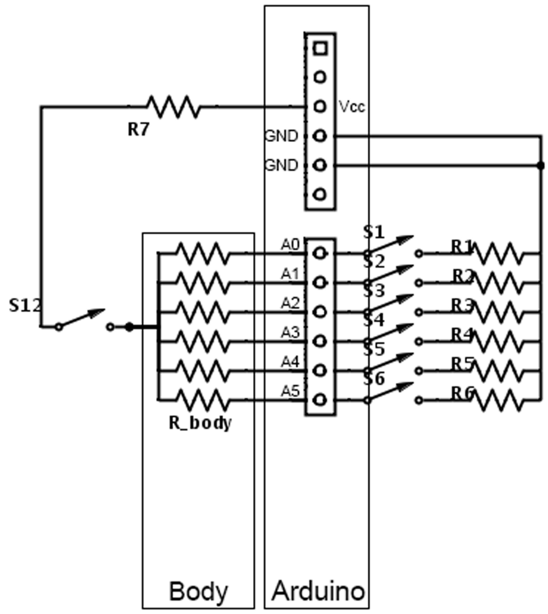

Figure 8.2 Schematics of the implant circuit. The reference/probe electrodes were attached to the analog ports of the microcontroller board. Switchers S1–S6 define which probe to select as RE. S12 disconnects or reconnects the intracochlear electrode. Resistors R1–R7 all have a value of 1.5 kΩ.

The circuit is based on a number of voltage dividers. The schematics are depicted in Figure 8.2. The two resistors in series form a voltage divider circuit. One end of the resistor pair is fed 5 V, while the other end is connected to the ground. Five volts that the Arduino provides are divided between the two resistors depending on their resistance. The resistor, which holds the greater resistance, gets more of the voltage, according to Ohm's law formula. The voltage that falls across a component is directly proportional to the amount of resistance it contains (157):

$$R_{body} = \frac{R_{reference} - V_{in}}{V_{out}} - R_{reference} \text{ [8.7]}$$

Using this principle, a model can be set up to determine the resistance on the basis of the voltage division. With this data, all electrical characteristics of the circuit can be defined, i.e., impedances, currents, and voltages.

The software to control the implant was purposely designed by the research group. The electrode implantation was done on a temporal bone with the cochlea exposed to have access to a wider range of measurement points. The intracochlear electrode was inserted at 6 mm from the cochleostomy and then fixed. To facilitate electrical conduction, a steady uniform flow of saline solution was applied over the entire temporal bone throughout the test.

Measurements were taken at three different points across the temporal bone: the mastoid region, standard position, and root of the zygomatic arch of the temporal bone. The specific locations are indicated with black squares in Figure 8.3. The implant processor returned and average for each point based on 2000 measurements, to minimize the effect of noise on the signals.

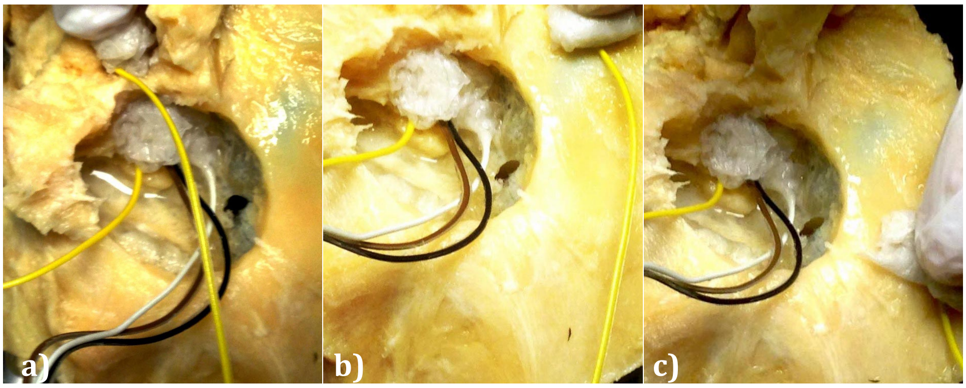

Figure 8.3. a-c. Real images showing the three RE positions studied: a) at the root of the zygomatic arch of the temporal bone; b) in the standard temporal bone position; and c) far away from the cochlear in the mastoid region.

The statistical analysis was performed using IBM SPSS Statistics for Macintosh, V. 21.0. (IBM Corporation; New York, USA).

8.1.4- Results

2D Numerical Model

The finite element method was applied to each of the generated geometries to solve the electrostatic problem, which is defined by the potential between the RE and intracochlear electrode. After performing the simulation, the electric field lines were evaluated. Figure 8.4 illustrates the results. It can be observed how as the electrodes get closer together, the field lines are closer to the intracochlear electrode, which means that the current generated becomes higher.

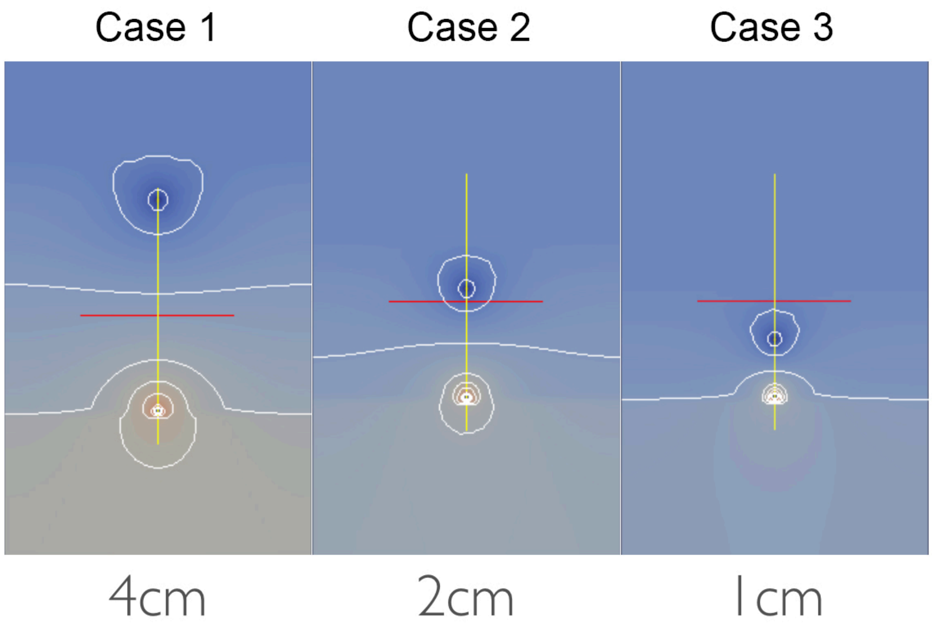

Figure 8.4. Electric field lines as a function of bone thickness. As the RE is placed closer to the intracochlear electrode, the field lines show higher intensity in the space between them, where the auditory fibers are located. The cross represents the origin of the coordinate system (0,0), and the white lines represent electric field contours of equal value.

The next measurement taken was the behavior of the current at the Y-axis, which is defined as the distance perpendicular to the longitudinal axis of the electrode circumference. The zero point was established in the intracochlear electrode, and the maximum point was 6-mm away from it. Figure 8.5 shows how in the case of the shortest distance (bone thickness, 1 cm), the value of the current is always higher than in the case of maximum distance measured (bone thickness, 4 cm). The maximum difference is observed in the region near the electrode, which is the target zone for an effective stimulation of the auditory nerve.

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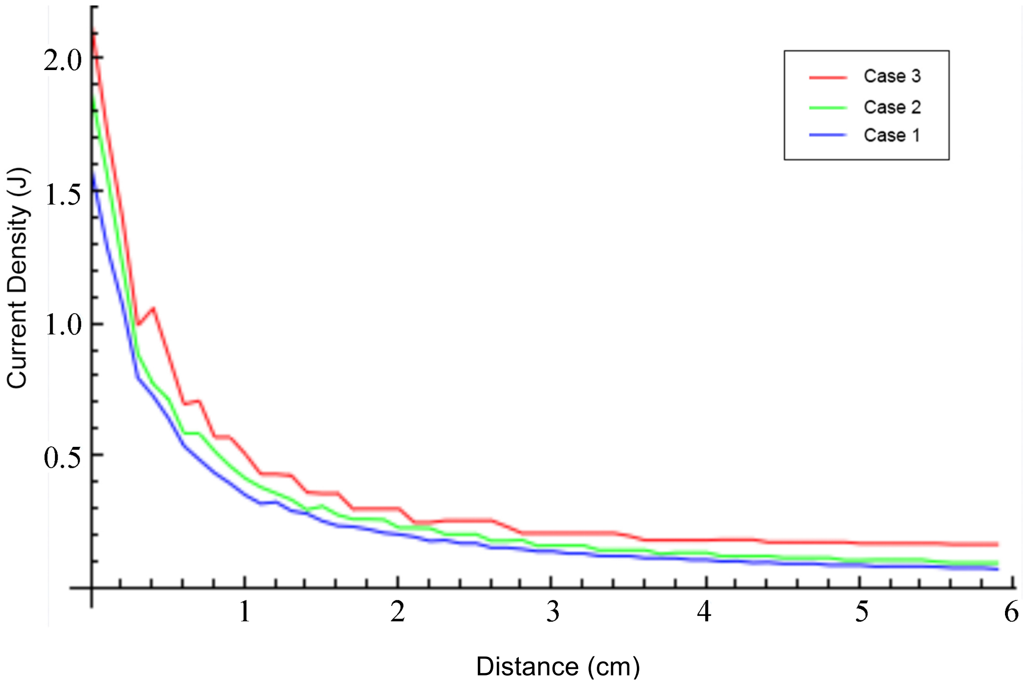

Figure 8.5. Current as a function of inter-electrode distance at the Y-axis (perpendicular to the electrode longitudinal circular axis). Each line represents a bone thickness (Case 1 being the one with the highest value and Case 3 with the lowest). All of them follow exponentially decaying profiles, although the values are higher for the case where bone thickness is lowest (Case 3).

Finally, the evolution of the flow at a distance of 2 mm from the electrode at the X-axis was studied. This was defined as the longitudinal axis of the electrode circumference. As observed in Figure 8.6, moving along the X-axis also shows how the current generated is higher in the cases where the inter-electrode distance is shortest.

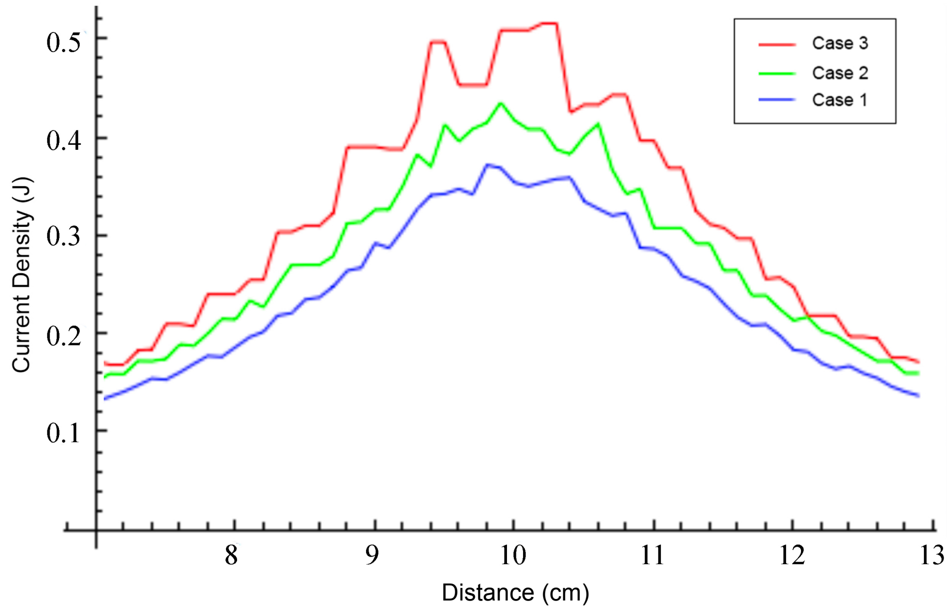

Figure 8.6. Current as a function of inter-electrode distance at the X-axis (along the electrode longitudinal circular axis). Each line represents the same bone thickness as in the previous figure. The three cases follow a similar profile, although the lowest thickness one is the one showing the highest current values.

However, the current density *j* generated for a fixed potential increases as the electrodes approach. Table 8.1 shows the normalized current density for each case. It can be observed how in the case where the electrodes are closest together, the current density is 33% higher than in the case where they are the furthest apart.

Case	Current Density
A (furthest away from the cochlea)	100%
B (intermediate distance)	115.16%
C (closest from the cochlea)	133.01%

Table 8.1 Current density as a function of bone thickness. The variables show an inverse relationship that is consistent with current findings from the previous method used in this study

Temporal Bone Measurements

Impedances, currents, and potentials that were produced on the temporal bone in the three positions described in the previous section were measured (Table 8.2). It can be observed that as the RE approaches the intracochlear one, the generated current increases. The power when the electrodes are closest (C) is 38.78% greater than in cases when they are farthest apart (A). This causes a current that is 176.12% stronger with a lower potential difference.

Case	Impedance (kΩ)	Voltage (V)	Current (mA)	Relative Power
A (furthest away)	45.88	4.38	0.0955	61.22%
B	30.80	4.20	0.1364	83.87%
C (closest)	24.14	4.05	0.1682	100.00%

Table 8.2 Resistance, voltage, current, and power consumption of each RE position of the temporal bone measurements.

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III. DISCUSSION

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9.- PSYCHOPHYSICS

9.1.- How the healthy ear understands speech in noise and the effects of training on it

The ability to focus and understand a speaker in a noisy environment is a critical social-cognitive capacity whose underlying mechanisms are still unclear. Many explanations have been proposed. However, none had looked at high frequencies as the possible cues for speech identification in noise. Thus, in the first psychophysical study of this thesis, the relation between speech recognition in noise and frequency bandwidth of the auditory stimulus was investigated.

Many studies have already demonstrated the importance of frequency bandwidth in the perception of speech in quiet. For example, Stelmachowicz et al. have studied these effects on the perception of the phoneme /s/ in Normal Hearing (NH) children and adults. The results showed a correlation between these two factors for all groups. A clear example where a limited hearing bandwidth can be found is in presbycusis, which involves the rise of auditory thresholds for high frequencies (161). In these patients, the most common complain is the inability to understand Speech In Noise (SIN), while they can still follow a conversation in a quiet environment with little difficulty. These two findings (that high frequencies are important for speech perception in quiet, and that the rise in high frequency hearing thresholds hinder SIN ability) seem to point in the direction of high frequencies as important cues for speech perception, specifically in the case of background noise.

A study by Stuart et al. analysed word recognition performance in 12 NH adults in continuous and interrupted broadband noise as a function of SNR with and without filtering at 2 kHz. However, their filtering threshold was lower than the actual 8 kHz that average implants have. Moreover, they focused primarily on the difference between continuous and interrupted noise, rather than the role of high frequencies. In a period where Cochlear Implants (CI) only reached around 4 kHz, Hornsby also demonstrated that better

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performance was accomplished when the frequency bandwidth was increased to 7 kHz. (162) (163)

The statistical outcome in this thesis is in accordance with the findings from Stuart and Hornsby, for NH subjects, when noise is either higher or lower than the speech signal. The results suggest that high frequency components above 8kHz may play a significant role in word recognition in noisy environments.

However, for the situation where noise and speech have equal levels of intensity this was not the case. The characteristics of the experiment do not allow for a clear explanation as to why this happens. According to the results obtained, there seems to be a difference in how humans recognize speech in different noise situations. This is in accordance with a neurophysiological study by Wong et al. where they measured cortical activity of NH subjects while they identified words in two noise conditions: below and above the speech level. They found cortical noise-dependent activation of the bilateral middle and left posterior portions of the superior temporal gyrus. These results likely reflect demands in acoustic analysis, auditory-motor integration and phonological memory, as well as auditory attention. Unfortunately, they did not study what happened when noise and speech were at the same level, so the activation sites in this case are still a unknown. Thus future research points in the direction of combined psychoacoustics and neurophysiological experiments, where neural activity is recorded while speech material is presented using the three SNR of the present experiment. (69)

Correlation studies between how much a word was altered by filtering and its score did not yield conclusive results. Significant correlation was only found for the case where noise was lower than speech. This was somewhat expected, since word recognition is a complex phenomenon. Vowels and consonants are perceived through different mechanisms. Nie et al. sustain that consonant recognition happens through temporal cues while vowel recognition relies on spectral cues. The results of the present experiment, in combination with the statistical outcome observed, seem to indicate that even small additions of high frequency content to sounds can lead to an overall improvement in speech recognition in noise. In the light of these arguments, it is hypothesized that vowel recognition assumes an important role in speech recognition in noise. Moreover, results from Friesen et al. indicate that proper vowel recognition requires appropriate frequency allocation. The implications of their findings and those obtained in this thesis lead to the thought that for appropriate speech recognition in noise, it will be necessary to fulfil two requirements: 1) a correct frequency allocation and 2) the preservation of high frequencies. This is because if the audible frequency is increased, but the allocation of such frequencies is not accurate, then the result will not be reconstructed properly by the brain. (164) (165)

The first study (section 6.1) thus showed how degraded high frequency thresholds can lead to decreased ability to understand SIN. Hereafter, the natural question to ask is: is there a way to prevent, or even reverse this problem? If the answer to this question were positive, this would mean that patients with hearing loss could significantly improve their SIN perception through an appropriate training protocol.

The search for the answer leaded to the second study of this thesis (section 6.2). The ability of air traffic controllers (ATC) to understand SIN under limited frequency bandwidth conditions is extraordinary. However, this does not seem to be an innate ability. These individuals all report having progressively acquired this faculty throughout the years, but specially during their two-year training period. Thus, the second experiment of the series of psychophysical experiments in this thesis tried to objectify the effect of training on SIN understanding.

Many studies have investigated the effect of SIN exercises. Song et al. studied training-related malleability using a program that incorporated cognitively based listening exercises to try to improve SIN perception in patients with hearing loss. Trained subjects exhibited significant improvements in SIN perception that were retained 6 months later.

Subcortical responses in noise demonstrated training-related enhancements in the encoding of pitch-related cues. This was the first time that short-term training was demonstrated to have the potential to improve the neural mechanisms for SIN perception. These results involve and define biological mechanisms that contribute to learning success, and they provide a conceptual advance to the understanding of the kind of training that can influence sensory processing in adulthood. (99)

More recently, Sweethow and Sabes prospectively assessed the generalization of SIN training in a cohort of individuals with hearing loss (166). Whitton et al. prospectively assessed the effect of signal-in-noise, audio-game training on SIN understanding with untrained materials (167). On the other hand, Fu and Galvin and Moore and Shannon have shown that targeted auditory training can further enhance the benefits of new implant devices and/or speech processing strategies (168) (169). These findings and those from this thesis suggest that SIN training could be beneficial both for CI users and for those with moderate hearing loss, since it equips patients with the cognitive tools to confront complex auditory situations.

However, the only groups that have been studied and that have some kind of long-term training are musicians, who are auditory experts. For example Parbery-Clark et al. investigated the effect of musical training on SIN performance. They found that musical experience improves the ability to understand speech in challenging listening environments. The results also suggest that this enhancement is derived in part from musicians' remarkable working memory and frequency discrimination. (170)

The results described in section 6.2 indicate that ATC, just like musicians, are more capable of identifying SIN. This ability is most apparent in the most adverse case, when the signal is lower than the noise. In all cases, they obtained better scores than the Control group. The proposed explanation is that this is due to the effect of training that occurs naturally during their daily activities. Even though the set of words used were completely unknown to them, it seems that the skills learnt in one particular auditory situation are transferable to a new and different one, in this case, the laboratory test. This is an interesting result, since the first thing that subjects commented after being informed of the purpose of the study was that they 'would probably not perform better because their "trick" is that they know which words to expect during radio communications'. However, it is clear from the results that this is not the case and they could identify more words than the standard NH population. Regretfully, the prevalence of this opinion was not quantified in this study. It would have been very interesting to see how much their beliefs correlate with the test results. The studies reviewed here favour the hypothesis that their subjects, and likely ATC as well, are better at focusing on the less degraded acoustic speech cues and then filling the gaps using cognitive skills in the presence of background noise. This way they require less time of adaptation and therefore miss less information.

Given the promising results, it was decided to investigate possible correlations between years working and age. However, this analysis did not return clear results. The two cases where significant correlations were found (inverse correlations between 8 kHz 5 SNR test results and 22 kHz 5 SNR) do not seem meaningful in the context of the other tests. If a correlation of any kind is to be established between these two variables, then consistency across tests is required. Thus, the results observed are likely to be caused by the limited number of subjects, which is an unfortunate but common aspect in clinical studies, as opposed to pure physics studies, due to the difficulty of recruiting volunteers. In such cases, statistics needs to be treated with care. It is possible and very likely, given the time taken by subjects in the studies mentioned before (approximately 6 months- 1 year) (97), that the learning curve is steeper during their first year of training and approaches a plateau thereafter. Thus, another study emerges as a very interesting continuation of the present one: to test ATC in training at different stages of their learning period, or to follow them longitudinally. Maybe looking more closely into this period of their working life will reveal a significant correlation between SIN identification and training duration.

The study also looked at how high frequencies affected SIN understanding in this population group. Interestingly, both control and target groups had statistically significantly better results when the unfiltered lists were used. The results indicate that performance decreases when high frequency components of speech are removed. Still, the ATC group performed better in both conditions, thus indicating an auditory learning effect, but the present study could not acceptably correlate it with either age of working years. Nevertheless, the role of high frequencies is remarkable, since it is present in both trained and untrained subjects. This further supports that frequencies above 8 kHz can help humans to better understand SIN. To draw further conclusions, a more specific experiment needs to be designed, where groups of words with similar high frequency contents are selected to try and establish a relationship between word identification and its frequency content, because the speech material used has balanced phoneme content.

The implications of these findings are profound because they can result in better rehabilitation therapies for people with hearing loss. Our aging society is bound to suffer SIN understanding impairments. As individuals grow older, this ability degrades even before auditory losses become clinically relevant. Therefore, it is capital to come up with strategies to try and slow this process down as much as possible. The results from section 6.2 have demonstrated that daily exposure to speech in noise can result in adaptation and enhanced performance in NH individuals. This is the first time that ATC are studied for this singular ability. In addition to the recent proofs that learning during adulthood is far greater than previously thought, we provide evidence that auditory training under degraded sound quality can lead to better SIN understanding and can have big implications in the way hearing loss is treated today. Hence, research in the line of this study will certainly aid in the development of efficient and effective training protocols and materials.

10.- ELECTRICALLY EVOKED PSYCHOACOUSTICS IN COCHLEAR IMPLANT RECIPIENTS

10.1.- How physical and psychophysical variables affect electrode discrimination in cochlear implant recipients

The aim of this study has been to measure the effect that the distance between the intrachoclear electrodes and the modiolus has on the ability of a patient to discriminate the origin of a certain stimulus, separating this effect from those of the NRT, T level and impedance, since these factors may also influence the ability to discriminate electrical pitch. The motivation for this study has been that a better ability to discriminate electrical pitch could help to design better training for CI patients, since it has been seen in section 9 how frequency resolution plays a major role.

The results provide evidence that the ability to discriminate between electrodes depends on multiple factors. In this study, NRT, T level, impedance and distance were identified. They were all statistically significant in explaining electrode discrimination.

Especifically, the statistical analysis provides empirical evidence in favour of the hypothesis that the electrode distance to the inner wall of the cochlea is a significant predicting variable on the electrode discrimination ability of the patient, measured as success rate. The results indicate that the greater the distance, the lower the success rate will be. Thus, distance to the inner wall will result in more difficulties to perceive electrode differences.

The explanation can be found in the extensive research that has been conducted to describe current field patterns for various electrode configurations within the cochlea (171) (172) (173) (105). These experiments show that the spatial gradient of the electric field increases as the distance between the electrodes and the neural ends decreases. Thus, an electrode that is closer to a surviving neural element would require less current to reach discharge threshold due to the steeper slope of the electric field. The result would be lower thresholds for proximally placed electrode arrays. Moreover, steeper electric fields also produce less current spread to adjacent neural ends, which would also reduce channel interaction. Therefore, in our experiment, a patient with closer electrode-inner wall distance has a narrower excitation region per electrode, and hence perceives two adjacent stimulations as being more dissimilar than a patient with a larger distance, under the same conditions of neural survival and impedance.

Apart from the computer model results, human experiments have also found that perimodiolar placement appears to improve electrode discrimination (174). Cohen et al. measured electrode discrimination ability in three subjects implanted with a perimodiolar electrode array. Two of the three subjects had a portion of the electrode array close to the modiolus. Those subjects had smaller pitch difference limens for electrodes located in this portion than those that occupied a more lateral position. These results also suggest better spatial selectivity for electrodes positioned closer to the neural ends. Hughes and Abbas also found that subjects with a perimodiolar array had both narrower ECAP-derived width and better electrode discrimination. (175)

Perimodiolar arrays therefore offer the potential for improvements in speech processing due, in particular, to narrower regions of neural excitation. It remains to be seen whether such changes would result in practical improvements to speech perception. Although such narrowing would be less for stimulation towards the upper end of the dynamic range, it has been found that the effect is significant for the Contour array relative to the straight, even at 80% of the loudness. (176)

In the present study, electrode discrimination also depended on T level. Previous findings indicate that electrode discrimination improve with increasing level (177) (109) (178). Other studies found that distance to the inner wall was positively correlated to lower thresholds and comfortable loudness levels (179) (180). There is, therefore, an unsolved contradiction between lower intensity and worse performance. If the theory developed by the models described above were true, then one would expect that, at low intensity levels, the patients would be most capable of discriminating electrodes. However, experience does not support this theory. Based on neurophysiological studies, the author proposes a theory to unify both concepts and explain why lower distances to the neural ends are correlated with increased pitch discrimination ability, but so is higher T level, as opposed to what might be expected, because a lower distance to a neural end causes a lower T level.

Raggio and Schreiner examined single and multi-unit responses to cochlear electrical stimulation in the cat auditory cortical region (181). They found parallel bands of low-threshold response separated by a band of high-threshold response. Each low-threshold band exhibited a cochleotopic map of the cochlear place of stimulation. Bierer and Middlebrooks showed that at supra-threshold stimulus levels, the focus of maximal cortical activity, or centroids, also followed a tonotopic organization (182). Spelman et al. predicted that a broad bipolar stimulus would produce two discrete foci of activity in the cochlea, which would presumably lead to discrete foci of cortical activation (183). This has been observed in many studies, but with a limitation imposed by the difference in electrode-neural end proximity across cochlear regions (182) (184). Apical electrodes are located closer to these neural ends due to the smaller diameter of the scala timpani in this region. Consequently, a stimulus coming from such an electrode can dominate over one from a less proximally located electrode, because the intensity received from the former is higher than for the later. What has been observed consistently is that there is a limit at which the auditory cortex images no longer show the two stimuli as two foci of activity (182) (185). Therefore, a plausible explanation for the controversy between high intensity but low electrode-neural end distance required for optimal pitch discrimination is that pitch is encoded as the centroid of a broad stimulation region, but within a certain spatial extension. In other words, the auditory cortex takes the broadening in the centroids, which determine the stimulus pitch, as intensity increases until a certain broadening value is obtained. When the stimulation region is too large, the sound starts to be perceived as noise and so two nearby electrodes cannot be recognized.

Impedance was also found to have the biggest correlation with electrode discrimination. The cochlear implant electrodes are the boundary between the electrical stimulus and the auditory nerve fibres that need to be stimulated. Therefore, to ensure an optimal stimulation, electrical impedance must be minimized. This parameter depends on electrode surface area, morphological processes and electrochemical processes initiated by electrical stimulation, and is a major factor in power consumption. Moreover, electrode impedance provides an indication of the status of the electrode-tissue interface.

Intracochlear fibrosis following cochlear implantation is very common (186). As the area becomes inflamed and healing after surgery occurs, scars develop in the cochlea, forming fibrotic tissue that progressively covers the electrode array (187). Such inflammatory processes compromise the survival of neural ends (188). Hence, after surgery, changes in electrode impedance may be expected even before commencement of electrical stimulation, because morphological changes start to occur right after the surgery at the electrode-tissue interface. The formations of scars can reduce the performance of cochlear implants by increasing impedance, which raises the threshold stimulation required for sound perception and hence the implant energy consumption. (189)

The implications of the results presented in this thesis are that perimodiolar electrodes could be a better solution in terms of pitch discrimination not only because they are closer to the modiolus, but also because their impedance is lower. Moreover, as re-implantation will be necessary for the infant population receiving cochlear implants nowadays, a perimodiolar electrode in the first intervention would lead to a less fibrotic cochlea in the second surgery and would lead to better pitch discrimination due to lower impedance than if implanted with a straight electrode array.

However, the study presents several limitations that must be discussed to help push forward any further research in this line. First, it would be highly recommendable to extend the sample size, given that the sample available does not allow the elimination of the patient effect. It is necessary to discard the possibility that correlations observed and quantified in table 7.3 and in the models estimated in section 7.2.4 could be due to a patient characteristic that was not controlled in the experiment. Therefore, a larger sample could potentially smooth out such individual differences. In this line, the enlargement of the sample must be carried out using a sample selection that fixes the patient demographics (years of deployment, age at which was implanted, medical team that conducted the implant, etc), in order to rule out such individual differences, creating a more carefully controlled system. This is a very hard requirement to fulfil in the field of medicine, specially regarding cochlear implants, due to the low proportion of the population that carries one, the very large number of possible pathologies that lead to implantation, the years of auditory deprivation, etc. Hence, a multicentre study seems like the favourable option to consider.

Another limitation is the imaging technique. The comparison of psychophysical outcomes for subjects using the straight and Contour arrays was made in the light of data from Computer Tomography (CT) scan analyses of the positions of the electrode arrays. The Contour electrode arrays were overall found to be positioned considerably closer to the modiolus than the straight electrode arrays. CT scan seems to have a clear advantage over the previously used plain film radiography with the standard Stenvers projection. With modern multislice CT scanning, more detailed information can be gathered on the intracochlear electrode position. However, when determining the exact position of the electrode in the cochlea, it was difficult to identify the separate electrodes in relation to the modiolus. This was mainly due to image degradation caused by partial volume effects, but metallic artifacts of the electrode array also posed a problem (190). The window depth in our software configuration limited us in visualizing the two extreme contrasts we wanted to investigate: the fluid compartment and the bony structures of the

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cochlea and modiolus; and the radiopaque electrode array. However, our software overcame this limitation by adjusting inter electrode distance to the total length of the array, which was clearly visible. In the future, technical improvements in both software and hardware will further improve the spatial resolution of this technique under such extreme contrast conditions. In fact, the CT cone beam technology that exists in some centers is currently being used to obtain higher image qualities that even allow independent electrode position. (191)

Finally, a third limitation can be due to differences in electrode design. Electrodes in the straight arrays consist of full platinum bands that encompass the entire circumference of the array. In contrast, the Contour array consists of half-band electrodes located on the medial portion of the precurved carrier. Thus, orientation and spread of current fields will be different for the two electrode designs. This could not be quantized in this study, but, given the large importance that the electric field patterns have in determining auditory nerve stimulation, further studies where this variable is analyzed would be needed in order to further increase the model's predictive power.

In the light of these findings, perhaps the path to substantial improvements lies in employing new speech processing algorithms that incorporate detailed knowledge of the neural response to stimulation of individual patients' arrays. A narrower excitation pattern produced by stimulation suing a perimodiolar array would provide a finer palette for the construction of desired excitation profiles, more closely related to those of normal hearing. However, a larger sample size is required as well as correlation studies between speech perception and electrode discrimination before giving conclusive statements.

11.- ELECTROMAGNETISM

11.1.-The role of the reference electrode in cochlear implants power consumption

In the previous studies of this thesis, different factors that affect the ability of patients with hearing loss to interact with society have been evaluated: auditory training, positioning of the electrode array and broadening of the frequency range. All of these result in a patient's better quality of life. However, little attention has been paid to the physical characteristics of the reference electrode and its effect on said quality of life. Especifically, the power consumption caused by its anatomical position can have a significant impact on the patient's autonomy and the economic costs of the maintenance of the implant.

The RE position, in fact, defines many aspects of the electrical behavior of the CI once implanted. As shown in the section 8.1.4, current intensity, impedance, and power that are necessary to stimulate the auditory nerve are all dependent on it. In this study, the effect of distance between the RE and intracochlear electrode on these parameters has been evaluated to determine the consumption difference caused by it. To this end, two independent and complementary methods were used. First, a numerical model was built using finite element analysis. Then, measurements were taken from a temporal bone.

The resulting data were used to validate the consistency of the two models because they all demonstrated the same physical behavior. These results were particularly good because one can then use them as complementary techniques to provide a wide range of direct and indirect measurements.

Studies on other areas have demonstrated the influence that a correctly positioned RE can have on the efficiency of electrical stimulation. Butson and McIntyre demonstrated that the inclusion of either electrode or tissue capacitance reduces the volume of tissue activated in a manner dependent on the capacitance magnitude and the stimulation parameters (amplitude and pulse width) in deep brain stimulation. (192)

On the basis of the available measurements and modeling results, the scala tympani is usually considered to be a preferential current pathway that acts like a leaky transmission line. Therefore, most studies assumed that current thresholds exponentially decay along the length of the scala tympani. Briare suggests that although the relatively well-conducting scala tympani turns out to be the main current pathway, the exponential decay of current is only a good description of the far-field behavior (106). In the vicinity of the electrodes, higher current densities are found, which are best described by a spherical spread of the current.

No studies have been published where the position, shape, and impedance of RE are investigated in detail. Yet there is evidence that a change in the impedance of the RE interface causes substantial current variations. The essential results that have been observed from the work of Black et al. and Spelman et al. and the present study is that the position of the RE lines affects impedance and currents generated and therefore, the power consumption of the device when stimulating the auditory nerve. (146) (193) (184)

Bone thickness was the adjustable variable of choice in both the theoretical and numerical models in this study because its constant resistivity (ρ) is larger than the other tissues'; therefore, it will have the strongest effect on conductivity. A real-life example is observed in patients with ossifications. In these cases, both current threshold and electrical impedance are higher than average. This occurs because the resistivity (ρ) increases. As shown in Eq. 8.3, for constant size and length, but increased resistivity coefficient, the resistance of the material increases. The increase in ρ is because of the increased bone density that is caused by ossification.

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Therefore, the experiment described in section 8 demonstrates that placing the RE closer to the edge of the surgical mastoidectomy may result in a decrease in IC power consumption, particularly in the case where the patient presents ossified regions of the cochlea. In such cases, a small decrease in distance between RE and intracochlear electrodes can mean the difference in deciding whether to keep a certain electrode activated or to disconnect it because of its high power consumption. (194)

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IV. CONCLUSIONS

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1. High frequency filtering of speech and background noise above 8 kHz causes a significant decrease in word recognition at SNR = ±5 dB SPL in normal hearing subjects.

2. Through involuntary auditory training derived from their professional activity, Air Traffic Controllers understand speech in noise significantly better than an ordinary normal hearing individual, as evidenced by the SIN laboratory test.

3. The differences in performance between filtered and non-filtered word recognition are smaller for Air Traffic Controllers than for the ordinary hearing individual.

4. Patients who have electrodes placed in a perimodiolar position have better electrode discrimination for low (25% of the DR) intensity stimuli than those who have lateral wall type electrodes.

5. Patients who have electrodes placed in a perimodiolar position have lower impedances.

6. Reference electrode placement has a measurable effect on consumption in *in-vitro* measurements using a test implant.

7. The reference electrode position has a functional effect on stimulation intensity in *in-vitro* measurements using a test implant.

8. The integration of a physicist in a hearing loss unit can result in innovative and multidisciplinary discoveries in the field of otology, as demonstrated by the variety of studies completed during this PhD, which will ultimately result in greater patient well-being.

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V. RESUMEN EN ESPAÑOL

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JUSTIFICACIÓN

En el mundo actual, donde la interdisciplinariedad juega un papel clave en la investigación, es vital cambiar los esquemas tradicionales que separan a los profesionales de acuerdo a su grado universitario. Para conseguir una verdadera revolución científica, los diferentes campos del conocimiento deben mezclarse, de modo que los problemas tradicionales se miren desde nuevas perspectivas. Así es como surgen soluciones innovadoras.

La tesis intenta ilustrar este concepto. La adición de un físico a una unidad de hipoacusia ha dado diferentes posibles soluciones a problemas clínicos. La presencia de un físico en una sesión clínica, que escucha las necesidades reales del personal sanitario, puede acelerar drásticamente la innovación en lo que respecta a la cirugía, programación del implante coclear, rehabilitación y evaluación del paciente. En esta tesis, se ha trabajado tratando de cubrir todos estos aspectos.

El hilo conductor de los estudios aquí presentados es la mejora de la percepción del sonido de los pacientes implantados cocleares-bajo diferentes condiciones. El trabajo se ha dividido en los diferentes campos físicos que se aplicaron para resolver cada pregunta inicial. Se evaluaron tres condiciones de sonido. En primer lugar, se estudiaron los mecanismos de percepción del habla en ruido. Después se investigó como la discriminación de tono eléctricamente evocado se ve afectada por la proximidad del electrodo a las terminaciones neuronales. Por último, se estudió la forma de mejorar la transmisión del sonido eléctricamente evocado en pacientes con altas impedancias.

La primera parte se centra en la psicoacústica, un campo híbrido entre la física y la psicología. Se realizaron dos estudios. El primero es un estudio sobre la importancia de las altas frecuencias para la percepción del habla en ruido en las personas con audición normal (como primera medida para poder extrapolarlo a la población implantada en un futuro). Esto demostró que los componentes de alta frecuencia del habla, por encima de 8 kHz, ayudan a entender las palabras en presencia de ruido ambiental. El segundo estudio es una continuación del primero y explora el efecto del entrenamiento a largo plazo en la percepción de ruido en los controladores de tráfico aéreo. Este sector de trabajo desarrolla su capacidad de percepción del habla durante el curso de su trabajo. Las condiciones acústicas especiales a las que están sometidos les hace entrenar sus habilidades cognitivas para que su cerebro consiga extraer la máxima cantidad de señales auditivas y poder establecer comunicaciones de radio. Dichas comunicaciones tienen el mismo ancho de banda que las prótesis auditivas (audífonos e implantes). Por lo tanto, el estudio demuestra cómo el entrenamiento mejora la capacidad de entender el habla en ruido y propone la inclusión de la palabra en la formación de ruido para las personas con pérdida auditiva.

La segunda parte se centra en la psicofísica de la audición eléctrica. Se centra en evaluar el efecto de la distancia de los electrodos al nervio auditivo como un factor importante para la discriminación de tono en pacientes implantados cocleares. Este estudio muestra que una posición más próxima al nervio auditivo produce una mejor discriminación de los electrodos.

La tercera parte utiliza los principios físicos del electromagnetismo para mejorar la eficiencia energética del implante coclear. Se sugiere que el reajuste de la posición del electrodo de referencia en el momento de la cirugía, -Colocación lo más cercana posible a la cóclea-, puede disminuir la corriente eléctrica necesaria para producir una respuesta del nervio auditivo. Esto a su vez traería beneficios económicos para el paciente, ya que se traduce en una mayor duración de las pilas.

Los resultados de estos estudios demuestran los beneficios de incluir un físico en la rutina diaria de un departamento médico. Esto puede generar soluciones personalizadas y específicas. Por lo tanto, el papel del físico debería tenerse en cuenta en los nuevos modelos de organización de los servicios de salud como motor de la innovación y los tratamientos individualizados.

OBJETIVOS

- Efecto de la supresión de frecuencias por encima de 8 khz sobre la comprensión del habla en ruido: con el fin de entender por qué los sujetos con pérdida auditiva tienen dificultades en la comprensión del habla en ruido, primero es necesario entender los procesos que permiten a los sujetos con audición normal dirigir la atención a la fuente sonora deseada. Sin embargo, estos mecanismos neuronales no se conocen con certeza. Por lo tanto, este estudio tiene como objetivo evaluar la forma en la que las frecuencias agudas percepción contribuye a la percepción del habla en ruido.

- **Efecto del entrenamiento auditivo en la percepción del habla es ruido en controladores de tráfico aéreo:** a diferencia de los trabajos anteriores, donde los pacientes reciben formación in situ y después se realiza el estudio, nuestro abordaje se centra en cómo el entrenamiento auditivo intrínseco del trabajo de los controladores puede dar lugar a un mejor rendimiento en las pruebas de laboratorio. Por otra parte, también se evaluó el papel de las frecuencias agudas por encima de 8 kHz, ya que las comunicaciones de radio son limitadas de igual modo que las prótesis auditivas a esta cifra. Por lo tanto, este estudio tiene dos objetivos principales. El primero consiste en evaluar si los controladores de tráfico aéreo tienen mayor capacidad de entender el habla en ruido en una prueba de laboratorio. Con este fin, se comparó este grupo con uno de control compuesto por sujetos con audición normal sin entrenamiento. El segundo objetivo es comparar su rendimiento al filtrar las frecuencia por encima de 8 kHz.

- **Comparación de la discriminación de percepción del tono eléctricamente evocado en pacientes portadores de guías de electrodos y recta perimodiolar:** el objetivo de este estudio es aportar evidencia de que los electrodos perimodiolares dan una mejor discriminación de electrodos a baja intensidad eléctrica, ya que están más cerca del modiolo. con este fin, se estudiaron las capacidades de discriminación de electrodo en usuarios de guías rectas (422) y perimodiolares (coclear CA (RE)) utilizando una plataforma desarrollada específicamente para esta tarea.

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- **Efecto de la posición del electrodo de referencia sobre el consumo de energía en los implantes cocleares:** El objetivo de este estudio es examinar la variabilidad en la posición del electrodo de referencia como un factor que contribuye a la variabilidad en la capacidad auditiva en pacientes portadores de implante coclear, en el contexto de dos preguntas, 1) si su colocación tiene un efecto medible sobre el consumo, y 2) si la posición RE tiene un efecto funcional en la intensidad de la estimulación.

CONCLUSIONES

1. EL filtrado de altas frecuencias, por encima de 8 kHz, de la palabra y el ruido produce una disminución significativa en el reconocimiento de bisilábicas a ratios de señal-ruido de ± 5 dB SPL en sujetos con audición normal.

2. Los controladores de tráfico aéreo, a través de un entrenamiento auditivo involuntario derivado de su actividad laboral, son capaces de comprender el habla en ruido significativamente mejor que una persona con audición normal sin entrenamiento auditivo.

3. Al igual que en los sujetos con audición normal sin entrenamiento, los controladores de tráfico aéreo experimentan diferencias en el rendimiento entre el reconocimiento de palabras cuando estas y el ruido son filtradas a partir de 8 kHz y no filtradas aunque las diferencias son más pequeñas en los controladores que en los sujetos sin entrenamiento.

4. Los pacientes que tienen electrodos colocados en una posición perimodiolar tienen una mejor discriminación de electrodos para estímulos de baja intensidad (25% de la RD) que los que tienen electrodos rectos, que están más próximos de pared lateral.

5. Los pacientes que tienen electrodos colocados en una posición perimodiolar tienen impedancias más bajas.

6. La colocación del electrodo de referencia tiene un efecto significativo sobre el consumo en mediciones in vitro usando un implante de prueba.

7. La posición del electrodo de referencia tiene un efecto funcional sobre la intensidad de la estimulación en las mediciones in vitro utilizando un implante de prueba.

8. La integración de un físico en una unidad de hipoacusia puede, por lo tanto, dar lugar a descubrimientos innovadores y multidisciplinares en el campo de la otología, como lo demuestra la variedad de estudios realizados durante este estudio de doctorado.

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VIII. TABLE INDEX

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TABLE INDEX

Table 6.1. Pearson correlation results for the comparison of autocorrelation of words pre and post filtering and their total score. The only significant correlation was found for the case 5 dB SNR.

Table 6.2. Pearson correlation of age, years working as ATC and overall test results. No correlation was found between tests results and either age or experience. Obviously, in the vast majority of cases, the older the ATC, the longer they have worked, so there is a correlation between these two variables.

Table 6.3. Correlation between years working, age and individual test results. Significant inversed correlations were found for: a) 8 kHz 5 SNR and years working; b) 8 kHz 5 SNR and age; c) for 22 kHz 5 SNR and age.

Table 7.1. Demographic data

Table 7.2. Descriptive statistics.

Table 7.3 Descriptive statistics of the variables: electrode-inner wall distance, NRT, T level and impedance.

Table 7.4. Pearson correlation of the variables: success rate, electrode-inner wall distance, NRT, T level and impedance.

Table 7.5. Estimation using the MCSM, MLCSM, LBGLM for the variable success rate.

Table 8.1 Current density as a function of bone thickness. The variables show an inverse relationship that is consistent with current findings from the previous method used in this study.

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Table 8.2 Resistance, voltage, current, and power consumption of each RE position of the temporal bone measurements.

IX. FIGURE INDEX

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FIGURE INDEX

Figure 2.1. A sound wave is caused by the compression and rarefaction of the particles in the medium along which it propagates. This displacement pattern takes the form of a sinusoidal wave when represented along the time axis. Part b shows the parameters that define a wave: amplitude, frequency, period, phase and phase constant.

Figure 2.2. Frequency determines how fast the particle returns to the equilibrium position, in other words, how fast it vibrates.

Figure 2.3. Sound intensity decreases as distance from the source increases because the same initial radiated energy is distributed through a larger area.

Figure 2.4. Relative response of each weighing SPL system as a function of frequency. It can be seen how A is most sensitive to frequencies up to 1000 Hz, while B and C tend to flatten out sooner.

Figure 2.5. When two single frequency waves are combined, the result is a waveform that is the sum of them.

Figure 2.6. Fast Fourier Transform of function f(t). a) f(t) in the time domain shows a sound wave as a function of time. b) FFT of f(t) gives a function F(f) which is a frequency profile of f(t).

Figure 3.1. Anatomy of the ear. The pinna and external auditory canal form the outer ear, which is separated from the middle ear by the tympanic membrane. The middle ear houses three ossicles, the malleus, incus and stapes. Together they form the sound conducting mechanism. The inner ear consists of the cochlea, which transduces vibration to a nervous impulse and the vestibular labyrinth that houses the organ of balance.

Figure 3.2. The ear canal acts as a tube open at one end, which forces the closed end to be a node and the open end and antinode. This causes the minimum resonant frequency (1st harmonic) to be four times the length of the ear canal as. The next harmonic that follows these boundary conditions is an odd multiple of this resonant frequency.

Figure 3.3. Anatomy of the cochlea with a close-up diagram (a) and a real histological image (b) of the organ of Corti, the three Scalae and the spiral ganglion (14).

Figure 3.4. Basilar membrane vibration for high and low frequencies. The maximum displacement point along this membrane can be seen to be frequency dependent, thus making the cochlea a fine tuned frequency analyser.

Figure 3.5. Hair cells have stereocilia arranged in a directional way, which causes hiperpolarization of the cell when bent in one direction, and depolarization when bent the opposite way. These changes release neurotransmitters that are received by the neural ends and sent to the brain as auditory input.

Figure 3.6. The auditory nerve will tend to fire at a particular phase of a stimulating low-frequency tone. So the inter-spike intervals tend to occur at integer multiples of the period of the tone. With high frequency tones (> 5kHz) phase locking is not possible, because the capacitance of inner hair cells prevents them from changing voltage sufficiently rapidly.

Figure 4.1. Temporal and place coding of sound. In temporal coding (left), neurons fire action potentials in phase with the sound waves. Place coding (right) refers to the perception of pitch depending on the site of stimulation. This is possible because the auditory cortex that responds to sound is arranged tonotopically.

Figure 4.2. Equal-loudness contours for loudness levels from 10 to 120 phones for sounds presented binaurally from the frontal direction. The absolute threshold curve (MAF) is also shown (19).

Figure 4.3. Basic structure of Moore et al. loudness model.

Figure 5.1. Main components of a cochlear implant: a) speech processor; b) transmitter; c) receiver/stimulator; d) electrode array.

Figure 5.2. Block diagram of a CI system. Audio signal is acquired by the processor microphone and then amplified. Then it is passed to a filter bank in order to separate it into different frequency bands. The output of each filter is then passed through an envelope detector. The dynamic range adaptation block then transforms the acoustic dynamic range for each channel into the electrical dynamic range necessary for each electrode. Finally, the processor generates the stimulation pulses representing the current level to be presented at each electrode and at each time instant.

Figure 5.3. Simultaneous electrode stimulation effect. A, and B represent single electrode stimulation patterns. C shows the overlap caused by simultaneous stimulation.

Figure 5.4. Block diagram of the CIS strategy. The signal is pre-amplified and filtered into frequency bands. Full-wave rectification and low-pass filtering then extract the envelopes of the filtered waveforms. Then the signal is compressed to fit the patient's dynamic range and then modulated with biphasic pulses. The biphasic pulses are transmitted to the electrodes in an interleaved way.

Figure 5.5. Growth function of discharge rate versus electrical pulse intensity.

Figure 6.1. Butterworth filtered FFT of a single word. The original signal is passed through a filter that removes the high frequency part of the spectrum, leaving the lower components unaltered.

Figure 6.2. Preparation of sound stimuli:6 sets of lists were divided into two groups (A and B) and three noise files were generated at 60, 65 and 70 dB. Each of the lists in group A were played along one of the three noise conditions. Group B lists and three noise files of the same characteristics as before were first filtered using a Butterworth filter. Similarly, each list was reproduced together with one of the three noise files.

Figure 6.3. Experimental set up. Subjects were seated in a sound treated room, 1m away from two adjacent loudspeakers. The word lists and noise files were played separately from each of the loudspeakers.

Figure 6.4. Example of cross-correlation results of words 1 and 2 of list A1. The x axis represents sound frequency and the y axis represents the degree of correlation of each frequency value.

Figure 6.5. Success Rate for each SNR condition in both filtered and unfiltered situations. Statistically better performance is observed when high frequencies are preserved for ±5 dB SNR ($p < 0.05$). However, for SNR= 0 no statistical difference is observed.

Figure 6.6. ATC age histogram.

Figure 6.7. Control group age histogram.

Figure 6.8. Years working as ATC histogram.

Figure 6.9. Success rate of ATC and control groups as a function of the test. The ATC group outperforms the control group in all SNR conditions and both in non-filtered (A=22 kHz) and filtered (B=8 kHz) conditions.

Figure 6.10. Success rate as a function of years working as ATC.

Figure 6.11. Success rate as a function of age for the ATC group.

Figure 7.1. Case of Use Diagram. The patient interacts with the researcher by providing answers for each of the test steps: T and C level calculation (MCL and THL, respectively), loudness balance and electrode discrimination test.

Figure 7.2. Data Base schematics. Tests are nested within their previous required step, the variable patient being the main sorting variable.

Figure 7.3. Insert Patient. Here demographic, clinical data and electrode position with respect to the modiolus can be registered.

Figure 7.4. Stimulus definition window. PPS: pulses per second; Stimulation mode: ground electrode location (1 for ground electrode in the mastoid, 2 for internal processor case and 1+2 for a combination of both); # PPS: number of pulses per second (defines stimulus duration); silence: defines time lapse between pulse trains.

Figure 7.5. Dynamic Range Map. C and T levels for each electrode are determined using the up-down method.

Figure 7.6. Loudness Balance window. Adjacent electrodes are loudness balanced using the confluence method.

Figure 7.7. Patient view for electrode loudness balance. The patient has to state which stimulus sounds louder.

Figure 7.8. Alternative forced choice experiment for electrode discrimination.

Figure 7.9. Success rate histogram representation

Figure 7.10. Success rate mean values and standard deviations per patient (pac).

Figure 7.11. Relationship between the mean electrode-inner wall distance per patient and mean success rate.

Figure 7.12. Comparison of performance between an average and a low score patient using the three models calculated. The LBGLM model has the most appropriate shape, showing ceiling and floor effects for perfect performance (value one) and lowest performance (value 0).

Figure 8.1 Geometries used in the numerical model. The interelectrode bone layer thickness was modified to study its effect on impedance, and thus, electrical consumption.

Figure 8.2 Schematics of the implant circuit. The reference/probe electrodes were attached to the analog ports of the microcontroller board. Switchers S1–S6 define which probe to select as RE. S12 disconnects or reconnects the intracochlear electrode. Resistors R1–R7 all have a value of 1.5 kΩ.

Figure 8.3. a-c. Real images showing the three RE positions studied: a) at the root of the zygomatic arch of the temporal bone; b) in the standard temporal bone position; and c) far away from the cochlear in the mastoid region.

Figure 8.4. Electric field lines as a function of bone thickness. As the RE is placed closer to the intracochlear electrode, the field lines show higher intensity in the space between them, where the auditory fibers are located. The cross represents the origin of the coordinate system (0,0), and the white lines represent electric field contours of equal value.

Figure 8.5. Current as a function of inter-electrode distance at the Y-axis (perpendicular to the electrode longitudinal circular axis). Each line represents a bone thickness (Case 1 being the one with the highest value and Case 3

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with the lowest). All of them follow exponentially decaying profiles, although the values are higher for the case where bone thickness is lowest (Case 3).

Figure 8.6. Current as a function of inter-electrode distance at the X-axis (along the electrode longitudinal circular axis). Each line represents the same bone thick- ness as in the previous figure. The three cases follow a similar profile, although the lowest thickness one is the one showing the highest current values.

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X. APPENDIX

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APPENDIX I: ETHICS COMMITTEE APPROVALS

V O L C @ N
Plan Estratégico del CHUIMI
« 2013 - 2016 »

Gobierno de Canarias

Servicio Canario de la Salud Complejo Hospitalario Universitario Insular-Materno Infantil	**Certificado**	**Informe Protocolo:** Id:CEIC-CHUIMI-2014/693

Vicente Olmo Quintana, Presidente del Comité Ético de Investigación Clínica del Complejo Hospitalario Universitario Insular-Materno Infantil,

C E R T I F I C A:

Que este Comité, en la sesión celebrada el 27 de febrero de 2014, evaluó la propuesta del Dr. Ángel Ramos Macías, Servicio de Otorrinolaringología, Cabeza y Cuello del CHUIMI, para la realización del estudio de investigación abajo señalado, y que dicho estudio fue aprobado el día 5 de marzo de 2014, constando como investigadora colaboradora **Dª Mª Teresa Pérez Zaballos.**

> **"Importancia de las frecuencias agudas en la percepción del habla en ambiente ruidoso en personas normo-oyentes"**
>
> Protocolo: Versión 1 de febrero de 2014
> HIP y CI: Versión 1 de febrero de 2014

Lo que firmo para que conste a los efectos oportunos.

Las Palmas de Gran Canaria, a 26 de abril de 2016

PRESIDENTE DEL COMITÉ

Dr. D. Vicente J. Olmo Quintana

VOLC@N
Plan Estratégico del CHUIMI
« 2013 - 2016 »

Gobierno de Canarias

Servicio Canario de la Salud Complejo Hospitalario Universitario Insular-Materno Infantil	**Certificado**	**Informe Protocolo:** Id:CEIC-CHUIMI-2014/744

Vicente Olmo Quintana, Presidente del Comité Ético de Investigación Clínica del Complejo Hospitalario Universitario Insular-Materno Infantil,

C E R T I F I C A:

Que este Comité, en la sesión celebrada el 30 de octubre de 2014, evaluó la propuesta del Dr. Ángel Ramos Macías, Servicio de Otorrinolaringología, Cabeza y Cuello del CHUIMI, para la realización del estudio de investigación abajo señalado, y que dicho estudio fue aprobado el día 27 de noviembre de 2014, constando como investigadora colaboradora **Dª Mª Teresa Pérez Zaballos.**

> **"Estudio psicoacústico sobre la discriminación tonal en los implantados cocleares".**
>
> Protocolo: Versión noviembre 2014
> HIP y CI: Versión noviembre 2014

Lo que firmo para que conste a los efectos oportunos.

Las Palmas de Gran Canaria, a 26 de abril de 2016

SERVICIO CANARIO DE LA SALUD
LAS PALMAS

PRESIDENTE DEL COMITÉ

Dr. D. Vicente J. Olmo Quintana

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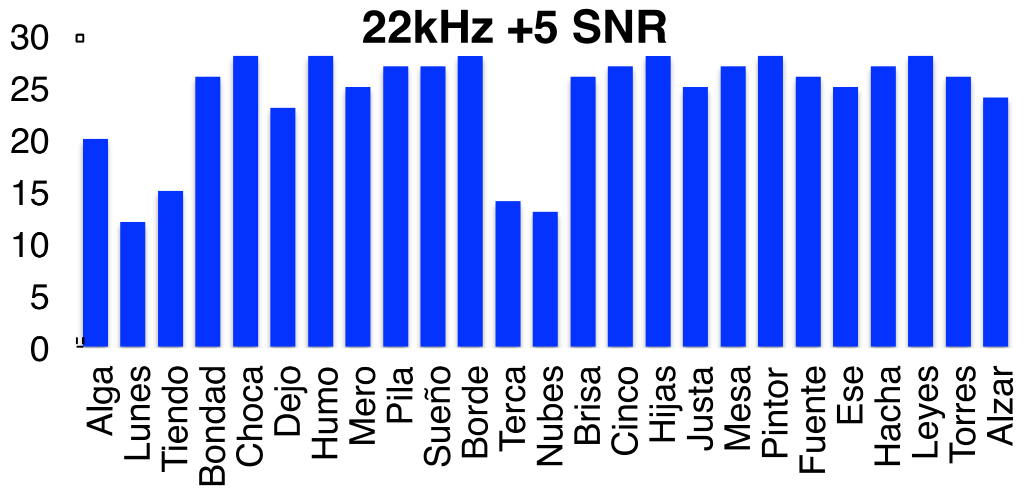
APPENDIX II

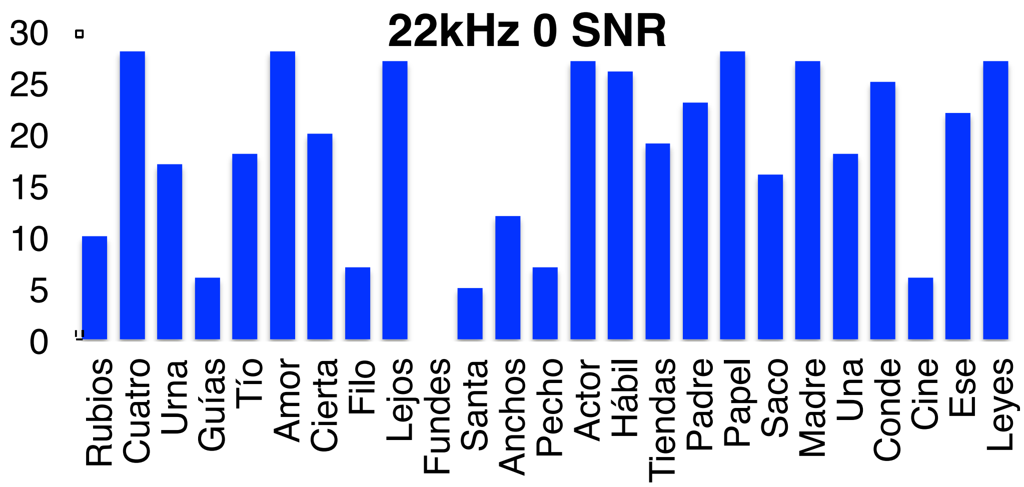
Lists of words used for each SNR and filtering/non filtering condition.

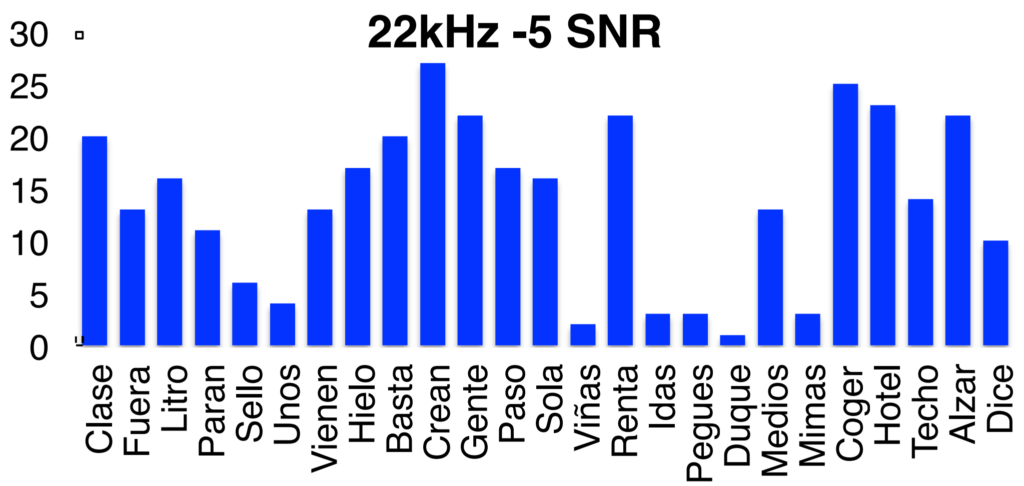
8kHz +5 SNR	8kHz 0 SNR	8kHz -5 SNR	22kHz +5 SNR	22kHz 0 SNR	22kHz -5 SNR
Boina	Mujer	Tardes	Alga	Rubios	Clase
Ecos	Portal	Anís	Lunes	Cuatro	Fuera
Real	Tierra	Cedo	Tiendo	Urna	Litro
Doce	Quince	Crema	Bondad	Guías	Paran
Prisa	Hotel	Guapa	Choca	Tío	Sello
Cero	Resta	Luces	Dejo	Amor	Unos
Dure	Yema	Pelas	Humo	Cierta	Vienen
Jabón	Alga	Ruegas	Mero	Filo	Hielo
Mosca	Canto	Vuelas	Pila	Lejos	Basta
Toser	Correr	Cientos	Sueño	Fundes	Crean
Flema	Fuerza	Llaves	Borde	Santa	Gente
Fuerte	León	Tía	Terca	Anchos	Paso
Donde	Mulo	Bajo	Nubes	Pecho	Sola
Seda	Queso	Curas	Brisa	Actor	Viñas
Chino	Valles	Hierba	Cinco	Hábil	Renta
Juego	Veinte	Mantel	Hijas	Tiendas	Idas
Caspa	Jefe	Perros	Justa	Padre	Pegues
Año	Sede	Sartén	Mesa	Papel	Duque
Ligo	Sastre	Fleco	Pintor	Saco	Medios
También	Alma	Saca	Fuente	Madre	Mimas
Tinte	Higos	Coche	Ese	Una	Coger
Nunca	Puso	Cada	Hacha	Conde	Hotel
Hilos	Diga	Tiempo	Leyes	Cine	Techo
Olla	Día	Montón	Torres	Ese	Alzar
Primas	Piso	Noche	Alzar	Leyes	Dice

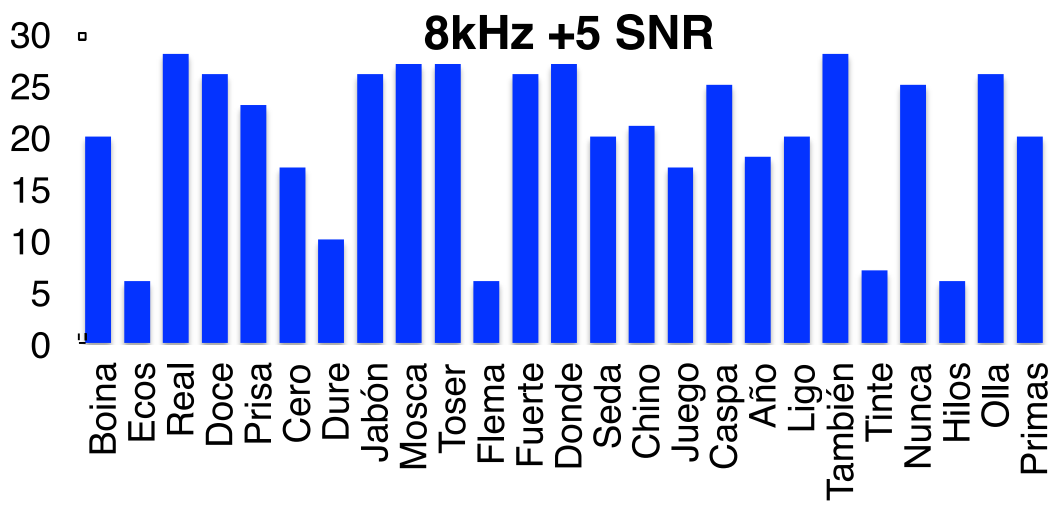
APPENDIX III

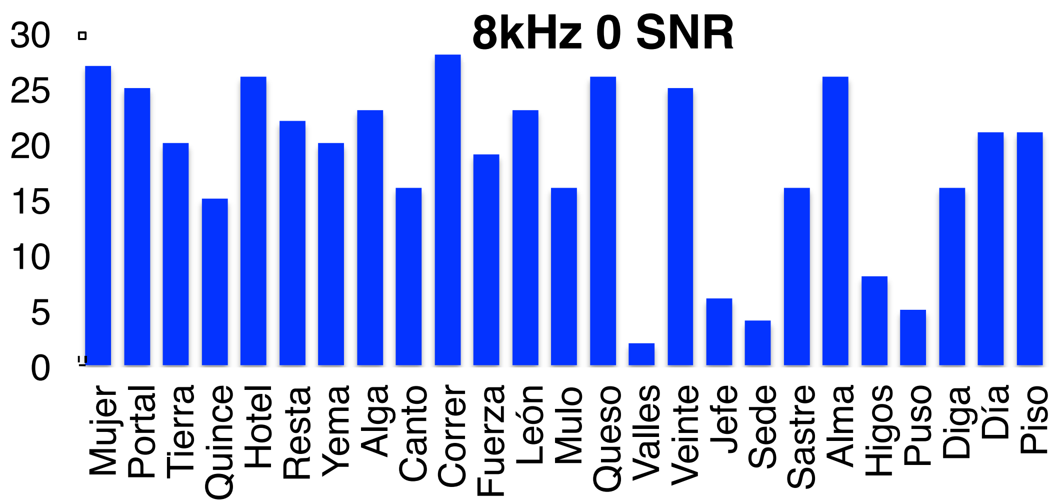
Individual word scores per list. The words were extracted from the validated Spanish Disyllabic Speech material for hearing loss assessment. As expected, word identification decreases as noise increases. It can also be observed how word recognition is higher in the cases ±5 dB SNR.








8kHz +5 SNR
30
25
20
15
10
5
0
Boina
Ecos
Real
Doce
Prisa
Cero
Dure
Jabón
Mosca
Toser
Flema
Fuerte
Donde
Seda
Chino
Juego
Caspa
Año
Ligo
También
Tinte
Nunca
Hilos
Olla
Primas


8kHz 0 SNR
30
25
20
15
10
5
0
Mujer
Portal
Tierra
Quince
Hotel
Resta
Yema
Alga
Canto
Correr
Fuerza
León
Mulo
Queso
Valles
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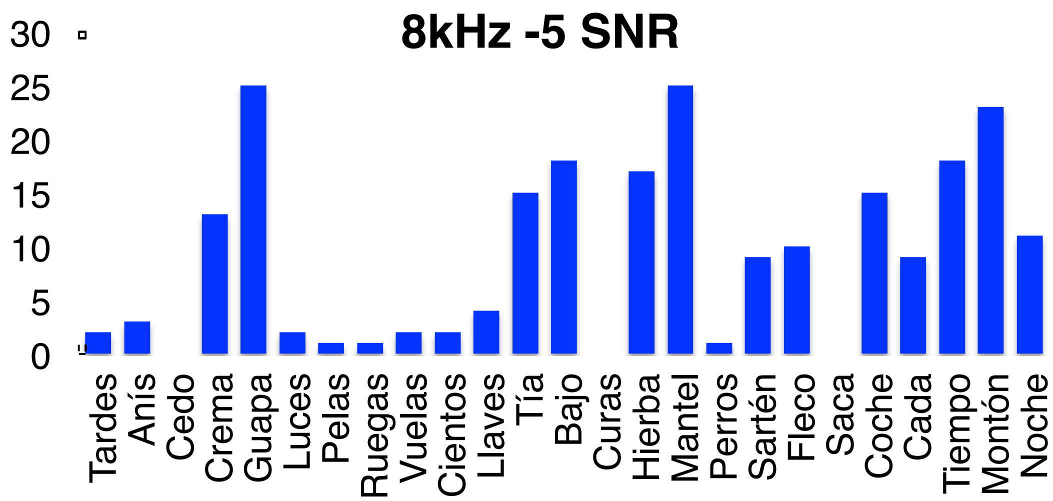
Sede
Sastre
Alma
Higos
Puso
Diga
Día
Piso


8kHz -5 SNR
30
25
20
15
10
5
0
Tardes
Anís
Cedo
Crema
Guapa
Luces
Pelas
Ruegas
Vuelas
Cientos
Llaves
Tía
Bajo
Curas
Hierba
Mantel
Perros
Sartén
Fleco
Saca
Coche
Cada
Tiempo
Montón
Noche

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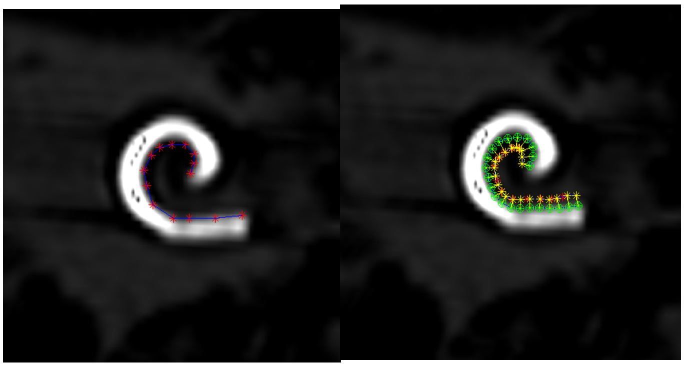
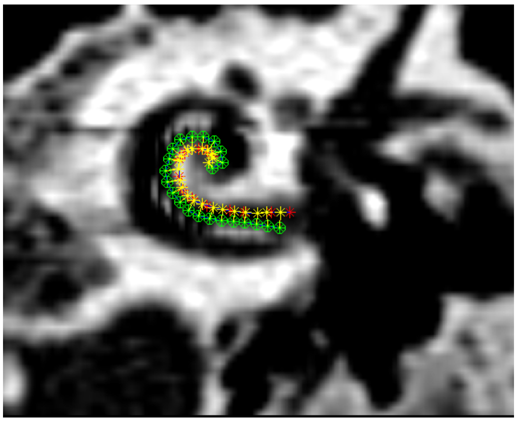
APPENDIX IV

Individual image treatment to obtain electrode-inner wall distance for each of the participating patients. For patient 2, the first and second images are shown for illustration of the procedure. For the rest of the patients, only the final image is shown.



Patient 2



Patient 3

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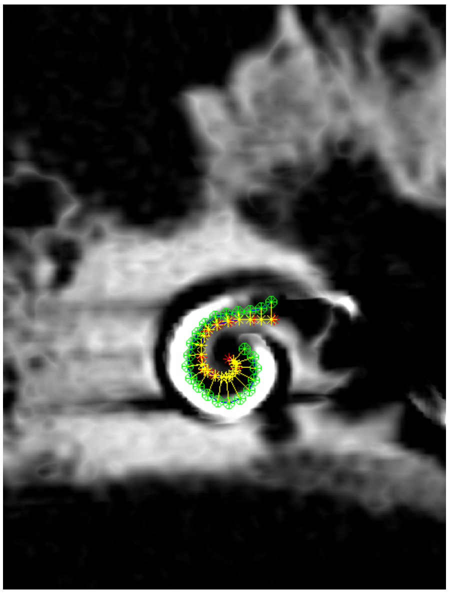
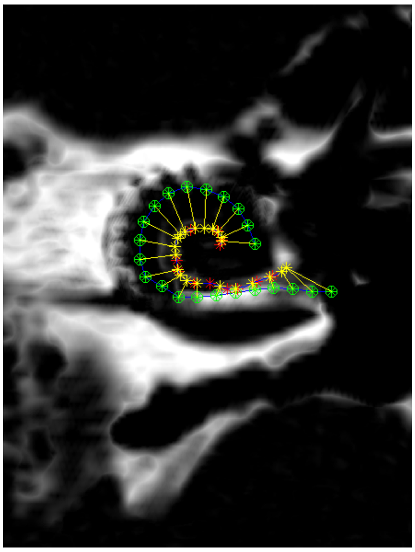


Patient 4



Patient 5

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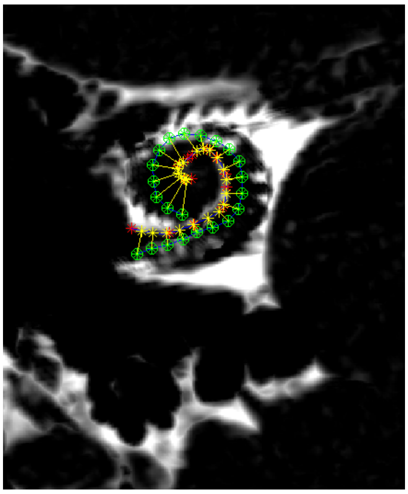
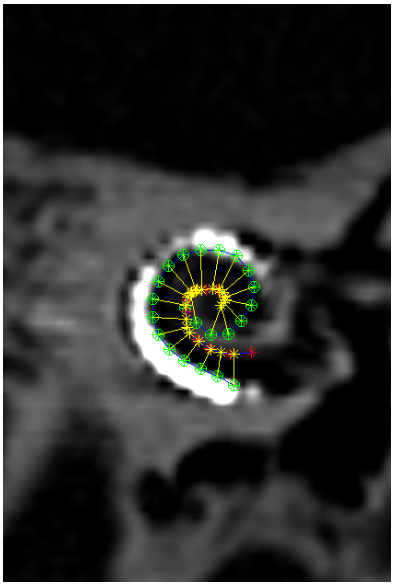


Patient 6



Patient 7

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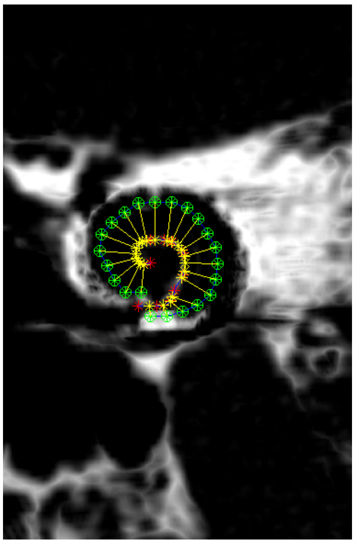
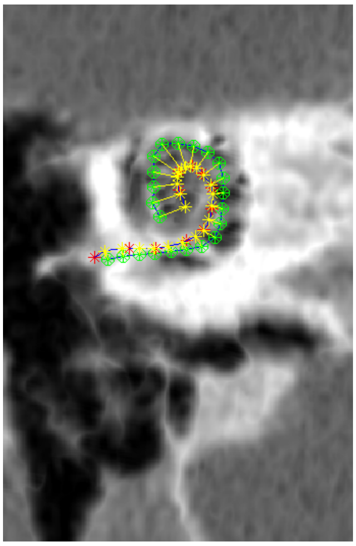


Patient 8



Patient 9

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APPENDIX IV

Certificate of authorship

Las Palmas de gran Canaria, a 22 de Abril de 2016

En cumplimiento de los siguientes artículos del REGLAMENTO DE ESTUDIOS DE DOCTORADO DE LA UNIVERSIDAD DE LAS PALMAS DE GRAN CANARIA, Aprobado por el Consejo de Gobierno el 17 de diciembre de 2012 (BOULPGC de 9 de enero de 2013) y modificado por el Consejo de Gobierno de 23 de octubre de 2013 (BOULPGC de 4 de noviembre de 2013)

Artículo 6.- Derechos y obligaciones del Doctorando, el Tutor y el Director de la Tesis Doctoral. Punto 2.

c. "a figurar, en los términos que se establecen en la legislación vigente sobre la materia, como autor o coautor en todos los trabajos, artículos o comunicaciones en los que haya participado durante su formación doctoral, particularmente los que se produzcan a partir de los resultados de su tesis doctoral o en otros en los que se expongan resultados de la investigación en los que su aportación pueda considerarse sustancial y efectiva."

d.- "a ser reconocido, en los términos que se establecen en la legislación vigente sobre la materia, como titular o cotitular de los derechos de propiedad intelectual que se produzcan a partir de los resultados de las investigaciones en las que haya participado durante su formación doctoral, particularmente los que se produzcan a partir de los resultados de su tesis doctoral o de

otros que se produzcan a partir de resultados de la investigación en los que su aportación pueda considerarse sustancial y efectiva."

Artículo 8.- Autorización y Depósito de la tesis doctoral. Punto 2

La Comisión Académica del Programa de Doctorado, en un plazo máximo de 15 días, dará o no la conformidad para su tramitación. Para ello, la Comisión Académica establecerá criterios objetivos que incluirán, al menos, la obligatoriedad de que el doctorando haya producido, como primer autor o autor principal, una contribución científica (publicaciones en revistas, libros o capítulos de libro, patentes u obras artísticas) de calidad contrastada y que contribuya al sostenimiento de Programa de Doctorado, derivada de la Tesis Doctoral, antes de su depósito. Para acreditar la condición de autor principal, esta deberá ser reconocida por el resto de los autores de la contribución científica, al mismo tiempo que ninguno de los otros autores podrá presentar la misma contribución científica en calidad de autor principal para obtener la conformidad para la tramitación de su propia tesis doctoral.

Yo María Teresa Pérez Zaballos, con DNI 44737183K, con el objetivo de presentar para su defensa la tesis doctoral titulada

Integración de principios físicos en una unidad de hipoacusia

(Integration of physical principles in a hearing loss unit)

y, dado que algunos de sus resultados ya han sido publicados en las siguientes publicaciones:

RELACIÓN DE PUBLICACIONES. Deben citarse de forma completa.

1) de Miguel, Á. R., Zaballos, M. T. P., Macías, Á. R., Barreiro, S. A. B., González, J. C. F., & Plasencia, D. P. (2015). Effects of High-Frequency Suppression for Speech Recognition in Noise in Spanish Normal-Hearing Subjects. Otology & Neurotology, 36(4), 720-726.

2) Ramos-Miguel, A., Ramos-Macías, A., Artiles, J. V., & Zaballos, M. T. P. (2015). The Effect of Reference Electrode Position in Cochlear Implants. J Int Adv Otol, 11(3), 222-8.

3)Zaballos, M. T. P., de Miguel, Á. R., Plasencia, D. P., González, M. L. Z., & Macías, Á. R. (2015). Effects of long-term speech-in-noise training in air traffic controllers and high frequency suppression. A control group study. J Int Adv Otol, 11(3), 212-7.

4) Zaballos, M. P., de Miguel, A. R., Killian, M., & Macías, A. R. (2016, February). A Psychophysics experimental software to evaluate electrical pitch discrimination in Nucleus cochlear implanted patients. In Journal of Physics: Conference Series (Vol. 689, No. 1, p. 012030). IOP Publishing.

Con la finalidad de dar cumplimiento al apartado b, del punto primero de artículo 12 del REGLAMENTO DE ESTUDIOS DE DOCTORADO DE LA UNIVERSIDAD DE LAS PALMAS DE GRAN CANARIA, adjunto la firma de los coautores de las publicaciones que conforman la tesis, reconociendo cada uno de ellos y, por tanto, acreditando con dicha firma mi participación como autor principal en cada uno de los trabajos, y renunciado por parte de los firmantes al uso de dichas publicaciones como núcleo principal en otras tesis.

Apellidos, Nombre	DNI	Firma
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Silvia Andrea Borkoski Barreiro	X9332001 6	

Daniel Pérez Plasencia	43752506C	
Juan Carlos Falcón González	42827044D	
Mathijjs Killian		
Angel Ramos Macías	42811433S	
Jaime Viera Artiles	54082221-K	
María Luisa Zaballos González		

Daniel Pérez Plasencia		
Juan Carlos Falcón González		
Mathijjs Killian		Cochlear
Angel Ramos Macías		
Jaime Viera Artiles		
María Luisa Zaballos González		

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