The Influence of Hearing on Posture

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The candidate confirms that the work submitted is his own and that appropriate credit has been given where reference has been made to the work of others.

Dedication

In memory of my grandparents, Mohammad and Alwa Alshamrani.

You left fingerprints of grace on my life.

You shall not be forgotten.

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Upon the completion of this project, I am appreciative to a number of people who have had great contribution to this research. Without their help, hardly would I have finished the work. I am grateful to my most respected supervisors Dr Nicholas Thyer and Dr Ruth Brooke for their precious time, tolerance, expertise and also for helping me get through my stressful moments. They have provided me with extensive advice and guidance on the project from its earlier stages right to the end.

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Abstract

Maintaining upright posture requires integration of somatosensory, visual and vestibular information, and traditionally considered an automatic and effortless task. Recent research indicates that postural maintenance requires attention which interacts with other common tasks that share cognitive resources. This thesis investigates the influence of one such task, listening, on posture control and explores the novel idea that increased listening effort arising from hearing impairment uses extra attentional resources leaving less for posture control leading to an increased risk of positional or situational instability.

A dual-task study to explore the relationship between listening and posture suggested that listening has a destabilizing effect on posture control. This effect was detected by mean velocity, a centre of pressure measure. To further understand this effect, the mechanisms underpins this relationship was studied using stabilogram diffusion analysis. The results suggested that open-loop and closed-loop strategies were utilised to maintain upright posture. This relationship was further investigated under simulated hearing loss conditions, and the results revealed increased postural sway and longer open-loop times before switching to closed-loop mechanisms.

The deterioration of postural performance whilst listening may be explained by the idea that working memory has limited resource capacity to execute tasks, and that both tasks here requires attention and are competing for cognitive resources. Therefore, when the simulated hearing loss was introduced, it exacerbated this effect suggesting that performance was compromised due to the raised listening effort. It was concluded that mean velocity was sensitive to changes in postural sway resulted from performing a concurrent listening task. It was also concluded that stabilogram diffusion analysis would provide a comprehensive understanding of the postural strategies individuals adopted to maintain erect posture under normal and simulated hearing loss listening conditions.

List of Abbreviations

Normal Stance **(NS)**

Romberg Stance **(RS)**

Normal Stance eyes Open **(NSO)**

Normal Stance eyes Open with Auditory listening task **(NSOA)**

Normal Stance eyes Closed **(NSC)**

Normal Stance eyes Closed with Auditory listening task **(NSCA)**

Romberg Stance eyes Open **(RSO)**

Romberg Stance eyes Open with Auditory listening task **(RSOA)**

Romberg Stance eyes Closed **(RSC)**

Romberg Stance eyes Closed with Auditory listening task **(RSCA)**

Centre of Pressure **(COP)**

Centre of Mass **(COM)**

base of support **(BOS)**

Anterior-Posterior direction **(AP)**

Medio-Lateral direction **(ML)**

Standard Deviation **(SD)**

Mean Amplitude **(MA)**

Standard Deviation of Sway Amplitude **(SDSA)**

Movement Range **(MR)**

Standard Deviation of Velocity **(SDV)**

Phase Plane Portrait **(PPP)**

Mean Velocity **(MV)**

Planar Deviation **(PD)**

Sway Area **(SA)**

Total Phase Plane Portrait **(TPPP)**

Analysis of Variance **(ANOVA)**

Working Memory **(WM)**

Reaction Time **(RT)**

Central Nervous System **(CNS)**

General Practitioners **(GPs)**

Digital Signal Processing **(DSP)**

Sensorineural Hearing Loss **(SNHL)**

Simulated Hearing Loss **(SHL)**

Table of Contents

List of Tables

List of Figures

1 Chapter One: Introduction

1.1 Introduction

Postural control is defined as the capability to maintain upright stance by maintaining the centre of gravity within the area of the base of support (Nashner 1981) for the purpose of balance and orientation (Shumway-Cook and Woollacott 2007). Postural control is essential to complete nearly all human activities. However, with approximately two third of the body mass found above the trunk, no one stands entirely still and so the human body is constantly moving. In other words, we are not naturally stable (Winter, Patla and Frank 1990).

Postural control is maintained through the integration of internal and external forces and environmental factors (Yaggie and McGregor 2002). It is a dynamic activity with multiple controlling systems that change within the individual and with the surrounding environment (Winter, Patla and Frank 1990; Oie, Kiemel and Jeka 2002). It requires a complex processes between musculoskeletal and sensory systems (Shumway-Cook and Woollacott 2012). There are three sensory systems involved in the maintenance of posture, the visual, vestibular and somatosensory systems. These three systems help to control posture and equilibrium as they adapt, reweight, depending on the task and environment (Woollacott, Shumway-Cook and Nashner 1986; Shumway-Cook and Horak 1986; Horak 2006).

Normally, the central nervous system (CNS) integrates and organises information from various sensory receptors and activates the suitable motor responses to maintain postural control (Peterka 2002). Information regarding the state of the body and its surroundings from the visual, vestibular and proprioceptive systems is gathered concurrently and transferred from the sensory receptors to the CNS via afferent pathways, where it is continually monitored to update postural control through the muscular system (Borghese and Calvi 2003; Yardley 1994).

Vision plays an important part in maintain posture by delivering information regarding the position and movement of the head in relation to the environment (Lopez et al. 2006). The vestibular apparatus, on the other hand, provides information regarding head position and speed of head movement with the aim of stabilising vision. The proprioceptors provide the system with the relative location of body parts to one another and position and movement of the body with respect to the surrounding environment (Lackner and DiZio 2005).

Maintenance of posture, although a complex process, is typically controlled below the threshold of conscious awareness; as such it is seemingly thought of as an automatic and effortless task that does not require attention (Neumann 1984). However, several studies indicate that it requires attention as it involves executive functions and that there can be interactions between maintaining correct posture and performing other cognitive tasks as they share common cognitive resources (Andersson, Yardley and Luxon 1998; Balasubramaniam and Wing 2002; Camicioli et al. 1997; Kerr, Condon and McDonald 1985; Lajoie et al. 1993; Lundin-Olsson, Nyberg and Gustafson 1997; Maylor and Wing 1996; Shumway-Cook and Woollacott 2000; Stelmach, Zelaznik and Lowe 1990; Teasdale et al. 1993; Jamet et al. 2004; Riley, Baker and Schmit 2003; Andersson et al. 2002; Woollacott and Shumway-Cook 2002; Fraizer and Mitra 2008).

These studies have revealed that when postural control and cognitive tasks are performed simultaneously they will compete for attentional resources. The amount of attention required for postural control appears to vary depending on the tasks (both postural and cognitive) being performed i.e. tasks' level of difficulty as well as other intrinsic factors such as age and sensorimotor expertise.

Humans usually maintain their posture, either static (standing or sitting) or dynamic (walking or running), while performing other cognitive tasks. One such task is listening. Listening is a pertinent aspect to this thesis. Sounds are all around us and there is a difference in the way these sounds are processed – some are heard, some are listened to, and some are comprehended requiring an active response. Nonetheless, all of these sounds form part of a communication situation.

The hearing, listening, comprehension and communication of these sounds are categorised as different functions (Kiessling et al. 2003; Pichora-Fuller and Singh 2006) and are reported to require attention as they involve executive functions (Mishra et al. 2010; Rudner et al. 2011; Rudner, Rönnberg and Lunner 2011; Sorqvist and Ronnberg 2012; Rönnberg et al. 2013b). Hearing can be described as an automatic auditory process, allowing for the perception of sounds, and the information about them such as their pitch, location or volume.

Conversely, listening is the function of intentionally hearing a sound and giving it full attention. It is an active mental process which requires concentration as it occurs at a conscious level and involves interpretation of what is being heard for the purpose of receiving information. Since hearing can be described as ability, happens as a passive process and requires minimal processing (occurs at a subconscious level), it can be assumed that listening involves more mental processing since it is a skill which happens as an active process (Kiessling et al. 2003; Pichora-Fuller and Singh 2006).

Beyond hearing and listening is the activity of comprehending. Comprehension is a function which enables us to understand the meaning and intent of received information. It allows us to understand and follow instructions or information in a message we have heard, and for this reason, it can be reasonably concluded that comprehension requires yet more concentration and effort than both hearing and listening (Rudner et al. 2013). The final function is communication, which is the passing of information, ideas, or intention between any number of individuals. Effective communication requires each participant to hear, listen, and comprehend (Kiessling et al. 2003; Pichora-Fuller and Singh 2006).

The processes involved in the ability to hear and perceive sounds are known as bottom-up processes, while those responsible for comprehension and communication are referred to as top-down processes (Zekveld et al. 2006; Besser et al. 2013; Davis and Johnsrude 2007; Avivi-Reich, Daneman and Schneider 2014).

To illustrate, bottom-up processes include automatic and unconscious responses such as the acoustic analysis and coding of speech, whereas top-down processes are responsible for linguistic processing and the representation of internal speech

(Zekveld et al. 2006). Top-down processing is also used when trying to understand speech in a noisy situation, or other problematic listening environment (Pichora-Fuller, Schneider and Daneman 1995).

Recently, the relation between sound and posture has gained increasing interest. Some researchers have reported a stabilizing effect of sound on posture and some others have also reported a destabilizing effect of sound on posture. Discrepancies in previous research have been referred to number of reasons including but not limited to: nature of sound, investigated populations and experimental protocols. The literature concerning the relation between sound and posture is covered in more details in Chapter Two, section 2.4.

Although the influence of sound on postural control has been reported by many researchers (Juntunen et al. 1987; Raper and Soames 1991; Sakellari and Soames 1996; Soames and Raper 1992; Kilburn, Warshaw and Hanscom 1992; Park et al. 2011a) the effect of listening on postural control is less well-documented. The above-mentioned studies have investigated how posture control is influenced by pure tones, noise, and moving or stationary sound signals. Thus, these investigations have examined the effect of sound on posture from a sensory point of view, but with minimal consideration given to listening attentional demands. These studies was conducted with the aim of investigating whether sound has an effect on posture, to explore what type of sounds produce this effect and if sound cues helpful in maintaining upright posture especially in populations requiring posture rehabilitation.

Listening, unlike passive hearing, extends beyond hearing sensitivity thresholds measured by hearing tests (Jerger and Musiek 2000; Iliadou, Chermak and Bamiou 2015) It is the day-to-day situations that individuals are exposed to, such as speech perception, that is known to be related to cognition (Akeroyd 2008; Gatehouse, Naylor and Elberling 2006a; Gatehouse, Naylor and Elberling 2006b; Ronnberg 2003), or like attending to complex sounds in an acoustically challenging environment. The relation between cognition and listening, especially in adverse listening situations such as listening in the presence of background noise (Mattys et al. 2012), hearing loss (Stenfelt and Rönnberg 2009) or when the cognitive demands of listening are increased (Mattys et al. 2012) has been established. Investigations of this relation revealed that listening requires cognitive processing and that adverse listening conditions similar to the ones described above requires more cognitive resources to be processed i.e. more or raised effort. This raised effort is reported to be more noticeable in individuals with hearing loss, according to both subjective (Gatehouse and Noble 2004) and objective measures (Zekveld, Kramer and Festen 2011).

Since previous research revealed that 1) both, maintaining upright posture and listening, require attention and that they involve cognitive processing and executive functions and that 2) sound has an effect on posture, it is important now to understand if listening, which presumably requires more cognitive processing than hearing, has an effect on posture control. This line of research will investigate the relation between listening and posture control using a dual-task paradigm and explore the attentional demands listening might have on posture control in normal individuals under normal and simulated hearing loss conditions.

Exploring the relation between listening and posture will allow a better understanding of the nature of this relation in the presence of hearing impairment. It might help explain other listening and posture control health-related problems, for example, falls, reduced situational awareness and the attentional cost of hearing loss. It may also inform the way that current rehabilitation plans are provided; evidence may be delivered to support a link between the relationship between listening, maintaining upright posture and the risk of falls, and it may also provide evidence supporting a link between this relationship and the low uptake of hearing aids for some individuals.

1.2 Thesis focus

As described previously, maintaining upright posture and listening both require attention, mental processing and involve executive functions in order to be executed successfully, either alone or simultaneously. Numerous psychological experimental studies (Baddeley 2000; Cowan 2005; Kahneman 1973) have suggested that humans have limited mental capacity and processing resources, and that if the task or tasks being performed require processing resources more than ones capacity, task execution is affected and hence task performance will deteriorate.

Although postural control, sound and posture and the impact of hearing impairment are well-established in the literature, it is apparent that thus far, minimal research has been directed towards the attentional cost of 1) listening and 2) listening in the presence of hearing impairment and its relationship to posture control in adults. As the relation between listening and posture control in adults is currently unclear, the questions that this thesis addresses come from linking three central ideas, namely: 1) humans have limited cognitive processing resources (Kahneman 1973; Baddeley 2012; Cowan 2005; Wickens 1984; Wickens 2002; Spence and Driver 2004), 2) maintaining upright posture requires cognitive processing (Shumway-Cook and Woollacott 2000), and, finally, 3) listening requires cognitive processing (Anderson Gosselin and Gagne 2011; Collins and De Luca 1994; McGarrigle et al. 2014) and it even demands more cognitive processing, i.e. raised effort, in the presence of hearing impairment (Gosselin and Gagné 2010; Shinn-Cunningham and Best 2008; Strauss et al. 2010; McCoy et al. 2005; Wingfield, Tun and McCoy 2005).

To understand how listening influences posture control in the presence of hearing impairment, this process has to be understood and the concept be established in healthy adults with normal hearing first, and measures capable of detecting this effect have to be identified. The following section will illustrate the structure of the thesis and outline the role of each chapter.

1.3 Thesis structure

In **Chapter One**, this chapter, an introduction of the current research, thesis focus and thesis structure were provided. **Chapter Two** will provide a brief background concerning the human auditory system, balance system and maintenance of erect posture. This will be followed by a review of the literature concerning posture and cognition, posture and sound, and working memory (WM) models for information processing in humans. Finally, the aims and objectives of this thesis will be stated.

Chapter Three covers the first experimental chapter that addresses the first aim of this thesis. In this chapter, the influence of listening on posture control in normal-hearing individuals will be investigated. Using a dual-task paradigm, posture control while listening will be studied and body sway is computed and different global sway measures are calculated. This chapter will explore whether listening has any effect on posture control and if any, it will identify the tools suitable to measure this effect.

Chapter Four comprises the second experimental chapter. It will describe a dual-task experiment which was designed to examine the influence of listening on postural control in normal-hearing individuals. Body sway is computed and then analyzed using a method called Stabilogram Diffusion Analysis (SDA) which will allow to study the body sway behavior during listening using a structural body sway measure. This structural sway measure should provide insight into the underpinning strategies and mechanisms that normal individuals use to maintain upright posture whilst listening.

Chapter Five covers a hearing simulation method which was used to develop a listening task with simulated hearing loss in order to address the third objective of this research. In this chapter, hearing simulation was introduced and the selected method of simulation was analyzed, validated and an experiment was conducted to ensure that the developed listening task with simulated hearing loss was of an accurate representation of hearing loss. The developed task was employed in the final experimental study presented in Chapter Six.

Chapter Six details the third experiment, which investigates the influence of listening on posture control in normal-hearing individuals using the listening task with simulated hearing loss developed in Chapter Five. In this chapter, global and structural sway measures, found to be sensitive to the effect of listening on posture control in Chapter Three (Experiment One) and Chapter Four (Experiment Two), will be calculated. This study employs a dual-task paradigm.

Chapter Seven discusses general thesis conclusions, highlighting the experimental works' main findings, presenting the thesis implications and also identifying future research avenues based on the current results.

2 Chapter Two:

Background and Literature Review

2.1 Introduction

The beginning of this chapter will contain a short overall picture of the human hearing system and the methods humans use to regulate their posture. Subsequently, extant literature on how sound and posture are connected and cognitively processed will be submitted, before an outline of the aims and objectives is drawn and a statement of the thesis' research questions is delivered.

2.2 Human hearing

2.2.1 The hearing system

There are numerous structures that make up the human ear. Its complexity can be categorised as three principal entities: the outer, middle, and inner ears (Figure 2-1). The experience of sound occurs via a set of connected episodes, as follows: the pinna focuses air vibrations and multiplies them in the external auditory canal, from where they hit the tympanic membrane. This subsequently pulsates in line with the sound's waveform (Behrbohm, Verse and Stammberger 2009). There is a sequence of three bones that are connected to the tympanic membrane (these being the malleus, incus and stapes, or the ossicles, collectively); they proceed in line with the tympanic membrane's pulsations. The stapes, which is the final link in the ossicles' chain, is connected to the cochlea, the fluid-filled, sensory hearing organ, by a composition called the oval window; this is a pliable membrane that divides the middle and inner ears.

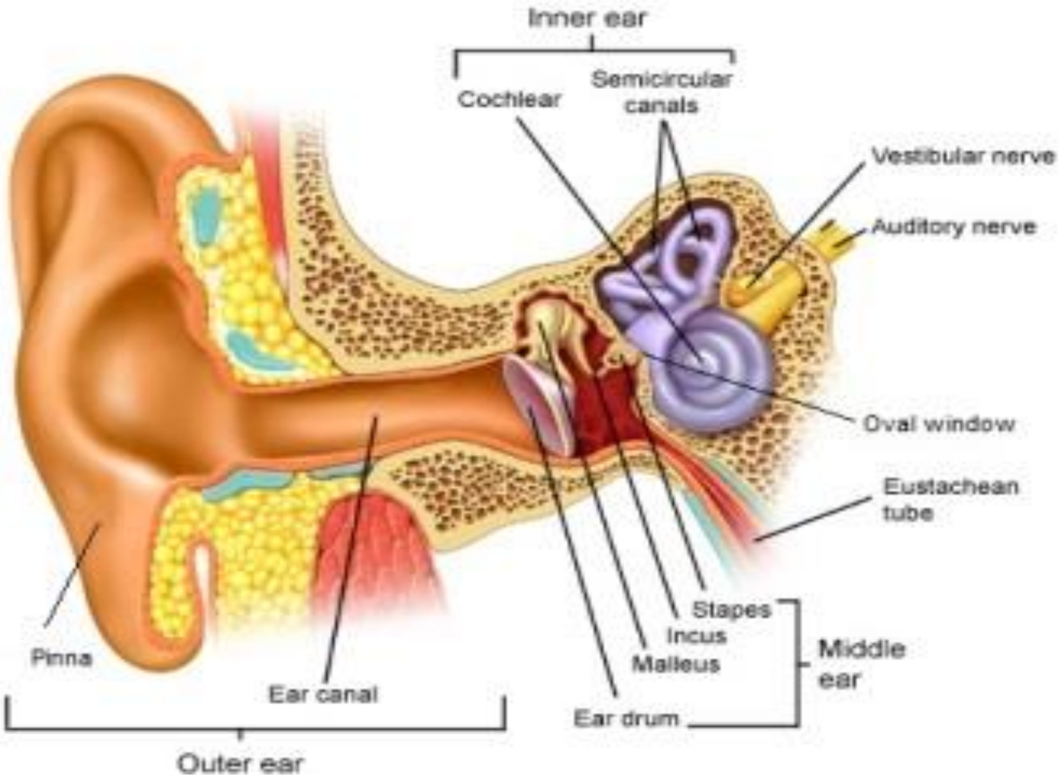


Figure 2-1 The anatomy of the ear. Adapted from Introduction to Audiology (Stach 2008).

The ossicles have a combined purpose with the tympanic membrane – to surmount an impedance misalignment of the middle ear, which is full of air, and the inner ear, which is full of fluid, and thereby enable the satisfactory transference of sound to the inner ear (Moore 2007). Impedance misalignment, or impedance mismatch, means the variation in terms of resistance between the level of air (low impedance) and the extent of the inner ear's fluid (high impedance); this means that were sound to act on the oval window, the majority of it would just reflect reflexively, resulting in a significant leak (between 30 and 40 dB) in acoustic energy (Goode 1986).

Inside the cochlea are three chambers, otherwise called scalae; the oval window is joined to the scalavestibuli. Waves result from the stapes' lateral movement; these waves are compatible with the acoustic source and they move through the fluid inside the cochlea. In turn, the waves incite the cochlea's different membranes to move, but the movement ceases at the round window.

Fundamentally, the cochlea is cylindrical in shape, split evenly down its reach via a configuration called the Organ of Corti. Within the Organ of Corti there are lines of two separate hair cell forms (stereocillia), these being set down the cochlea's length: Inner Hair Cells (IHCs) and Outer Hair Cells (OHCs). When fluid moves inside the cochlea, a configuration inside the Organ of Corti, the basilar membrane (where the stereocillia can be found), pulsates, which deflects the separate forms of stereocillia. There is a gradual physiological change in the basilar membrane, from the cochlea's bottom to its crown; the basilar membrane is fairly narrow and rigid at the bottom, but more flexible and wider at the top. Because of the mechanical characteristics of moving waves, various frequencies create peaks at diverse positions on the basilar membrane. Such an arrangement is called tonotopic , which indicates that the arrangement of the cochlea is progressive (high–low) and frequency-specific (Seikel, Drumright and King 2015).

The IHCs' deflection incites their depolarisation, resulting in action potentials being despatched to higher auditory centres from the cochlear nerve (Stach 2008). Considering that the cochlea's physiology is frequency-specific, the response by the IHCs to a particular frequency will be dependent on their position in the cochlea. This feature is also replicated to some extent in the OHCs, but rather than transmitting auditory data to higher auditory centres, they have an active function inside the cochlea: they are constituents of an efferent feedback loop, utilising somatic electromotility (they oscillate down their full length via cell body pulsations) (Brownell 1990). The OHCs' movement raises the level at which the basilar membrane pulsates to specific frequencies, thereby serving to amplify and fine-tune the response to frequencies inside the cochlea. IHCs can thus be defined as the 'true sensory cells of the inner ear', because sensory data on sound is not relayed to higher auditory centres by OHCs (Yanz 2002). There is then transference of nerve impulses resulting from IHCs to the brainstem and auditory cortex, from where they can be interpreted. Having established the workings of the human hearing system, the next section will explore the question of how to evaluate whether its function is regular, as well as looking at how to measure people's hearing thresholds.

2.2.2 Measurement of hearing thresholds

Pure tone audiometry is the accepted technique for hearing evaluation in a clinical setting. This involves the deployment of an audiometer, which is an instrument that can generate sinusoidal pure tones over various frequencies at a given intensity. The audiometer determines the lowest sounds capable of being heard by someone between the frequencies of 250 and 8000 Hz, these being defined as absolute thresholds. Pure tones are created; they vary in frequency and intensity and have a duration of between one and three seconds. Subjects are invited to press a button whenever they hear such sounds; the frequencies used are 0.25, 0.5, 1, 2, 4 and 8 kHz, and each ear of a patient is tested individually. To determine the absolute threshold, the staircase technique is deployed.

Usually, a soundproof booth is used to restrict interference from ambient noise in a room, and the plotting of each ear's absolute thresholds is marked on an audiogram. The respective thresholds for air conduction and bone conduction are evaluated as part of the testing, air conduction being determined via headphones that allow the passage through the whole auditory configuration of outer, middle and inner ears. Bone conduction, on the other hand, is measured through the placing of a pulsating pad over the mastoid bone, which is located at the rear of the outer ear. This enables sound to be multiplied by vibration in the skull, circumventing the outer and middle ears and transmitting direct to the inner ear, thereby evaluating fundamental sensory ability. When the absolute thresholds of air and bone conduction differ, this tells of a loss of conductive hearing. An average, normally from data at 250, 500, 1000, 2000, and 4000 Hz, is gleaned from the hearing thresholds to determine the level of impairment that an individual may have, and the standard UK clinical classification guidelines are shown in table (2-1).

Average hearing threshold (dB HL)	**Descriptor**
20 – 40	Mild hearing loss
41 – 70	Moderate hearing loss
71 – 95	Severe hearing loss
> 95	Profound hearing loss

Table 2-1: UK hearing loss severity guidelines. Data at 0.25, 0.5, 1, 2, and 4 kHz are the sources from which the average thresholds for each specific ear are determined. Source: (British Society of Audiology 2011).

Despite the fact that pure tone audiometry measures physiological capacity to some extent, it cannot yield data on the extent of impairment to various hair cell types inside the cochlea (that is, it cannot distinguish between damage from OHCs and damage from IHCs). Nor can it establish whether the central auditory processing deficit is influenced. Considering the range of functions allied to the various configurations that the auditory system exhibits (e.g., frequency, temporal and loudness processing), a prediction of the level to which someone may encounter effects in everyday life as a result of sensory impairment cannot be made. The degree to which the ear can detect two separate frequencies, for instance, is not particularly discernible by audiometric thresholds (Simon and Yund 1993). In fact, hearing sensitivity standards like pure tone audiometry are frequently inadequately aligned with hearing ability records that are more subjective (Weinstein and Ventry 1983; Newman et al. 1990; Stephens and Zhao 1996; Nondahl et al. 1998).

To determine the impact of the loss of hearing in a way that is more valid, ecologically, the two principal techniques used have been the testing of speech and people's own self-reported experiences. Speech is tested by a previously arranged set of sentences or words being relayed to the subject at various intensities, after which the subject must audibly repeat what they heard. A plot can then be made of the percentage accurate score as a function of intensity, which becomes a speech audiogram. The test validity can be further augmented by assessing how speech is perceived in the presence of noise, thereby yielding an assessment of the capability to obtain significant auditory data from an entity that masks the background.

To help determine how big a problem a person may have with their hearing, self-reported experiences are also employed, the method used being conducted via a validated questionnaire to give structure to the process. When hearing loss is self-reported, it relates the degree of impact that hearing loss has on the way that a person carries out actions and roles. This leads to the explanation of the phenomena of objective and subjective tests not always being linked, even though links might be expected to exist (Weinstein and Ventry 1983; Newman et al. 1990; Stephens and Zhao 1996; Nondahl et al. 1998).

There can be variations in hearing loss effects despite people sometimes presenting absolute thresholds that are exactly the same (Halpin and Rauch 2009), the reason being that pure tone audiometry is incapable of discerning the impairment of various configurations in the auditory system, instead producing a yardstick of overall function. A variety of perceptual results can occur because of separate elements of the auditory system being impaired (e.g., if IHCs were completely lost, the consequences would be completely different than if OHCs were totally lost). In Section 2.2.4, there will be a concise overall picture given of the perceptual consequences that occur following what is referred to as Sensorineural Hearing Loss (SNHL), but before the implications of hearing loss are examined, the different forms of hearing loss will be explored.

2.2.3 Hearing loss

Hearing loss can be defined as complete or partial inability to hear sound in one or both ears. Total hearing loss means that the people affected cannot hear any sound, while people with partial hearing loss have a sense of sound, but it can be of lower intensity or distorted because their auditory system is impaired. If the structures inside the outer, middle, or inner ear are compromised by harm or abnormalities, loss of hearing can result. It can also occur if higher auditory centres are similarly compromised. When hearing loss takes place at the inner ear's gross division (or thereafter), it is referred to as the condition SNHL, whereas conductive hearing loss happens at (or before) the middle ear's gross division. Nevertheless, both conditions can simultaneously impact an ear if the air conduction thresholds are higher than normal, i.e., when they are > 20 dBHL and gaps of > 10 dBHL between air and bone conduction thresholds. Mixed hearing loss is the term used for this condition.

The condition of hearing loss does not result just from a sound's attenuation, but also from its distortion (Plomp 1986). Sound attenuation is a consequence of the auditory system's sensitivity deficit; sound distortion results when the sounds that are above the hearing threshold decline in quality. As regards the performance of the various structures that are involved in hearing, there is a greater complexity to SNHL's perceptual consequences than those resulting from conductive hearing loss (Moore 2007). The latter tends to just lead to a deficit in auditory sensitivity, considering that the impact on the auditory system concerns how it may overcome the misalignment in impedance between the middle ear, which is filled with air, and the inner ear, which is filled with fluid. An active element operates inside the inner ear, (Moore 1996a), which indicates that the result of the structures being harmed is not just less audibility, but sound distortion.

Sound distortion is significant, as there can be no reversal of it. Although increased audibility can be achieved by using rehabilitative measures like hearing aids, sound distortion's influence on the comprehension of speech continues (Moore 1996a), even though sound may be clearly audible. This phenomenon is pronounced by elements such as background noise (Ricketts 2001). Sometimes, increased levels of sound can result in less understanding, rather than more, for people with hearing loss (Studebaker et al. 1999).

Many adults experience hearing loss (Roth, Hanebuth and Probst 2011). UK adults between the ages of 18 and 80 experience a rate of hearing loss of one in 6.1, the loss being a minimum of 25dB in the less affected ear; one in 12 the loss being a minimum of 35dB in the less affected ear and one in 17 a minimum of 40 dB (Akeroyd, Foreman and Holman 2014). Akeroyd, Foreman and Holman (2014) figures were determined from data extracted from a national hearing study (Davis 1995). These data indicated a rise in hearing loss of 12% across two specific decades. Because the condition of hearing loss is associated with age (Davis 1995), its growth seems to be accelerating as the number of older people increases (Laplante-Lévesque, Hickson and Worrall 2010). Additionally, younger people's relationship with sound (e.g., extended use of music devices) may indicate that damage to hearing could turn into a significant issue for younger people (Niskar et al. 2001; Crandell, Mills and Gauthier 2004; Chung et al. 2005).

2.2.4 Impact of Sensorineural hearing loss on perception

The perceptual outcomes from sensorineural hearing loss vary from person to person, even if their absolute hearing thresholds are identical. The consequences of sensorineural hearing loss will be detailed further in the following subsection.

2.2.4.1 Threshold elevation and loudness recruitment

The hearing loss element that pure tone audiometry determines is elevation of absolute threshold. This relates to the condition of sound having to be increased before a person with hearing loss registers it. An additional difficulty with sensorineural hearing loss is loudness recruitment, which the majority, if not the entirety, of people with SNHL display to some degree (Moore 2007). Loudness recruitment concerns an extraordinary increase of perceived loudness when the physical sound level reaches intensities over the absolute threshold. Elevated absolute thresholds and loudness recruitment can result in several consequences such as: lower dynamic range, changes in loudness cues and an absence of auditory information.

2.2.4.2 Reduced frequency selectivity

The auditory system's capacity to delineate the constituents of a sound constructed of numerous items is frequency selectivity (Moore 2007). Should a person with normal hearing hear two distinct, pure tones, for instance, they could distinguish each separate tone, but for people with sensorineural hearing loss, there is less frequency selectivity – wider frequency bands, in other words – which makes the sound quality distorted. As it has a significant function in many features of auditory perception, frequency selectivity is a key element of the auditory system (Moore 2007). Lower frequency selectivity can result in various consequences: effective listening is greatly affected by extraneous masking noises, perception of spectral components in sound is lower, and there is damaged timbre appreciation.

2.2.4.3 Reduced temporal resolution

The capability to discern alterations in sound over a period of time is known as temporal resolution (Moore 2007). Low levels of temporal resolution are evident in people with SNHL, which is a result of impairment of some of the internal cochlea structures, resulting in temporal processing being affected. People with damaged hearing, for example, show reduced propensity to: 1) discern when a sound has gaps (Fitzgibbons and Wightman 1982); 2) discern gap length (Irwin and Purdy 1982); 3) notice a low-volume sound after a high-volume sound (Moore 2012); 4) identify the authentic fluctuation of sound; instead, they can register these as internal gaps (Glasberg and Moore 1992).

Such effects on the temporal processing capacity can disadvantage individuals with hearing loss when they try to comprehend speech, especially when there is ambient background noise. Moore (2007) contends that the majority of everyday noises exhibit rapid amplitude fluctuations from instant to instant, and when temporal processing is impaired, people suffering hearing loss will struggle to keep up with these sounds' temporal profile. Additionally, in situations where the acoustics are challenging – a surfeit of background sound – a person's capacity to temporally ascertain the source of interest of a sound may be adversely affected by noise that varies and shifts, and which therefore competes with the sound source of interest.

2.2.4.4 Conclusion of impact of SNHL on perception

This last section has demonstrated how SNHL's consequences hugely exceed a simple sensitivity loss inside the auditory system. While SNHL constitutes a lesser capability to register acoustic data, various difficulties emerge in terms of perceiving sound when it reaches audibility. This problem in the ability to process and comprehend auditory data has caused authors to conjecture that people with SNHL must expend extra effort when listening if they are able to enjoy effective auditory perception (Hornsby 2013). This extra effort employed in listening when the acoustics in an environment are difficult, or when conditions are tough, or when there is hearing loss, is an extremely relevant issue for this thesis. Therefore, how the ability to listen in such conditions affects cognitive processing is further examined in section 2.5.2. Now that human hearing has been introduced, the following section should provide a background about the other pertinent aspect of this thesis, that is posture.

2.3 Human balance and posture

2.3.1 Definitions

Posture control, as a term, is often confused with balance and equilibrium, which can be unhelpful, so it is useful to establish correct definitions of balance and posture before moving on to the subject of postural control in the next section. The ability of the body to retain its centre of gravity above the supporting base may be termed as balance, whereas posture relates to the synchronisation of parts of the body and the body's limbs, as well as the guidance of body position (Highstein, Fay and Popper 2004). We can therefore state that balance can be achieved and retained through postural alignment. If balance and posture are not correctly maintained then the body can lose balance. Pollock et al. (2000) exquisitely stated postural control to be "the act of maintaining, achieving or restoring a state of balance during any posture or activity". Human posture will be examined in the next section, and elements of correct postural control will be examined.

2.3.2 Posture control

Expanding on the aforementioned definitions, postural control can be said to the method of regulating the spatial position of the body in order to afford stability and orientation (Shumway-Cook and Woollacott 1995; Shumway-Cook and Woollacott 2001). A person is thus compelled to retain their centre of mass (COM) constrained in their base of support (BOS). The BOS may be said to be the body area that connects with the support surface (this body area normally being the feet) and thus permits supporting forces to be generated: ground reaction forces at the level of the feet, for example (Binder 2009). The capability to keep a suitable correlation between the body's segments and the environment within which it is functioning is known as postural orientation (Shumway-Cook and Woollacott 2016). Keeping effective postural control depends on biomechanical constraints, motor strategies, sensory strategies, orientation-in-space and cognitive control (Horak 2006). In the ensuing subsections, the ways in which these elements support the maintaining of posture will be explored briefly.

2.3.2.1 Biomechanical constraints

The COM's movement depends on various biomechanical factors such as the height, weight and muscular configuration of a person (Chiari, Rocchi and Cappello 2002). The BOS's form, size and characteristics are also instrumental in the performance of balance, but it is the quality and magnitude of the BOS that is balance's biggest biomechanical constraint (Horak 2006). The BOS's breadth has a lot of bearing on the amount of COM sway and its direction. When a standard stance is adopted, for example, the areas of COM sway are compact and focused on the A-P direction, whereas when walking, the same areas are bigger, and mainly focused on the M-L direction (Winter 1991). Staying upright while standing demands that the body uses specific muscles: in the standing position, the line of gravity moves before the thoracic spine and neck elements, just to the rear of the hip joint, and before the knee and ankle joints, as the line indicates what muscular activity must take place so that the body can maintain a standing position, resisting gravity (Kendall et al. 2005). The process entails the involvement of muscles in the calf, hip and trunk; if these muscles are weak, imbalance may ensue.

2.3.2.2 Motor strategies

The principal motor strategies deployed in achieving posture are those that relate to the hip, ankle and stepping motion. Shear forces are involved in hip strategy, in order to retain the COM inside the BOS (Horak and Nashner 1986). If the hips are bent considerably during this process, altered positioning of the COM will occur. This is principally deployed when the COM is near the boundaries of stability or when the BOS is compact (Horak 1987). Using hip strategy demands rapid motions of the axial body elements and the gathering of trustworthy data from visual and vestibular systems. The advantage of using ankle strategy is that the COM once more becomes stable, because motions arranged around the ankle joints are used. This technique tends to be used in everyday actions, and is generally engaged when the person is on solid ground, with the COM and BOS close together. Stepping strategy, the third method, involves rapid motions of a lower limb in the falling COM’s direction. It thus shifts the BOS to wherever the COM is about to fall, in order to regain balance (Horak 1987).

2.3.2.3 Sensory strategies

Data from the vestibular, visual, and somatosensory systems is crucial for postural control. The particular collection of sensory inputs that are vital for postural control depend on a person’s characteristics, the prevailing environmental conditions and the individual postural control action that is occurring (Shumway-Cook and Woollacott 2016). Data for the CNS regarding the head’s motion and position relating to inertial and gravitational forces is given by the vestibular system, which supplies a frame of reference, as regards gravity, for postural control. Signals of this type are not sufficient on their own, though, to enable an accurate picture of the movement of the body through space. Extra knowledge about the head's movement and position is given by the visual system. Although visual senses are not strictly needed to keep balance, they are still very helpful. We can stay upright while our eyes are shut, but we sway more, which implies a lesser degree of postural control (Shumway-Cook and Woollacott 2016). Lastly, more data about the body's movement and position as regards the supporting surface is given by the somatosensory system; inputs of this nature to the body also relay knowledge about how body parts are inter-relating. Similar to vision, when somatosensory inputs are not optimal, either due to pathology or stance condition i.e. standing on a foam, body sway will increase resulting in less stability (Asai et al. 1990). In what follows, the way in which vestibular, visual, and somatosensory systems assist posture control will be highlighted.

2.3.2.3.1 Vestibular system

A gravito-inertial frame of reference for postural control is supplied by the vestibular system; it does this by discerning how the head's movement and position respond to gravity and inertial forces (Shumway-Cook and Woollacott 2016). The vestibular system is comprised of the saccule, utricle, and semi-circular canals of the inner ear. There is a dense and compact area in the saccule and utricle called the macula; this is the vestibular system organ that acts as a receptor (Coren, Ward and Enns 2004; Yardley 1994), from which hair cells stick out. The cells are covered in a glycoprotein layer, gelatinous and dense in texture, which is called the otolithic membrane. Across this membrane's surface there are calcium carbonate crystals called otoliths. As the head leans forward, gravity draws the otoliths and the otolithic membrane in front of the hair cells in the direction in which the head is leaning, which causes the hair cells to bend, thus inciting various nerve impulses (Tortora and Nielsen 2009; Baloh and Kerber 2011). Within the vestibular branch of the vestibulocochlear nerve, hair cells synapse with first-order sensory neurons. There are two branches to the vestibular nerve; the superior branch is the first, running directly alongside and parallel with the facial nerve in the internal auditory canal. The superior branch is the source of supply for the anterior and lateral semi-circular canals, and also the utricle. Second is the inferior branch, running directly alongside and parallel with the cochlear nerve; this branch innervates both the saccule and posterior semi-circular canals (Fetter 2000). Most of the nerve fibres feed off to the vestibular nuclei from here, although projections also go to the cerebellum and cortex (Coren, Ward and Enns 2004).

2.3.2.3.2 Visual system

Data on the head's position and movement with regard to adjacent objects is given by the visual system, which provides a verticality reference point (Shumway-Cook and Woollacott 2016). The process in a standard visual system is that light goes into the eye via the cornea, progresses to the anterior chamber, pupil, lens, and posterior chamber, finally reaching the retina (Schiller and Tehovnik 2015). The retina acts as a lining for the rear 75 percent of the eye and forms the visual pathway's start

(Tortora and Nielsen 2009). There are two different photoreceptors present: 1) rods, which are sensitive for black and white vision in low light conditions; and 2) cones, which receive colour vision. Upon stimulation the photoreceptor cells emit neurotransmitters that alter bipolar and horizontal cells; these are linked to amacrine cells and retinal ganglion (Schiller and Tehovnik 2015). This activity – the photoreceptor stimulation and consequent emission of neurotransmitters – incites nerve impulses being generated. Axons from retinal ganglion cells reach out behind to the optic disc, or blind spot, before leaving the eyeball at the optic nerve. From there, the pathway leads to the optic chiasm, the point at which the optic nerves cross, prior to transformation into the optic tracts. Following this, the optic tracts go into the brain and end in the lateral geniculate nucleus of the thalamus. Once at this point, they again synapse with neurons that create optic radiations, which subsequently project onto the cerebral cortex and occipital lobe's principal visual fields, thereby creating vision (Tortora and Nielsen 2009).

2.3.2.3.3 Somatosensory system

Data concerning the spatial positions of our limbs and body parts and the status of their movement is given by the somatosensory system (Blumenfeld 2002; Luxon and Davies 1997). Such information involves the proprioceptive sensations, which include the ankle, vibration sense, pain, and pressure. The tendons and muscles (mainly the postural muscles) are the sites in which proprioceptors are embedded. Primary sensory neuron axons take proprioceptive information into the spinal cord through the dorsal root ganglion and progress to the caudal medulla; once there they synapse with secondary sensory neurons before decussating. When decussating is complete, the secondary neurons progress their ascent and synapse in the thalamus. At that point, the neurons project to the primary somatosensory cortex in the postcentral gyrus (Blumenfeld 2002). The cerebellum receives proprioceptive data from the anterior and posterior spinocerebellar tracts. The cerebellum is thus appraised, via proprioceptive inputs, of motion, which the cerebellum can synchronise, polish and refine, in addition to keeping good posture and balance (Tortora and Nielsen 2009).

2.3.2.4 Orientation-in-space

A person's capacity to appreciate their body parts' spatial location and to navigate their surroundings so as to recognise and place objects satisfactorily is known as orientation-in-space (Massion 1992; van der Kooij et al. 2001). In order to stay balanced, there must be full integration of sensory information, which must be interpreted in light of a stable frame of reference that relates directly to the postural activity being enacted. People in good health experience that their nervous system adapts to the body's orientation-in-space as a matter of course, depending on what the activity is and its context (Horak 2006). Difficulties with this dimension of postural control may be the reason why people with otherwise normally operating sensory systems are still prone to dizziness (Cattaneo and Jonsdottir 2009).

2.3.2.5 Cognitive control

Previous experiences and environmental dangers are assessed by the CNS in the interests of keeping the right stance and choosing the relevant motor responses for the particular activity and environment. Achieving this needs the integrated actions of sensory collection, interpretation and action if the right balance is to be retained with a minimum amount of conscious effort. Such a procedure is automatic and needs the engagement of cognitive resources as well as the use of cortical regions. Even in apparently simple functions like standing still, cognitive processing is needed to retain balance (Horak 2006). Research on greater levels of difficulty in postural actions and dual task paradigms demonstrates the need for cognitive processing to be active in postural control.

The greater the difficulty in a postural action, the higher the level of cognitive processing will be needed. An action such as standing still, with the feet placed a shoulder width apart, seems an elementary action that would appear to need little effort, in theory, but if the postural action was more complicated, extra cognitive processing could result in more interference, necessitating more attention to be given to the action. Section 2.5.3 will provide more detail about the demands of cognitive processing and attention.

2.3.3 Measurement of posture

Physiological postural sway in a body that is standing upright in a static position can be specified as continuous corrective movements that enable the body's centre of gravity to retain postural control (Shumway-Cook and Woollacott 2016; Shumway-Cook and Woollacott 2001). An established line of thought is that in the action of standing still, the quantification of postural sway may be achieved by reporting the COP movement over a period of time (Chagdes et al. 2009). COP establishes the exact position of the resultant vertical ground reaction force (Winter 1995a), which enables it to be worked out. COP indicates the centre of body mass's trajectory and how much torque must be delivered to the support surface in order to regulate body-mass acceleration (Winter, Patla and Frank 1990). COG and COP vary in that centre of gravity (COG) is a variable that is directed by the postural control system, and is passive in nature (it cannot be obtained straight from a force platform) (Winter 1995b), while COP refers to the weighted average of the total pressures that ensue from the contact point with the support surface; unlike the COG, it is derived from a force platform (Winter, Patla and Frank 1990).

2.3.3.1 Parameters of COP

A number of COP parameters have been utilised in the attempt to define postural control, and there is controversy concerning the most appropriate measure to choose, as there are contrasting opinions on the measure that best registers sensitivity and records fluctuations inside the postural control system (Palmieri et al. 2002). The COP parameters that are most regularly used include but not limited to COP displacement, velocity and root mean square amplitude. When COP sway is higher, this has been taken to assume a lower level of postural control, but more contemporary studies indicate that in order to understand why there might be higher levels of sway, the environment needs to be taken more into consideration (Carpenter et al. 2001).

Calculations for evaluating balance strategies regularly assess the trajectories of COP or COM. It is a technique that uses force-plates to assess COP's projection and can also be engaged to evaluate COM displacement (Shumway-Cook and Woollacott 2016). The COP is the central point at which all the force that is being generated onto the supporting surface is distributed. It shifts around the COM on a continuous basis, in the interest of retaining it inside the BOS, thereby adding to stability. The sensitivity of this is very high: force-plates can register tiny movements that the naked human eye cannot perceive.

2.4 Hearing and posture

Despite the fact that it has not been a subject of interest for some time, in recent years there has been a growth in attention given to the affiliation between sound and posture. Literature on this subject is sparse, and even that seems to lack consistency and contain contradictions. Detailed investigation of this subject, however, indicates that apparent discrepancies in the studies are the result of varying conditions in the experiments that were conducted and the methods employed. In this section, an overarching view of the literature relevant to this subject will be given, foregrounding a number of hypotheses concerning how sound affects posture control.

There is common agreement that the maintenance of a person’s upright stance is dependent on the central system working to combine a number of sensory cues (Maurer, Mergner and Peterka 2006). There has been a lot of documentation of the function that visual, vestibular and proprioceptive inputs have performed. There has not been as much interest historically, however, in what part the function of audition in postural control has played, despite earlier research indicating that a deficit in auditory input can have an adverse effect on postural control (Era and Heikkinen 1985; Juntunen et al. 1987). In recent years, though, more interest has been focused on this subject (Gago et al. 2015; Gandemer et al. 2014; Mangiore 2012; Ross and Balasubramaniam 2015; Rumalla, Karim and Hullar 2015; Vitkovic et al. 2016; Zhong and Yost 2013). The results from all this research, from a variety of contexts, have indicated that posture can be profoundly affected by sound; specifically, when people assimilate auditory data it can assist in lessening postural sway. Gandemer et al. (2014), for example, conducted experiments involving 20 participants (12 men and 8 women); these experiments were held in a soundproof space, with the help of an Ambisonic system, which creates a field of sound around a subject to try and generate an experience of immersion in sound. The experiments were embellished through the use of a rotating low-pass, which served as a filter for any white noise or auditory stimulus. In the view of the authors, sounds that rotate tend to stabilise postural sway, a consequence that grew (improved postural control) with the enlargement of sound source velocity.

Mangiore (2012) conducted research whereby 13 subjects (5 males and 8 females), including three users of bilateral cochlear implants and 10 users of bimodals (i.e., having, oppositely, a cochlear implant and a hearing aid), allowed their postural control to be examined under amplification and then with amplification absent, via a stimulus of white noise in a variety of auditory circumstances (without sound, then with just the left and right speakers, then with front and rear speakers, and finally with the full set of four speakers). Calculations of the rate of body sway were made, from which it was deduced that the level of balance in users of cochlear implants was markedly better when an external sound source was used.

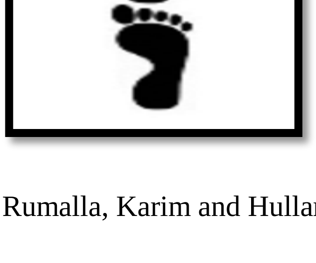
Another example is Vitkovic et al. (2016) exploration of the possibility that balance can be controlled via a hearing ‘map’ of the immediate environment. In a study involving 50 people in possession of normal hearing, 28 who had hearing loss and

19 who had vestibular dysfunction, exploration of how sound affected postural sway was made by analysing centres of pressure. The experiment made use of sound cues in the acoustic surroundings that could be either in evidence or absent. The results indicated that people with normal hearing made use of auditory cues to raise the quality of their postural sway. This capacity was lessened for people with hearing loss, although hearing aids seemed to redress the balance. There was much more use of auditory cues by people with extra vestibular deficits, prompting the possible conclusion that sensory weighting might enhance the deployment of auditory cues to compensate for sensory redundancy.

2.4.1 Hearing loss and posture

Primary research on what function audition has in postural control looked at how auditory loss affects postural sway. Era and Heikkinen (1985) demonstrated that young adults who had been subjected to a greater level of work-related noise exhibited more postural sway than others who had not experienced the same exposure. Confirmation of this came in later research by Juntunen et al. (1987), who examined how auditory deficits affected the postural control of soldiers, who had inevitably been exposed to sporadic, intense noise from guns. These subjects exhibited a much higher degree of body sway compared to the control group; this was much more pronounced in subjects who experienced extreme hearing loss.

Studies on workers (Kilburn, Warshaw and Hanscom 1992), children who were congenitally deaf (Suarez et al. 2007) and adults (Mangiore 2012) yielded similar findings. The area most explored, though, was loss of hearing in elderly people, with its attendant danger of falls (Viljanen et al. 2009; Lin and Ferrucci 2012). The explanation advanced about this by certain authors is that vestibular function deficit is universal, and that auditory loss resulting in impaired balance is just an expression of vestibular deficits. Research by Rumalla, Karim and Hullar (2015), though, made a comparison of the postural sway experienced by over-65-years-old users of bilateral hearing aids when hearing was aided and unaided respectively. Measurement of postural stability was taken under tandem stance: standing heel-to-toe, see figure 2-2, while the volunteers were subjected to broadband white noise at a level of 65 dB. The people in the experiment were asked to stand for three 30-second trials with their hearing aids switched on and off respectively. The aided state was much more beneficial to the people's postural performance than unaided, which confirms the advantage of auditory input being completely accessible. Lastly, Kanegaonkar, Amin and Clarke (2012) research made a comparison of how people in a variety of auditory environments experienced postural sway: in a standard room compared to a soundproof room, with and without ear defenders, eyes open and eyes shut. When their eyes were open, people experienced more sway when in a soundproof room compared to a standard room, and when using ear defenders, as opposed to not using them.



2-2 Tandem stance described by Rumalla, Karim and Hullar (2015)

The common feature of this research is that a shortage of auditory input leads to reduced postural control, a finding that might encourage people to incorporate sound in their postural control practice, facilitating interaction between sound and posture. As we shall observe in what follows, though, little research has yet been made in this area.

2.4.2 Sound has a stabilising effect on posture

Some research investigating static sound stimulation detected reduced sway when sound stimuli were present. Easton et al. (1998) research tested people who assumed a tandem Romberg stance (heel-to-toe position) and then had sources of sound placed on either side of their head, first with their eyes open, then with their eyes shut. When auditory cues were present there was a reduction of 10% in their sway, but this increased to 60% when visual cues were given. The lesser effect of sound compared to vision is very much evident, while in a later study, which also entailed people adopting the tandem Romberg stance, the result was reported that sway reduced by 9% when people were faced with a sound source in front of them (Zhong and Yost 2013). Zhong and Yost (2013) set out to explore if balance was strengthened when the auditory spatial cues were emitted from a particular source of sound, as opposed to when vision was used. Eight males and eleven females undertook postural activities (tandem Romberg and Fukuda stepping) with their eyes respectively open and shut when subject to a sound signal or broadband white noise, first when these were on and then when they were off. It was apparent from the results that a sound source in a fixed position can supply the central nervous system with enough spatial cues to improve the stability of the posture, but the visual cues were shown to be much stronger than the auditory ones.

Further research has dealt with how mobile sound sources function. Deviterne et al. (2005) study observed how old subjects responded to the rotation of sound stimuli around them. Two forms of rotating stimulations were contrasted: an auditory communication that had no particular meaning (a 440 Hz continuous tone) and a narrative with meaning – a brief recorded story. In this latter auditory form, the participants were requested to concentrate their attention on the story and then answer some subsequent questions. Following this experiment, the subjects experienced stabilisation: the fact that they had to concentrate on the sound compelled them to take the stimulation's rhythm and rotation into account; this facilitated auditory anchorage and thus stability of posture. Agaeva and Altman (2005) experiment was to have an arc of loudspeakers play sounds that travelled on a sagittal plane. While the sound travelled, the participants experienced slightly less postural sway, and exhibited an inclination to lean slightly forward when sound was present.

Loudspeakers were used to carry sound stimuli in all these experiments, therefore spatial data was able to be provided by the auditory stimulations, through local cues, on how much space was around the subjects (Blauert 1997); the outcomes were reported in terms of what auditory anchorage resulted: via the transmitted spatial data, the sources of sound produced landmarks, which helped the subjects to reduce body sway.

Headphones were used in other research, as auditory stimulation travelling through headphones is linked to the movement of the subject. In such instances, the sound cannot deliver spatial cues on a subject's surroundings., which could indicate why Palm et al. (2009) research involving 23 participants failed to detect any major fluctuations in postural sway when a subject was with or without headphones, listening to an auditory input such as instrumental music, which lacks specificity. Ross and Balasubramaniam (2015) recent work, however, showed a marked decrease in the body sway of subjects when in receipt of auditory stimuli via headphones. This experiment involved 19 participants in good health undertaking 20 30-second trails of stance in conditions ranging from eyes shut when silent, open when silent, closed in conditions of noise and closed under silence. The utilised stimulus was white noise that had consistent spectral density; this was produced by MATLAB and relayed via noise reduction headphones at 75dB. Variation of postural sway was evident when auditory noise was present, even though there was no visual data. The decreased sway cannot be seen to follow from spatial auditory cues being integrated. One of the ideas from the authors is that the phenomenon of stochastic resonance could be responsible for the results. Stochastic resonance happens when the data in a sensory signal is insufficient to come to the attention of the central nervous system, i.e., it is considered 'subthreshold'. In such circumstances, amalgamating noise - a signal without information - with the original sensory input increases the signal enough to raise it over the threshold and thereby integrate it. Proprioception has been known to result from this phenomenon: application to the soles of the feet by sub-sensory vibrations has exhibited markedly less postural sway (Priplata et al. 2002). Such an occurrence can also happen in audition, according to Ross and Balasubramaniam (2015). Although this is a promising development, the weakness of this idea is that no original auditory information was present that could be augmented through the addition of noise. In fact, the headphones worn by the participants were intended to decrease external noise, thereby under alternate noise and silence, no auditory data from the surroundings could be discerned by the subjects. Such results could indicate, however, that the phenomena engaged in reciprocal posture and auditory perception action could be multisensory and complicated. The results also posit that auditory noise could be used to rehabilitate people in certain circumstances.

2.4.3 Sound has a destabilising effect on posture

Several research experiments have failed to highlight how exposure to sound stimuli can decrease sway. One such experiment involved twelve subjects separated into two categories of age - some old, some young. They were subjected to a white noise auditory stimulus travelling right to left, then left to right: lateral body sway in the older group was more affected by lateral moving auditory stimulation compared to the young contingent (Tanaka et al. 2001). However, regulation of posture was not compared with sound absent or present respectively, meaning that comparison with previous studies was hard to achieve. Park et al. (2011b) study explored how a range of sound frequencies and amplitudes affected stability of posture. The outcome was that sway increased with higher sound frequency, but the absence of results without sound stimulation meant that comparison with previous results could not be effectively made.

A further set of studies entailed moving and static sounds (delivered through four loudspeakers) respectively: (Raper and Soames 1991; Soames and Raper 1992) highlighted a destabilizing effect of sound on the posture of subjects, in which the sound stimuli that were used were pure tone and background conversation. In the same vein, Gago et al. (2015) study showed how background noise could adversely affect standing subjects' regulation of posture. Gago et al. (2015) gathered 24 AD subjects. They matched age and health and made laboratory observations of postural stability; background noise was unremarkable and four conditions were set: standing on a level, hard floor for 30 seconds with eyes shut and eyes open, with ear defenders and without them. The outcomes were that postural stability was adversely affected by visual suppression, while deleting background noise markedly improved postural stability. The conclusion prevailed that audition is crucial for postural stability, but that background noise did not impart any information, and may have been more of a distraction than a complete absence of auditory data.

2.4.4 Summary of sound and posture

Different researchers have reported sound to have varying effects on stability: some say it improves stability, others say it impairs it. There are a range of reasons as to why this discrepancy in the literature occurs such as; experimental protocols and populations of interest. The sound source’s quality may be an important factor, and may result in variations in the literature. When the sound is uninformative, it seems that it cannot be amalgamated into the process of postural control. It can be confirmed from these findings that postural control is definitely affected by sound, and that it must be taken into account, just as other sensory modalities are.

2.5 Cognitive processing of sound and posture

This section will review the literature which explained how sound and posture are cognitively processed by humans and how they are successfully executed alone or concurrently. This section will begin by providing an overview of theories of information processing in humans which will then followed by subsections focusing on 1) processing sound and understanding speech and 2) maintaining upright posture and postural dual-task.

2.5.1 Information processing in human

Previously, there have been two principal approaches used in the modelling of dual-task interference: one focus on processing in serial and another on focusing in parallel. Parallel task processing is considered to be infeasible by serial-processing models, because processing capacity is restricted to just one stage of serial processing; in other words, it is a bottleneck. Resources at this point can only be delegated to one task at a time; the presumption is that there will be a priority task given preference, so the second task must be positioned in a sensory store prior to completion. Tasks cannot then be carried out simultaneously, which will result in the response to either or both being delayed.

Several authors (Broadbent 1958; Treisman 1960; Deutsch and Deutsch 1963) have put forward their theories as to where in the processing chain this might happen, and whether it happens at an early or late stage. There has been some support, empirically, for the counterpart to parallel processing models - serial-processing models, but satisfactory explanation as to why certain tasks might be simultaneously conducted and time-shared, with apparent minimal or zero decline in performance, has not been made (Edwards 2010). Consequently, these models are regarded as unsuitable for the work detailed in this thesis and no more discussion of them will take place thereafter.

In contrast, models that use parallel processing (Kahneman 1973; Wickens 1984), posit the view that there can be time sharing of these two tasks, but that there is a limited amount of resources available that might be allocated between them. If there is competition for resources between the tasks, at least one of them may be detrimentally affected in the view of some authors, so one way in which successful time-sharing can be accomplished is through central-resources models. Previous studies have suggested two possible models of parallel processing: One of these is a single resource model, which posits that people have just one bank of processing resources, and that the effort involved in conducting just one task imposes a strain on them. When the demands from processing tasks are more than the limited resources available, a decline in performance ensues.

Kahneman (1973) suggested this type of model, claiming that the resource pool allotted to a stage of processing or a task might be undifferentiated. Therefore, complete efficiency of time-sharing by the respective tasks is not completely expected by this model, but instead it predicts that if the totality of tasks conducted simultaneously raises the level of available resources too high, decline in performance will happen inevitably. Certain authors have expressed doubt as to whether this model might provide a sufficient explanation of the detailed form of attention, and disregarded it because of its comparatively simple nature (Niesser 1976; Allport 1993). The incidence of tasks resembling each other is so high-profile in dual-task research that it represents a further problem. Although separate modality tasks can undergo a dual-task cost, this can increase when there is simultaneous conducting of two tasks of identical modality (Lund 2002). Models that posit the processing of information by a range of resources tend to explain this finding better. Yet more authors have surmised that several single processing tools, or multiple resources, might exist, such pools being conceived as modules, or mental resources that are specialised (Navon and Gopher 1979; Allport 1980); these are separate resources of processing that are finite in capacity and are concerned with specific abilities or skills (Lund 2002). In such a scenario: for example, if a visual task and an auditory task have different modalities, they could still be conducted at the same time, as they do not use the same resources.

Although Allport (1980) concept backs particular studies that have identified instances where time sharing in distinct modality tasks has been successful (Shaffer 1975; Allport, Antonis and Reynolds 1972; Spelke, Hirst and Neisser 1976) it fails to illustrate why some findings show a dual-task cost related to two tasks that are modality-specific; in theory, each task should draw on separate processing resources. Additionally, these studies fail to articulate how many modules exist, or relate the entities with which the modules have to negotiate. Model testing in this area has problems, therefore, because the explanation of these results can be that a separate new model has emerged; in other words, the model is not falsifiable. Similarly, the question of the type of interaction between the models is not answered; co-ordination from the senses is needed, but there is no explanation of how the data from the modules is inter-connected. Navon and Gopher (1979) is more erudite: although they retain the idea of modules being specialised, they see information processing in terms of economics, holding that there may be trade-offs between modality performance, depending on what specific elements any situation might contain. How these distinct resources are cross-linked is not explained by such theories nor how there is occurrence of cross-modal links (Lund 2002). Wickens (1984), in contrast, additionally backs modules (Navon and Gopher 1979; Allport 1980), although his position is that the use of modules should be altered to allow them to negotiate separate elements of a specific task. Wickens' Multiple Resources Theory (MRT) (Wickens 2008; Wickens 1984; Wickens 2002) contends that modules function should be for separate task processing stages: input mode, processing mechanism, and output mode.

2.5.1.1 Multiple Resources Theory

Wickens' MRT (Wickens 1984; Wickens 2002; Wickens 2008) was created to allow for variations in dual-task performance over separate modalities. The human body's processing system comprises separate processing functions over five dichotomous resource pools: input modality (visual/auditory), processing code (spatial/verbal), visual processing (focal/ambient), processing stage (perception-cognition/response selection) and response type (manual spatial/vocal verbal).

Approaching dual-task modelling in this way presumes a restricted capacity to process, but the theory is that such restrictions only apply to distinct resource pools. Consequently, reductions in performance are surmised to be subject to the degree of sharing between tasks' processing stages/demands. Because all of these dimensions are supposed to contain separate processing resources, specific tasks are thought to be able to be conducted in parallel, as long as they do not use identical processing resources. Stimuli of auditory and visual natures can be simultaneously perceived, for instance, because different resource pools are drawn on. Wickens (2008), in contrast, contends that two apparently different tasks still individually possess processing demands in common, therefore there is unlikely to be good time-sharing. In consideration of this theory's structure, and the type of experimental work therein, which was devised from the position that situations of hearing loss or difficulty in hearing result in cognitive resources employed in comprehending speech being increased, the reasons for simultaneous tasks like retaining good posture being affected cannot be completely identified. MRT holds that increased effort in listening should not affect the retention of good posture, because different resource pools control these tasks' input modalities, therefore performance should not decrease. MRT may therefore not be the optimum model for explaining how people with hearing loss deal with it, and therefore MRT may not adequately explain this research's work.

2.5.1.2 Working memory

Baddeley and Hitch (1974) and Baddeley (Baddeley 1992; Baddeley 2000; Baddeley 2003a) developed an approach to attention and performance limitations that was in effect a synthesis of two other models – multiple-resource models and the central capacity model. This new model was termed the Working Memory (WM) model, and it attempts to explain how the brain temporarily stores and manipulates information. However, the concept of a working memory initially emerged by making a link between perception, long-term memory and action (Baddeley 2003a; Baddeley and Hitch 1974). It can therefore be considered as a way of predicting processing performance on a single or multiple tasks.

Working memory is generally viewed as a limited capacity model. It possesses a hierarchical structure and its purpose is the temporary storage and processing of the information needed to perform complex cognitive tasks (Rönnberg et al. 2013a). One of the main assumptions behind this model is the existence of modality-specific limited capacity pools and a general limited capacity store (Eysenck and Keane 2013). The former are depicted as two, non-interfering slave systems that are coordinated by the general resource pool, which is otherwise called the central executive (Figure 2-2).

There is a phonological loop, which stores speech-based information for just a couple of seconds. However, this information need not be lost if sub- vocal rehearsal is undertaken (Baddeley 1997). There is also a visuospatial sketchpad, which manipulates and processes spatial and visual information. A third slave system, the ‘episodic buffer', also under the control of the central executive, was added to the original model at a later date by Baddeley (2000); this is a non-modality-specific temporary storage system that is believed to incorporate data from the other two slave systems and from long-term memory.

Although the structures of Working Memory and MRT are similar, Working Memory differs as it incorporates the central executive. Only recently has this aspect of working Memory become better understood, as initially Baddeley et al. (1986) described it as a ‘ragbag’, whose role was to process any poorly understood or complex phenomena. As a result of further research, it is now believed that the central executive is responsible for focusing, dividing or switching attention (Baddeley 2003a). Therefore, while there is no direct interference or communication between the phonological loop and the visuospatial sketchpad, these two systems are coordinated by the central executive. The efficient functioning of these two sub-systems therefore depends on the central executive’s resources being available, and not being diverted elsewhere or otherwise engaged.

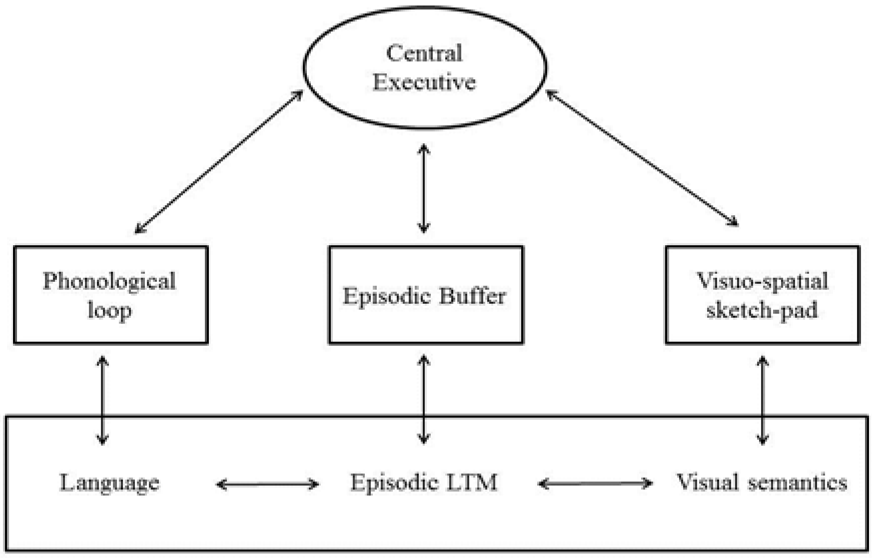


Figure 2-3 Baddeley's working memory model. Source: Working Memory: Theories, Models, and Controversies (Baddeley 2012).

Since the switching of attention between the two sub-systems is controlled by the central executive, it is reasonable to assume that a decrease in performance will be seen if both sub-systems require input from the central executive, or if the central executive is subject to a high load. This explains dual-task interference in working memory when the two slave systems are carrying out simultaneous tasks. In this research study, performance limitations can be accounted for by working memory. For example, when there is a hearing impairment present, or when the listener is in a challenging listening situation, the phonological loop will require more input from the central executive to process the auditory information. This will reduce the attentional control of the visuospatial sketchpad and episodic buffer, meaning that the individual is less able to perform tasks that require input from the central executive. Consequently, it is likely that we will see decreased performance in tasks that utilise these constructs, for example maintaining an upright posture. Working memory is therefore a suitable model for employment in this thesis.

2.5.2 Hearing loss and cognition and understanding speech

2.5.2.1 Impact of hearing loss on cognitive processing

A loss of hearing does not simply result in a reduction in audibility; it also distorts sounds that are above the hearing threshold. Understanding speech is a complex task involving multiple stages, and it utilises several different cognitive processes (Fritz et al. 2007), of which perception is only one. Craik (2007) emphasises that "attention, perception, comprehension, memory and thinking are all aspects of the same cognitive system and that, as such, deficiencies in one aspect will have consequences for other aspects”. Craik goes on to describe the complex processes involved in listening, but despite this, for adults with normal hearing, understanding speech is usually effortless and automatic. This is not so for many adults with an SNHL however (Pichora-Fuller, Schneider and Daneman 1995). Increasing sound intensity does not rectify the problem, because efficient listening processes are not restored by simply increasing the volume, due to the perceptual consequences associated with SNHL (Killion et al. 2004).

However, although SNHL distorts some parts of an auditory signal so that they cannot be recognised or correctly perceived, individuals can make up for this shortfall with their linguistic knowledge, to the extent that a spoken message can be reconstructed. For example, in a study conducted by Warren (1970), subjects were asked to listen to recordings of sentences where a cough replaced single phonemes. With a typed version of the sentences in front of them, they listened and then circled where they had heard the cough, and indicated whether it had completely replaced this circled section. Out of twenty subjects, nineteen reported the sentences as being complete, and the other one incorrectly indicated where the cough had occurred. They had all therefore ‘perceptually completed’ the missing part of the sentence, a phenomenon that was called ‘phonemic restoration’ by the authors. Further research has shown that this perceptual filling also occurs for non-speech sounds, when it is then called ‘auditory induction’ (Warren 2013).

As even older adults still maintain good linguistic knowledge (Wingfield, Tun and McCoy 2005) they may be able to use phonemic restoration to adapt to living with SNHL. However, researchers have reported that individuals with normal hearing are better at performing phonemic restoration than those with severe hearing loss (Başkent, Eiler and Edwards 2010). Moreover, cognitive resources are also required when attempting to inform discourse by retrieving and employing linguistic knowledge (Baddeley 2003b), and several studies have confirmed that individuals with hearing loss require more cognitive resources to comprehend an auditory message, compared to normal hearing individuals (Pichora-Fuller, Schneider and Daneman 1995; Edwards 2007; Stenfelt and Rönnberg 2009). When attempting to listen in conditions that are less than ideal, it comes as no surprise therefore that cognitive skills have a bearing on the understanding of speech. Akeroyd (2008) conducted a review of twenty papers published in the last thirty years that examined correlations between understanding speech in a noisy environment and various aspects of cognitive function. He came to the conclusion that there was an association between the two, secondary to the predictive power of hearing loss.

The link between cognition and hearing loss was explained by Stenfelt and Rönnberg (2009). They proposed that in optimal conditions, listening is implicit and suffers no delay, but when the auditory signal is distorted by hearing loss or by artificial means, then the listener is forced to employ an additional top-down strategy, explicitly decoding phonological content. This necessarily increases the demand on cognitive resources. Top-down processing means that the individual chooses to allocate their attention to the task in hand (Klingberg 2010), which with reference to the working memory model involves engaging the central executive. An individual with SNHL would therefore use a top-down listening strategy (and allocate increased resources to sound processing) more frequently than a normal hearing person would (Hornsby 2013).

The Ease of Language Understanding Model (ELU) by Stenfelt and Rönnberg (2009) uses working memory to explain that cognitive resource requirements are increased for successful listening (Ronnberg 2003; Rönnberg et al. 2013a; Rönnberg et al. 2008). In this model, language understanding is viewed as a system that is part of the working memory. The model describes and predicts the dynamic interactions between explicit and implicit cognitive functions, particularly when dealing with poorly perceived acoustic signals.

2.5.2.2 The Ease of Language Understanding model

The research that underlies the ELU spans the past thirty years. Rönnberg et al. (2008) propose a strong link between the understanding of language and working memory, therefore working memory processes are at the heart of the model. The basis of this proposition is work undertaken to demonstrate that working memory capacity is a strong predictor of speech understanding (Lyxell and Rönnberg 1989; Lunner 2003). Rönnberg et al. (2013b) suggest that people with hearing loss have to contend with several processes during a conversation – they must coordinate long-term memory, contextual cues and the perceptual input, which in their case is distorted. In their model, the skills utilised in complex working memory tasks are key to how well these compensatory interactions are managed in people with hearing loss.

Pichora-Fuller (1996) also propose a link between the comprehension of language and working memory. This comprehension requires the individual to perceive and recognise words, but the information acquired must also be connected with information kept in the long-term memory for correct interpretation, as well as with more recently stored information that was received earlier in the discourse. The author highlights the fact that when trying to listen in sub-optimal conditions, when listening requires more effort, mental resources give priority to word recognition, meaning that fewer resources can be utilised for storing or processing the information in greater depth. This conclusion was drawn from the observation that word recall diminishes when stimuli are presented at low signal-to-noise ratios (Rabbitt 1968; Pichora-Fuller, Schneider and Daneman 1995).

The ELU draws and builds upon this work by Pichora-Fuller (1996), as it explains how and under which situations working memory is employed to support listening in less than optimal conditions. Rönnberg et al. (2008) referred to a process known as ‘RAMBPHO’ whereby multimodal speech information is Rapidly, Automatically, and Multimodally Bound into a PHOnological representation in an episodic buffer. They make the assumption that the listener then searches for a match between this phonological representation and others lodged in the semantic long-term memory.

This seems to be based on studies such as those conducted by Poeppel, Idsardi and Van Wassenhove (2007) and Bendixen, Schröger and Winkler (2009), which reported that the auditory system has ‘predictability’, i.e. it can use contextual factors to predict which sound will come next. If RAMBPHO corresponds to the phonological representations being heard, then top-down processing is not needed, as lexical access and speech understanding can succeed without it. In this situation, lexical retrieval occurs quickly and implicitly, which is why individuals with normal hearing can understand speech effortlessly in favourable listening environments.

However, if the information from RAMBPHO is unclear and the listener is unable to immediately or unequivocally match it with phonological representations stored in the semantic long-term memory, top-down processing strategies are engaged. This supposition finds support in other research, for example a study by Foo et al. (2007). The participants in the study were habitual hearing aid users, and the compression settings of their hearing aids were altering, forcing a mismatch between RAMBPHO and phonological-lexical representation. The authors concluded that a person’s ability to overcome such mismatches depended on their working memory abilities, with the inference being that listeners employed a more explicit method of processing when this mismatch was encountered.

Rönnberg et al. (2008) highlighted the fact that under favourable listening conditions the ELU functions separately to the Working Memory model, but in less than optimal listening conditions, a more explicit approach to language understanding is engaged, and this bears a close resemblance to the Working Memory model. This is seen as being an example of the efficient use of resources, using implicit processing as far as possible for language processing, and only engaging processes that require more effort when the listening conditions are less than optimal.

Therefore, following the ELU, it is reasonable to expect suboptimal listening conditions to have a significant impact on the performance of concurrent working memory processes, as under these conditions, cognitive resources are redirected to assist with the comprehension of speech. The ELU proposes that several cognitive processes are engaged when a RAMBPHO mismatch is encountered (inference-making, semantic integration, switching of attention, storing of information, and inhibiting irrelevant information); Rönnberg et al. (2013a) therefore envisage that the central executive plays an important role in language comprehension. Following the original model's premise that explicit processes are drawn from a general capacity store (the central executive), if working memory processes are called upon to assist with comprehending speech, fewer resources will be available to other tasks being performed at the same time, leading to a decrease in performance.

Of course, the conditions in which speech occurs is constantly fluctuating, in terms of the acoustic environment, the volume of speech and ambient noise. Therefore, implicit and explicit processes are likely to be called upon in different proportions throughout a conversation. However, for those with hearing loss, the acoustic input will be of even poorer quality and such individuals are likely to experience mismatches between RAMBPHO and long-term memory stores much more frequently. It is therefore a reasonable conclusion that people with hearing loss will have to employ explicit, effortful processing more often than normal hearing individuals in the course of a conversation.

Empirical studies by other people support the theory that hearing loss may distort auditory signals sufficiently to impact on working memory processes (Rabbitt 1991; McCoy et al. 2005). Rabbitt (1991) revisited an earlier study (Rabbitt 1968) with the hypothesis if he degraded an auditory stimulus with a masking noise, the effect on participants with hearing loss would still be seen if the masking noise was removed. His hypothesis was supported and he drew the conclusion that “hearing loss will cause recognition errors, but may also impose an additional load on information processing capacity which prevents individuals from optimally rehearsing even those words that they have correctly heard”.

Therefore, the ELU can explain how a degraded auditory signal can impose additional demands on working memory processes. The model has recently been updated (Rönnberg et al. 2013b), with the new model including an explicit processing loop that provides feedback information to the early stages of speech comprehension, informing RAMBPHO information until the listener obtains some degree of understanding. This understanding goes on to induce a semantic framing of the next processing loop, providing the listener with some context, which helps them to gain an understanding of what is being said. This new model upholds the original premise that a match is sought between RAMBPHO information and lexical information stored in long-term memory, and when no direct match is found, explicit processes are engaged. A degraded auditory signal therefore places an additional load on working memory processes, which explains why hearing loss may require an individual to exert extraneous effects for task performance when required to dual task.

2.5.3 Posture and cognition

Established thinking on the neural processing relating to posture control was that it occurred automatically within the reflex circuits relating to the spine and brainstem, but the results of later research using a dual-task postural paradigm indicate that aspects of posture are influenced by cerebral cortex and cognitive processing. The process of controlling stability appears to require attentional resources. Research also indicates that the control of posture is governed by multiple sensory systems and that postural task requirements change the level of influence of the different sensory systems as people rapidly and continuously select the most relevant and appropriate combination of sensory feedback for a given context.(Horak, Nashner and Nutt 1988; Horak and Nashner 1986; Macpherson and Fung 1999; Macpherson and Inglis 1992; Stapley et al. 2002).

For example, when someone stands on the deck of a boat, their body increases weighting on the incoming flow of visual information and decreases weighting on signals from foot and ankle somatosensory receptors, as they are less useful in this context, in order to establish and maintain vertical orientation. However, when standing on a bus or a train, where forward and backward movement of the support surface is experienced the CNS relays more on signals from foot and ankle somatosensory receptors as they are more useful in this situation. This ability to re-order sensory inputs under changing incoming sensory conditions can be defined as CNS adaptability to events, or a type of redundancy that extends and increases postural control efficiency (Nashner 1982).

When more than one attentionally demanding task is being performed, this can lead to competition for processing resources and thereby lead to task interference (Bourke 1996). In posture cognition dual tasking literature, the primary task is considered to be postural and the secondary task is cognitive. Reduced performance of either or both of these tasks has been ascribed to interference or limited attentional resources shared between them (Woollacott and Shumway-Cook 2002). Failure to allot attention to posture in dual task conditions, or the reduced amount of available attentional resources, are postulated to be causes of instability. The following subsection provides a brief overview of existing literature concerning the relation between posture and cognition and illustrates how researchers interpreted their diverse results.

2.5.3.1 Posture cognition dual tasking

Several studies have looked at the relationship between posture control and cognition by conducting experiments where individuals are asked to perform a cognitive task while at the same time carrying out a postural task – hence the term dual-task. This section presents an overview of the research carried out on this relationship in young, older and balance impaired adults.

Postural control dual-task studies show that when stance is *perturbed*, performance of the cognitive task is affected (Andersson et al. 2002; Andersson, Yardley and Luxon 1998; Brauer, Woollacott and Shumway-Cook 2001; Brown, Shumway-Cook and Woollacott 1999; Rapp, Krampe and Baltes 2006; Redfern et al. 2001; Redfern et al. 2004; Shumway-Cook and Woollacott 2000; Teasdale et al. 1993; Yardley et al. 2001). The decrement in the cognitive task performance whilst dual tasking is marked in older (Brauer, Woollacott and Shumway-Cook 2001; Brown, Shumway-Cook and Woollacott 1999; Rapp, Krampe and Baltes 2006; Shumway-Cook and Woollacott 2000; Teasdale et al. 1993) and balance impaired adults (Brauer, Woollacott and Shumway-Cook 2001; Rapp, Krampe and Baltes 2006; Redfern et al. 2004; Shumway-Cook and Woollacott 2000) when compared to young adults performance.

Manipulations of the secondary task when stance is *perturbed* are reported to increase postural sway (Andersson, Yardley and Luxon 1998; Brauer, Woollacott and Shumway-Cook 2001; Marsh and Geel 2000; McNevin and Wulf 2002; Pellecchia 2003; Pellecchia and Shockley 2005; Rapp, Krampe and Baltes 2006; Redfern et al. 2001; Redfern et al. 2004; Shumway-Cook and Woollacott 2000; Shumway-Cook et al. 1997). Manipulations of the secondary task were found to have a greater effect in older adults and patients (Brauer, Woollacott and Shumway-Cook 2001; Rapp, Krampe and Baltes 2006; Redfern et al. 2001; Shumway-Cook and Woollacott 2000; Shumway-Cook et al. 1997), however, a number of experiments conducted on healthy young adults revealed reduced sway (Andersson et al. 2002; Brauer, Woollacott and Shumway-Cook 2001; Dault et al. 2001; Dault, Yardley and Frank 2003; Riley, Baker and Schmit 2003; Riley et al. 2005).

In the study conducted by Shumway-Cook et al. (1997), for example, normal and altered foam support surfaces were used as postural conditions, and the secondary task involved the use of sentence completion for language processing and the judging of line orientation for visual spatial processing. They recruited young, older and balance impaired adults. No significant differences were noted between the young and older adults, however, the balance impaired group showed significant difference on both postural conditions.

During *unperturbed* stance, decrements in cognitive task performance were also reported when the postural task is made harder by for example reducing the base of support (Kerr, Condon and McDonald 1985; Mitra 2003). For example, Kerr, Condon and McDonald (1985) paired a spatial and non-spatial memory tasks with tandem Romberg stance. An increase in errors was found in the spatial memory task but not the non-spatial one. Hence the study reported that postural control was attentionally demanding, and the authors suggested that there is a sharing of resources between spatial memory and postural tasks, while resources for non-spatial cognitive tasks are accessed separately.

Secondary task manipulations during *unperturbed* stance have revealed inconsistent results, showing increased postural sway (Huxhold et al. 2006; Maki and Mcllroy 1996; Manckoundia et al. 2006; Marsh and Geel 2000; Melzer, Benjuya and Kaplanski 2001; Mitra 2003; Mitra and Fraizer 2004; Rapp, Krampe and Baltes 2006; Shumway-Cook et al. 1997), decreased postural sway (Siu and Woollacott 2007; Vuillerme, Nougier and Teasdale 2000; Weeks et al. 2003; Dault, Frank and Allard 2001; Dault et al. 2001; Dault, Yardley and Frank 2003; Deviterne et al. 2005; Hunter and Hoffman 2001; Riley et al. 2005), or unchanged postural sway (Teasdale et al. 1993; VanderVelde, Woollacott and Shumway-Cook 2005; Yardley et al. 2001).

For example, in the study conducted by Huxhold et al. (2006) they suggested that a low-demand cognitive task would steer young or older adults' attention away from automatic postural control processes, and thereby enable postural control to work more efficiently. They favoured grading the difficulty increments of the secondary task so as to affect attentional resource competition. To test this, four cognitive tasks were set under both sitting and standing conditions: 1) Watching a random series of digits between one and nine being presented on a computer screen; 2) Choice (digit) reaction time; 3) n-back [2] digit WM; and 4) [2]-back spatial WM. The postural results were presented for the standing mode, showing a continuum of difficulty, postural control deteriorated significantly, from low-high for cognitive tasks (1-4).

The decrement in postural dual task performance during *unperturbed* stance is also marked in older adults (Dault and Frank 2004; Huxhold et al. 2006; Maylor and Wing 1996; Rapp, Krampe and Baltes 2006) and patients (Hauer et al. 2003; Manckoundia et al. 2006; Marchese, Bove and Abbruzzese 2003; Rapp, Krampe and Baltes 2006) when compared to young adults performance.

To conclude, cognitive task performance is reported to be affected when individuals are trying to maintain erect stance during challenging balance conditions. Although manipulation of the cognitive task during postural dual tasking has been reported to affect postural performance, it is less-well understood in what ways it happens. Posture performance whilst dual tasking in older and balance-impaired adults seems to be hindered by concurrent cognitive tasks. Moreover, manipulations of the secondary task revealed inconsistent results in young adults. The posture-cognition dual tasking literature revealed diverse results, even in young adults and controls, and did not provide adequate answers as of how the concurrent cognitive task impact posture performance. Investigations should discover cognitive tasks that affect posture and understand how they influence it. It is this challenge that this research aiming to address by identifying if listening under different testing conditions has an effect on posture control and how.

2.6 Aims and Objectives

Information regarding the current understanding of the posture cognition relation and how may postural cognition dual tasking interference happens has been provided, however, many questions remain unanswered as to the precise nature and strength of the cognitive tasks that might interfere with postural control. The aim of this line of research is to carry out a number of linked investigations to question the impact of such a cognitive task. The cognitive task in investigation in this program of studies is listening and listening in the presence of hearing impairment. It is apparent that, thus far, there has been no active consideration of the relationship between listening and hearing impairment, and postural control in adults.

Investigations carried out in this thesis should discover if listening and listening in the presence of hearing impairment affect posture and understand how they influence it. It is this challenge that this research aiming to address by identifying if listening under different testing conditions has an effect on posture control and how.Stated below are the aims and objectives of this thesis:

Aim one: Explore the relationship between listening and posture control in healthy adults with normal hearing. The objectives are to investigate if listening has an influence on posture, and if so, to find which tools, i.e. body sway measures, are most appropriate to detect this effect.

Aim two: Explore the upright posture mechanisms of healthy adults with normal hearing, and how they adapt while actively listening, with the objective to observe any changes in the underpinning strategies to maintaining erect posture while dual tasking.

Aim three: Examine the influence of simulated sensorineural hearing loss on posture control in healthy adults, with the objective to explore if the presence of a simulated hearing loss will 1) reveal an exacerbated influence of listening on posture control and 2) if strategies and mechanisms to maintain erect posture while listening are different.

A series of linked experiments were designed to address these aims and objectives. In the first experiment, the influence of an auditory listening task on posture control will be investigated in the context of normal-hearing adults. Exploring the relationship between listening and postural control in normal individuals will help to address the second and the third aims of this thesis.

3 Chapter Three: Exploring the Influence of Listening on Posture in Healthy Adults with Normal Hearing

3.1 Introduction

The aim of this chapter is to explore the effect of listening on posture control using a dual-task paradigm. Dual-task paradigm requires an individual to perform two tasks at the same time. It is posited that the processes involved in attention are constituted by a single cognitive resource, and when multiple tasks are carried out simultaneously, each task competes for these resources (Kahneman 1973).

The findings from this chapter will provide important data to characterise the attentional interaction between listening and posture. If the anticipated attentional effect of listening on posture control is proved evident, this will raise further questions, such as what are the underpinning strategies and mechanisms behind this effect, and does hearing impairment exacerbate it? The following paragraphs of this introduction will briefly discuss posture control and postural control measurement, followed by the methods employed to conduct this experiment, results and discussion.

3.1.1 Posture

The body's positioning is regulated by the posture control system, which maintains balance and orientation. The system is based on an internal representation of the body's position in space, along with the central integration of visual, tactile, vestibular and proprioceptive information (Amblard et al. 1985; Black, Wall and Nashner 1983; Winter 1995b; Jiang and Chun 2001). The feedback received from this information is used by the body to continually update its position, and this internal representation is then used by the motor system to control the body's position in space (Massion 1994; Mergner and Rosemeier 1998).

The subcortical nervous structures regulate the everyday motor-balance skill of quiet standing (Lacour and Borel 1993). Despite this being a simple task in postural terms, quiet standing uses considerable cognitive resources (Lajoie et al. 1993). It has been established that, in younger adults, an undisturbed upright stance requires a low level of attentional resources. However, when faced with more challenging conditions such as balancing on one leg or walking in difficult conditions, postural tasks take more cognitive effort. When these tasks are difficult or complicated, or when the individual's balancing ability is decreased due to ageing, there is an increase in cortical structures contribution – primarily those involved in motor attention (Rushworth et al. 2003) as well as those associated with the internal representation of the body (Michel et al. 2003; Pérennou et al. 2001; Thier and Karnath 1997).

Everyday postural behaviours are usually accompanied by other cognitive processes. The majority of postural tasks, whether static or dynamic, are performed by individuals at the same time as cognitive tasks. An example of this type of behaviour would be running while listening to music, or chatting while taking a walk. In order for both asks to be performed effectively, attentional resources need to be divided; for this reason, it is worth questioning whether postural performance level is affected by these dual-task conditions.

It is possible to answer this question by comparing the results from a single postural task performed on its own with those from a postural and cognitive tasks performed simultaneously. Research suggests that in dual tasks, there is competition for central processing resources, which results in lower performance levels (Li, Krampe and Bondar 2005; Bowen et al. 2001). However, it has been clearly demonstrated that there are many factors that can significantly affect the level of performance during a dual-task paradigm such as the nature of each task and their environmental context (Woollacott and Shumway-Cook 2002). Further, more intrinsic factors have been reported as having an effect on dual-task performance, such as an individual’s sensorimotor expertise (Vuillerme and Nougier 2004).

3.1.2 Assessment of postural stability

One of the methods to assess postural stability is called posturography (Browne and O'Hare 2001). This method performs and provides a quantitative assessment of postural stability (Baratto et al. 2002). It records an individual’s centre of pressure (COP) trajectory through the use of a force platform. Force platforms can accurately assess balance and monitor an individual’s progress by detecting minor changes in stability (Pellecchia and Shockley 2005).

Postural stability in both dynamic and static conditions can be measured using force plates (Pivnickova, Dolinay and Vasek 2012), however, static posturography is the focus in this instance. The quantitative nature of static posturography is a useful way to measure postural stability (Browne and O'Hare 2001). This technique demonstrates that although an unperturbed upright stance appears to be static, there is in fact a gentle rocking action from the ankle, which has been compared to a simple inverted pendulum (Gage et al. 2004). The oscillation and slight swaying is measurable using the COP (Pellecchia and Shockley 2005).

Despite many quantitative assessments of postural stability being obtained from COP, it is still unclear which of these measurements are the most effective (Moghadam et al. 2011; Shepard and Janky 2010; Norris et al. 2005; Salavati et al. 2009). Moreover, there is no standard study design procedure set in place which results in difficulty comparing the design of experiments due to their variety and diversity (Ruhe, Fejer and Walker 2010; Ruhe, Fejer and Walker 2011). In addition, it is also difficult to interpret dual-task data, considering the evidence that both sensorimotor (Horak and Nashner 1986; Jette, Branch and Berlin 1990; Manchester et al. 1989) and cognitive functions gradually decrease with age (Craik and Salthouse 1992; Maylor, Allison and Wing 2001).

3.1.3 Aim and objectives

To address the primary question dealt with by this thesis, i.e. to what extent listening influences posture, a first stage would be exploring the relation mediated by attention between listening and postural control in healthy individuals with normal hearing. The experiment described in this chapter employed a dual-task paradigm in order to study this relation. Body sway whilst listening was recorded, with the aim of investigating whether listening has an effect on posture control, and to identify body sway measures capable of detecting this effect. It was hypothesised that:

1. Body sway will increase in dual-task conditions compared to their baseline conditions.
2. Body sway will increase in line with increased posture task difficulty.
3. Errors in the listening task will increase in line with increased posture task difficulty.
4. Overall performance during dual-task conditions will deteriorate.

To determine the cost of performing a listening task while maintaining upright posture, body sway was recorded from healthy individuals with normal hearing under single and dual-task conditions and different body sway measures were calculated.

3.2 Methods

3.2.1 Ethical approval

An ethical approval from the School of Healthcare Research Ethics Committee (SHREC) was obtained for this study (SHREC/RP/296) (see Appendix One).

3.2.2 Inclusion and exclusion criteria

Inclusion criteria were: age between 18 and 60 years; normal middle ear function (ear pressure from -500 to +500 pa, compliance between 0.3 to 1.6 cm^3 and ear canal volume between 0.6 to 1.5 cm^3); normal hearing bilaterally (hearing thresholds better than or equal to 20 dB HL across octave frequencies between 250 and 8 kHz); no personal history of balance problems; and English as the first language. Any participant who did not meet the inclusion criteria, reported neurological disorders, motor problems, severe visual impairments or unable to understand and follow verbal instructions, was excluded from the study. Participants were advised to visit their general practitioners (GPs) if abnormal hearing thresholds were noted.

3.2.3 participants

Sample size was determined using Cohen’s power tables (Cohen 1988). For alpha = 0.05 and r = 0.50 at 85% power, Thirty-two participants are required. Thirty-three normal healthy participants, 21 males and 12 females, aged between 19 and 59 years old (mean age 39.7 SD± 11.1) met the include criteria and were included in the study. The participants were mainly students and staff recruited from the University community, but participants from around Leeds city were also recruited. The study was advertised using posters (see Appendix Two) inside the University campus and around Leeds city, by circulating emails to university staff and students, and by the snowball strategy. Written informed consent (see Appendix Three) was obtained from all participants after oral information, printed information (see Appendix Four) and the opportunity to ask questions had been given. Participants were given vouchers to recompense them for their time.

3.2.4 Equipment

3.2.4.1 Screening equipment

A calibrated Interacoustics MT10 middle ear analyser was used to perform tympanometry testing using a 226 Hz probe tone to establish middle ear function and a calibrated FONIX Hearing evaluator, FA-12 Digital Audiometer, was used to conduct audiometry to evaluate the hearing thresholds.

3.2.4.2 Experimental equipment

For balance evaluation, COP data were collected using a Kistler force plate type 9286BA (600x400x35mm) and an amplifier or a data Acquisition Box (Kistler DAQ Type 5691A) controlled by a SAMSUNG laptop running proprietary software (BioWare v5.1.1.0) to control data collection. This Kistler force plate (Figure 3-1) contains four piezoelectric sensors, one at each corner, which have three pairs of quartz plates. These become electrically charged when the plate is stepped on. These electrical charges are collected and then converted to analogue voltages by the amplifier. Using the laptop and the analysis software, BioWare, these data are controlled and processed to give visual and statistical data of ground reaction forces, moments and most importantly, COP.

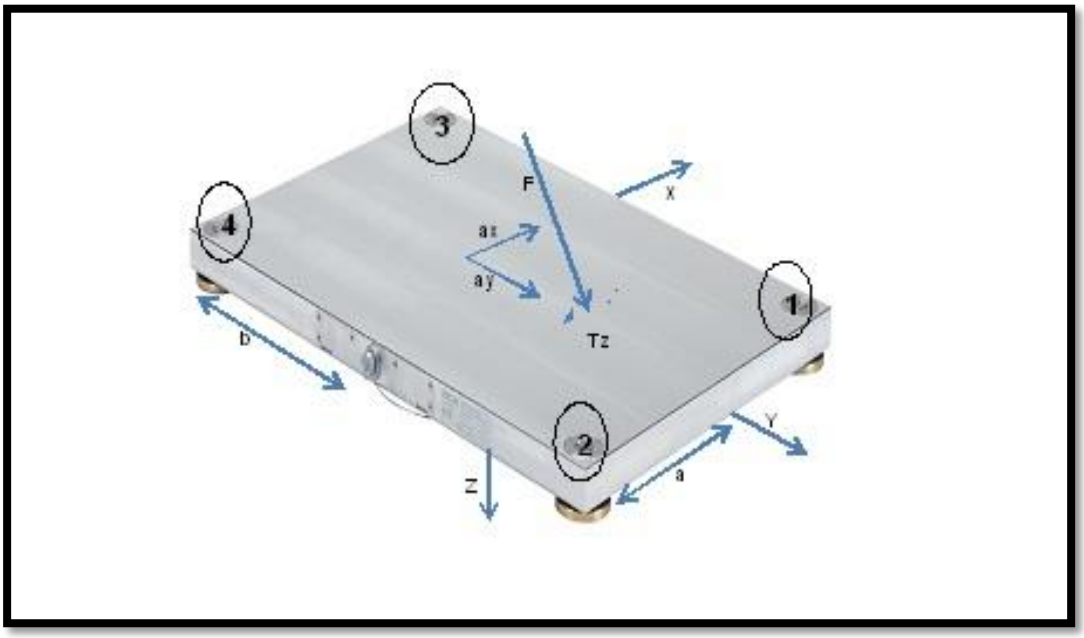


Figure 3-1 Force plate. Piezoelectric sensors 1, 2, 3 and 4. ML direction (x). AP direction (Y). Force (F). ax and ay: x and y coordinates of force application point. Tz free moment. Z vertical force.

3.2.4.2.1 Sway measures

Nine sway measure were calculated from the COP data in order to observe the hypothesised effect of listening on posture control. These measures were Mean Sway Amplitude (MSA) in both anterior-posterior (AP) and Mediolateral (ML) directions, Standard Deviation of Sway Amplitude (SDSA) in AP and ML directions, Movement Range (MR) in AP and ML directions, Standard Deviation of Velocity (SDV) in AP and ML directions, Phase Plane Portrait (PPP) in AP and ML directions, Mean Velocity (MV), Planar Deviation (PD), Sway Area (SA) and Total Phase Plane Portrait (TPPP). A code was written using R software to facilitate the calculations of these sway measures from the raw COP data (see Appendix Five).

PPP is a velocity versus displacement plot, it is evaluated based on whole body centre of gravity and centre of pressure and can be quantified from the root mean square variance of velocity and displacement. PPP value is defined by the ratio of amplitude to velocity by means of a predominant reference measurement, which is the time unit. PPP can be calculated in both AP and ML directions and also as a total, TPPP (Riley et al. 1995). PD is basically the square root of the sum of standard deviation in the mediolateral direction squared and the standard deviation in the anterior-posterior direction squared and is measured in (mm). Sway area can be described as the area spanned from the COP and it is measured in (mm^2/s) (Rocchi, Chiari and Cappello 2004). Formulas to calculate these measures are presented in table 3-1.

Table 3-1 Formulas to calculate global sway measures

Measure	Direction/description	Formula
SD of amplitude	AP	$\sigma_x = \sqrt{\frac{\sum(x_i - \bar{x})^2}{N-1}}$
	ML	$\sigma_y = \sqrt{\frac{\sum(y_i - \bar{y})^2}{N-1}}$
SD of velocity	AP	$\sigma_{v_x} = \sqrt{\frac{\sum(v_{x_i} - \bar{v})^2}{N-1}}$
	ML	$\sigma_{v_y} = \sqrt{\frac{\sum(v_{y_i} - \bar{v})^2}{N-1}}$
	Where:	$v_{x_i} = \frac{x_{i+1} - x_i}{t_{i+1} - t_i}$ $v_{y_i} = \frac{y_{i+1} - y_i}{t_{i+1} - t_i}$
PPP	AP	$\sigma_{r_x} = \sqrt{\sigma_x^2 + \sigma_{v_x}^2}$
	ML	$\sigma_{r_y} = \sqrt{\sigma_y^2 + \sigma_{v_y}^2}$
	Total PPP	$\sigma_r = \sqrt{\sigma_{r_x}^2 + \sigma_{r_y}^2}$
Mean Velocity		$\bar{v} = \frac{1}{T}\sum_1^T \sqrt{(\chi_{t+1} - \chi_t)^2 + (y_{t+1} - y_t)^2}$
Area		$A = 2\pi F_{0.05[2,N-2]}\sqrt{\sigma_x^2\sigma_y^2 - \sigma_{xy}^2}$
	Where:	$\sigma_{xy}^2 = \sum \frac{(x_{i-}\bar{x})(y_{i-}\bar{y})}{N}$

In the literature, many different body sway measures have been used to study postural stability; however, due to discrepancies in the literature it was difficult to decide which measures to include for the purpose of this study (Bauer et al. 2008; Demura, Kitabayashi and Aoki 2008; Doyle, Newton and Burnett 2005; Mancini et al. 2012; Raymakers, Samson and Verhaar 2005). All of the measures employed in this experiment have been shown useful in studying postural stability in the existing literature (Doyle et al. 2008; Jeka et al. 2004; Moghadam et al. 2011; Gray, Ivanova and Garland 2014; Santos et al. 2008; Karlsson and Frykberg 2000).

3.2.4.3 Listening stimuli

The listening task used in this experiment was a competing words test-free recall. This is a sub-test of the SCAN-3, which was originally designed to diagnose and describe central auditory processing disorders (Keith 1995). This sub-test comprised 20 test trials; in every trial, the participant hears two mono-syllabic words presented to opposing ears at the same time, for example, ‘soft’ and ‘last’ (see Appendix Six). This test was chosen as a listening task for the purpose of this study because it is a dichotic test requiring a participant response and that it targets binaural integration and involves brainstem, cortical and corpus callosum, and hence it is thought to tax cognitive functions (Katz et al. 2009).

A Sony CD player was connected to a calibrated Madsen Itera audiometer in order to control the presentation level of the auditory task and to deliver the test materials to the participants via TDH-39 headphones. The listening task was presented at sensation level, 30 - 40 dB above the participant’s average thresholds, which ensured stimuli were played at audible levels for all participants. Participant were required to listen to the words and to repeat back what they heard during the silent gaps between presentations. It was emphasized that if participants were uncertain about a word or only heard part of it, they should say what they thought they had heard. Only correct word recalls were recorded as a successful response from the participants.

3.2.5 Procedure

The tests were carried out in the audiology laboratory at the School of Healthcare at the University of Leeds. Every participant underwent a medical history screen (see Appendix Seven) followed by a routine clinical audiological assessment. The routine assessment involved visual inspection of the ears according to the British Society of Audiology's recommended procedure (British Society of Audiology 2010) to ensure that both ear canals and the tympanic membranes were normal, and to establish that there were no factors that might prevent testing, tympanometry testing according to the British Society of Audiology's recommended procedure (British Society of Audiology 2012) was carried out to check that the middle ear was functioning normally, and audiometry testing according to the British Society of Audiology's recommended procedure (British Society of Audiology 2011) was employed to ensure that the participant had hearing threshold levels of 20 dB or less at the following frequencies: 250, 500, 1000, 2000, 4000, 8000 Hz (see Appendix Eight for an Audiogram). Hearing assessments (audiometry) were always performed in a sound booth and the rest of the screening and experimental assessments were conducted in sound-treated rooms.

After screening, participants who met the inclusion criteria were tested on the dual-task experiment. The participants were asked to stand on the force plate adopting 1) normal stance, standing with their feet normal apart i.e. their most comfortable normal standing position and 2) Tandem stance, were participants stand in a heel-to-toe position (see figure 3-3), with eyes open and closed with and without the listening task, the SCAN-3 competing words sub-test. When there was a listening task, participants were required to respond by verbally repeating what they had heard. There were, therefore, eight conditions: Normal Standing with eyes Open (NSO), Normal Standing with eyes Open with Auditory task (NSOA), Normal Standing with eyes Closed (NSC), Normal Standing with eyes Closed with Auditory task (NSCA), Romberg Stance with eyes Open (RSO), Romberg Stance with eyes Open with Auditory task (RSOA), Romberg Stance with eyes Closed (RSC) and Romberg Stance with eyes Closed with Auditory task (RSCA). Every condition was recorded three times, each lasting 30 seconds. Participants also performed a baseline for the auditory task. For the baseline auditory task, although no balance measurement was recorded, the participants stood on the force plate. This is very important as all of the testing conditions needed to be exactly the same, with the exception of the absence or presence of the auditory task. Centre of pressure data was collected at a sampling rate of 100 Hz with a high pass filter of 10 Hz. Testing conditions were counterbalanced to control for order effects.

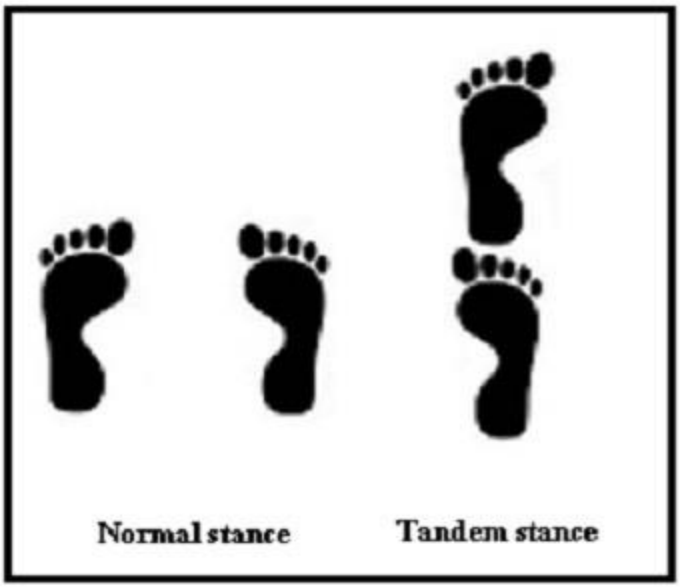


Figure 3-2 Normal and Tandem stance employed in this research

The idea beyond including different standing positions, NS and RS, and visual conditions, eyes open and closed, was 1) to introduce varying levels of difficulty of posture control to check for the effect that listening might have on posture which might not be detected under easy postural tasks especially that participants recruited in this program of research are healthy adults and 2) to help achieve one of the goals of this research which wanted to investigate and interpret this effect from an attention perspective. Including increasing levels of postural difficulty is assumed to demand more cognitive processing which as a result leave the listening task with less processing capacity. Changing the level of difficulty of the posture tasks through narrowing the base of support and/or removing visual information should help identify if the performance on the listening task remains constant or insignificantly changed or cause a performance deterioration. As this line of research was designed as a linked series of experiments, were one informs the next, it was thought that maintaining the same conditions in all experiments is important to facilitate a better understanding of the pattern, if any, that this effect of listening on posture control might has.

For recording these conditions there are factors to be considered. The length of the recording is an important factor as too short a time may result in an incorrect evaluation due to insufficient data, and too long a time recording may result in participant fatigue. Corriveau et al. (2000) and Lafond et al. (2004) recommended 1-2 minutes recording time in quiet standing, however, Le Clair and Riach (1996) suggest a recording time of 30 seconds when assessing adults. The latter was adopted in the current study as 1-2 minutes was considered likely to bias results through excessive fatigue. The recommended number of measurements of a single testing condition varied from two (Lafond et al. 2004) to four readings (Corriveau et al. 2000). Several readings may cause a learning effect, which will reduce the body sway, or may otherwise result in participant fatigue, which subsequently increases the body sway (Duarte and Freitas 2010).

Posture control for all participants was evaluated using the same equipment in the same laboratory settings (Figure 3-2). Participants were asked to stand on the force plate barefoot (shoes removed but socks on), as still as possible, with their arms at their sides and looking straight ahead, fixating on a 10 cm diameter black dot on the wall approximately 150 cm in front of them at eye level. In conditions where eyes were closed, the participants were asked to imagine looking at the target. The use of a fixation eye target has been found to affect postural stability. Positioning the target too high or too low would change the head position and therefore body balance (Paulus, Straube and Brandt 1984b; Prado, Stoffregen and Duarte 2007). Duarte and Freitas (2010) recommend that the fixation point is positioned at a height according to the participant's eye line. This method was adopted in the current study.

In dual-task conditions, participants were required to perform the posture task simultaneously with a listening task. The participants were asked to repeat back the words they had heard and they were informed that upon the initiation of the test, it would not be paused or interrupted. They were told that the stimuli were meaningful words and they were encouraged to repeat back the words they had heard in both ears in any order even if it was a part of a word or if it did not make sense. The participants responded during the silent intervals between the words. Strict scoring was used, i.e. correct verbs tense and singular or plural nouns - only the final response was recorded. The participants were not instructed to focus on one task over the other. This was fundamental as it may have altered their attention towards a particular task, which would accordingly increase the resources used to attend to it, which might result in an inaccurate deterioration of the concurrent task (Zok, Mazzà and Cappozzo 2008).

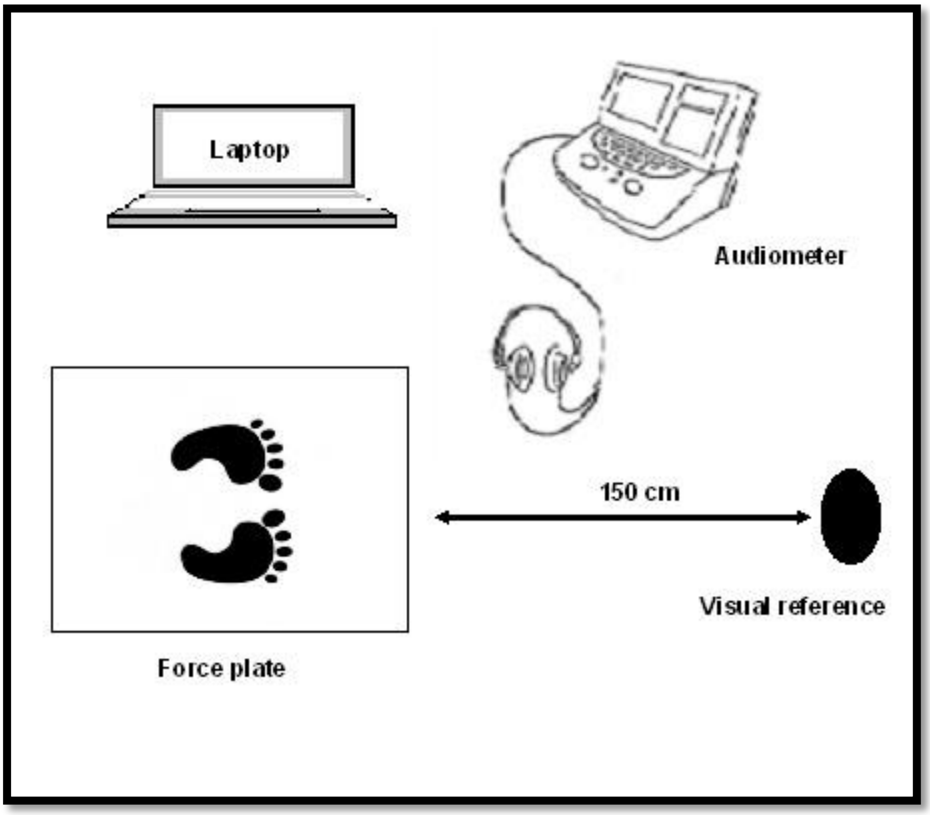


Figure 3-3 Schematic diagram representing room set-up

Other factors that may alter postural stability and hence recording, such as the base of support, foot position and wearing headphones, were also considered. An increased base of support, i.e. feet wider apart, would increase the participant's stability and vice versa (Duarte and Zatsiorsky 2002). Standardising foot position has been reported as important in evaluating postural stability (Chiari, Rocchi and Cappello 2002). However, regulating the position of the feet for all of the participants neglects the participants' characteristics, and so self-selecting the most comfortable position was recommended by (Duarte and Freitas 2010). In this study, the self-selection of the most comfortable position was adopted. Standing in the same position for each participant through all testing conditions was assured, as each participant's self-selected most comfortable position for the first condition was marked/recorded. During all of the testing conditions, the participants wore the headphones, regardless of the presence or absence of the auditory task, as taking off the headphones might change the participant's spatial awareness/orientation, and hence, postural stability.

3.3 Data and Results

3.3.1 Data

Every calculated sway measure showed an increased body sway in at least two testing conditions while dual tasking, compared to their baselines. However, there were measures that showed no change in postural stability in some conditions. In general, mean scores for the calculated measures revealed a trend of increasing postural stability in dual-task conditions and the harder the posture task became the more the stability decreased (Table 3-1), this is also available as a figure (15-1) in appendix 15. MSA_{ML}, MR_{AP} and MV revealed increased body sway every time the listening task was introduced and this increment built-up as the postural task increased in difficulty, either by narrowing the base of support in Romberg conditions or by vision deprivation during eyes closed conditions – i.e. from the easiest NSO to the hardest RSCA. Figure 3-3 shows this trend for MV.

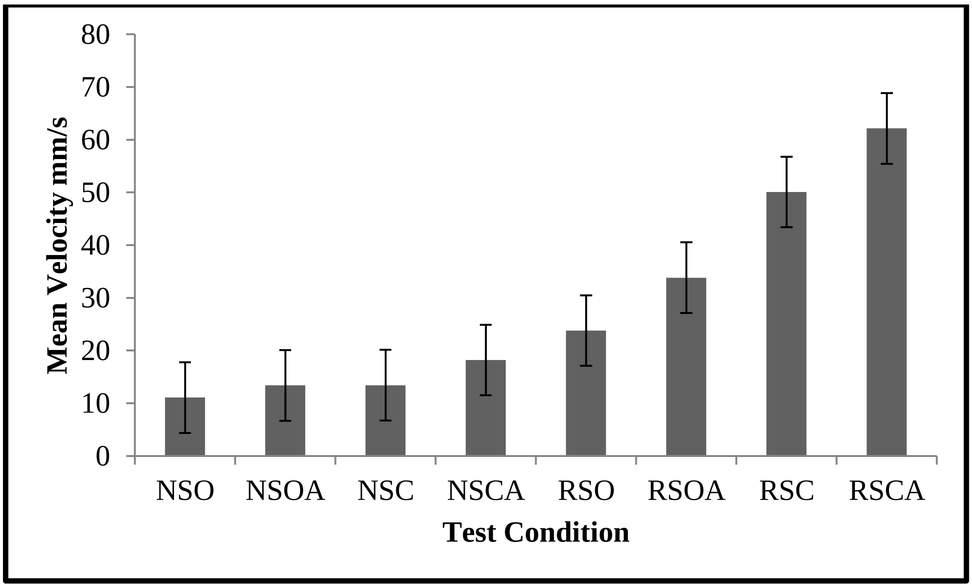


Figure 3-4 Changes in MV under different testing conditions. Y axis represents mean score for MV for testing conditions and X axis represents testing conditions.

Table 3-2 COP measures in different test conditions. Unites of COP measures are as follows: mm (amplitude/SD amplitude/Range/planar deviation); mm/s (SD velocity/mean velocity); mm²/s (Area); phase plane and total phase plane is in an arbitrary unit. Values are mean ± (standard deviation)

	NSO		NSC		RSO		RSC	
COP measures	**No Listening**	**Listening task**	**No Listening**	**Listening task**	**No Listening**	**Listening task**	**No Listening**	**Listening task**
MA (ML)	0.92 ± (13.41)	2.27 ± (12.99)	-0.53 ± (13.78)	1.67 ± (12.00)	-11.10 ± (23.42)	-15.60 ± (21.86)	-13.90 ± (25.09)	-15.69 ± (23.52)
MA (AP)	58.73 ± (29.23)	54.13 ± (36.34)	55.18 ± (39.12)	57.75 ± (33.74)	69.34 ± (34.51)	69.47 ± (36.84)	58.22 ± (33.57)	64.75 ± (37.07)
SDSA (ML)	2.94 ± (1.69)	2.94 ± (1.76)	2.89 ± (1.80)	3.18 ± (1.88)	5.34 ± (1.30)	5.45 ± (1.36)	10.39 ± (2.16)	9.96 ± (1.95)
SDSA (AP)	5.28 ± (2.66)	5.14 ± (2.89)	5.78 ± (2.02)	6.00 ± (2.78)	5.53 ± (2.54)	5.04 ± (2.56)	7.52 ± (3.27)	7.68 ± (3.39)
MR (ML)	17.39 ± (11.63)	16.93 ± (12.91)	15.98 ± (8.68)	18.52 ± (13.08)	28.72 ± (6.07)	29.54 ± (8.11)	53.27 ± (16.59)	50.62 ± (10.78)
MR (AP)	26.63 ± (18.37)	27.77 ± (18.41)	29.02 ± (10.01)	32.68 ± (18.43)	29.69 ± (12.14)	29.97 ± (19.09)	54.19 ± (31.42)	54.82 ± (38.14)
SDV (ML)	291.55±(167.27)	290.13±(174.39)	286.17±(178.39)	314.61±(186.15)	528.94±(128.60)	538.7 ±(134.92)	1028.55±(213.92)	985.55±(193.38)
SDV (AP)	522.53±(263.39)	511.72±(285.98)	572.51±(199.90)	594.04±(275.46)	547.72±(251.86)	498.15±(254.43)	744.34± (323.31)	760.71±(335.99)
PPP (ML)	291.57±(167.28)	290.14±(174.40)	286.18±(178.40)	314.63±(186.16)	528.97±(128.60)	538.81±(134.93)	1028.60±(213.93)	985.60±(193.39))
PPP (AP)	522.56±(263.41)	511.75±(286.00)	572.54±(199.91)	594.08±(275.48)	547.75±(251.87)	498.19±(254.44)	744.37± (323.32)	760.75± (336.01)
MV	11.07± (3.14)	13.37± (4.73)	13.41± (2.56)	18.19± (3.84)	23.78± (6.82)	33.82± (11.37)	50.07± (16.89)	62.14± (17.54)
PD	6.18± (2.88)	6.05± (3.18)	6.60± (2.34)	6.89± (3.14)	7.85± (2.37)	7.56± (2.49)	12.99± (3.32)	12.76± (3.25)
SA	5.90± (6.41)	5.90± (6.75)	5.83± (5.55)	7.10± (6.66)	9.80± (6.07)	9.22± (7.52)	25.73± (15.04)	25.09± (14.92)
TPPP	607.80±(272.80)	605.00±(298.86)	659.48±(215.48)	714.72±(326.44)	879.24±(322.99)	747.33±(235.58)	1130.59±(444.81)	1278.5± (317.71)

Several other trends were also evident. In the presence of the auditory task and regardless of the standing position, TPPP, PPP_{AP}, SDV_{AP} and $SDSA_{AP}$, all revealed increased body sway when the eyes were closed, and less body sway when the eyes were open. See Figure 3-4 where $SDSA_{AP}$ is plotted as an example.

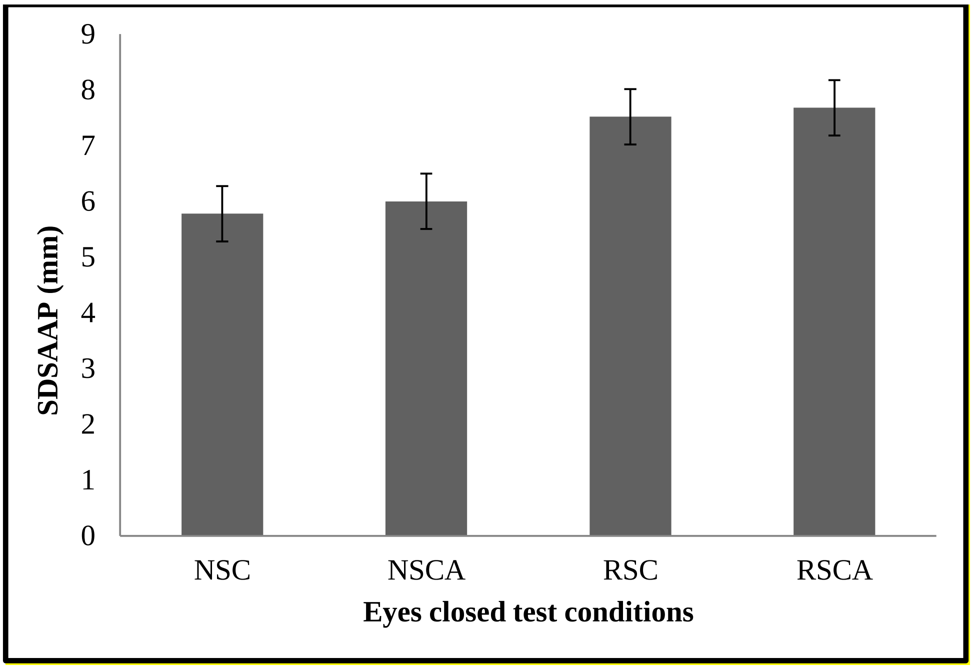


Figure 3-5 Changes in $SDSA_{AP}$ under testing conditions in presence of a listening task. Y axis represents mean score for $SDSA_{AP}$ measured in mm and X axis represents eyes closed testing conditions.

For the listening task, average scores in dual task conditions differ from their baselines, however, participants were able to score at least 75% correct in the most difficult dual task conditions. See Figure 3-5 where listening task average scores under the different posture task conditions are plotted. The performance of the listening task deteriorated as the posture task difficulty increased. The most difficult posture condition, RSCA, showed the least listening performance and this may be explained by a reallocation of attentional resource with increasing posture task complexity. Listening task scores under NSCA, RSOA and RSCA differed significantly from the baseline.

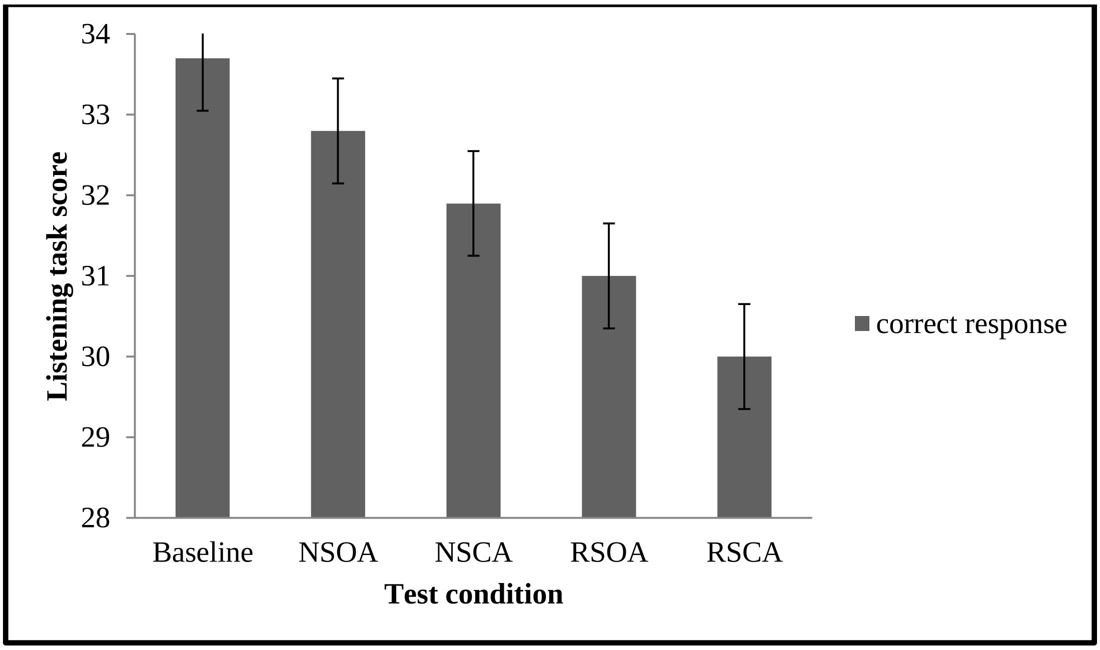


Figure 3-6 Listening task average scores under different test conditions.

3.3.2 Results

All of the nine calculated sway measures were statistically tested using repeated measures ANOVA with 2×2×2 design. The model was built based on two levels of the posture task (NS and RS), two levels of vision (eyes open and closed) and two levels of the listening task (no listening task and listening task requires responding). Test assumptions i.e. normality, sphericity and homogeneity were met. Results of the COP measures are presented in Table 3-2.

Table (3-3): Summary of ANOVA results on the effect of listening on posture control for all calculated COP measures. *F* = *F* value, *P*-value = significance, *MSE* = mean square error and η_P^2 = partial eta squared. Effect degree of freedom = 1 and error degree of freedom = 98 for all COP measures.

Measure	*F*	*P*-value	*MSE*	η_P^2
MSA (ML)	0.81	0.37	0.02	0.01
MSA (AP)	0.67	0.45	0.04	0.05
SDSA (ML)	0.07	0.79	0.02	0.01
SDSA (AP)	1.09	0.30	0.03	0.01
MR (ML)	0.01	0.91	0.03	0.00
MR (AP)	0.07	0.79	0.03	0.00
SDV (ML)	0.05	0.83	0.02	0.00
SDV (AP)	1.04	0.31	0.03	0.01
PPP (ML)	0.05	0.83	0.02	0.00
PPP (AP)	1.03	0.31	0.03	0.01
MV	34.22	**< 0.001**	0.01	**0.26**
PD	1.04	0.31	0.09	0.01
SA	0.56	0.46	0.07	0.01
TPPP	0.45	0.51	0.02	0.00

MV was the only COP measure to detect an effect of listening on posture control. At this measure, main effects were found for posture $\boldsymbol{F}$ (1, 98) = 1614.04, MSE = 0.03, $p < 0.001$, $\eta_P^2 = 0.94$, vision $\boldsymbol{F}$ (1, 98) = 871.43, MSE = 0.01, $p < 0.001$, $\eta_P^2 = 0.90$ and listening $\boldsymbol{F}$ (1, 98) = 34.22, MSE = 0.01, $p < 0.001$, $\eta_P^2 = 0.26$ (Figure 3-6). Results for posture and vision were no further explored, as it is a well-documented in the literature that eyes closed and tandem stance conditions are more difficult postures to maintain (Le Clair and Riach 1996). Furthermore, this study did not set out to compare different types of vision statuses or stances; its goal was to investigate whether inducing a more difficult posture (through vision deprivation or narrowing the base of support) would disrupt postural sway more with the addition of a listening task.

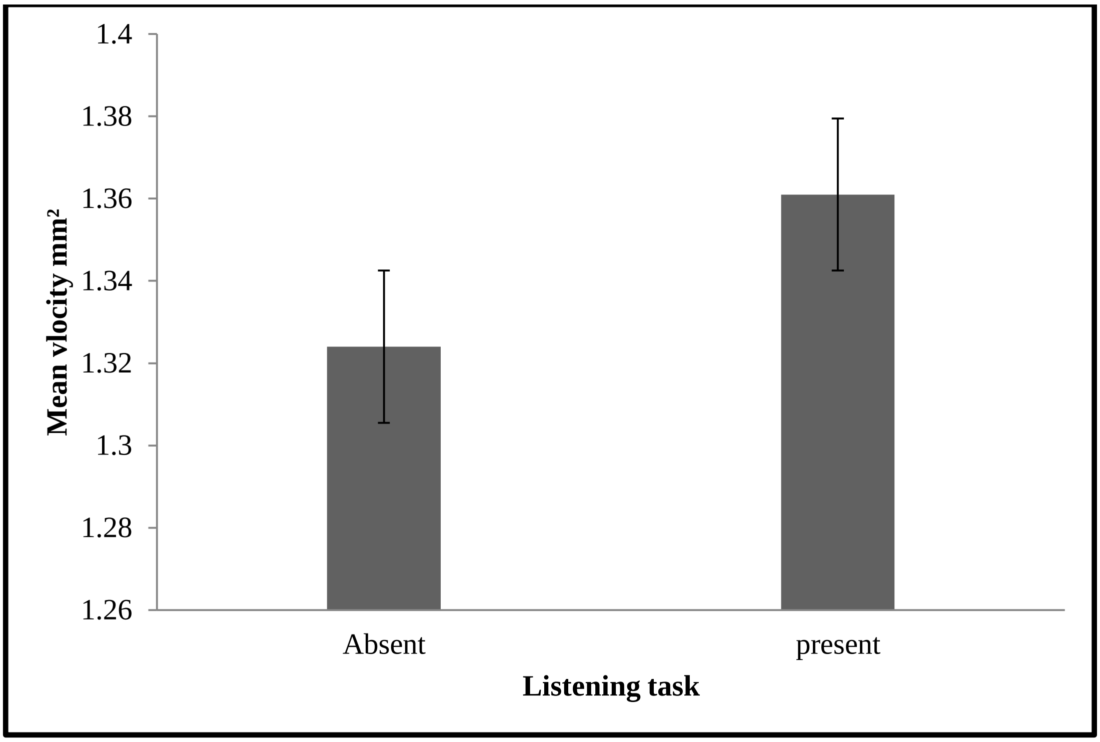


Figure 3-7 MV showed a significant main effect of listening on posture. Y axis represents the body sway measure, MV, and the X axis shows the condition of the listening task

Listening simple effects were examined using pairwise comparisons, which were conducted using a Bonferroni protected alpha level of p= 0.05. Pairwise comparisons revealed that when participants were standing normally, NS, mean velocity was able to detect a change in postural stability. Postural stability reduced when participants had to maintain upright posture and listen and respond compared to conditions were they had no listening task (mean difference = 0.48, SE = 0.01, $p < 0.001$, $\eta_P^2 = 0.16$). These results indicates that 16% of the variance is accounted for by listening during normal stance. Moreover, postural sway increased when the listening task was introduced to participants standing in the tandem position, RS, and they were required to verbally respond, compared to conditions were there was no listening task (mean difference = 0.25, SE = 0.01, $p < 0.001$, $\eta_P^2 = 0.14$). This means, while participants is standing in the RS and dual tasking, 14% of the variance observed is accounted for by the listening task. To conclude, MV indicated a significant increase in body sway during both NS and RS when participants were responding to a listening task under dual task conditions.

The correlation between the COP sway measures calculated in this experiment were also examined with the aim of exploring if the measures are interlinked. Although previous literature investigated heavily and focused on within-day, intra-session, and between-days reliability of several COP sway measures using intra-class correlation (Li et al. 2016; Bauer et al. 2008; Lafond et al. 2004), these studies examines the correlation between the same measure results and do not continue to explore between measures correlation. Yet, correlation between some of the measures calculated in this experiment but not all of them have been investigated before (Karlsson and Frykberg 2000). Table 3-3 represents Pearson's correlation coefficient between the COP sway measures calculated here: MA_{ML}, MA_{AP}, $SDSA_{ML}$, $SDSA_{AP}$, MR_{ML}, MR_{AP}, SDV_{ML}, SDV_{AP}, PPP_{ML}, PPP_{AP}, MV, PD, SA, TPPP.

From the correlation table it is evident that these measures are correlated. This is expected especially that they all come from the same entity, related to Centre of mass and are quantified from the same raw data (Winter ; Karlsson and Frykberg 2000).

Table (3-4): Pearson's correlation coefficient between calculated COP measures. Mean Amplitude medio-lateral (MAML), Mean Amplitude anterior-posterior (MAAP), Standard Deviation of Sway Amplitude medio-lateral (SDSAML), Standard Deviation of Sway Amplitude anterior-posterior (SDSAAP), Movement Range medio-lateral (MRML), Movement Range anterior-posterior (MRAP), Standard Deviation of Velocity medio-lateral (SDVML), Standard Deviation of Velocity anterior-posterior (SDVAP), Phase Plane Portrait medio-lateral (PPPML), Phase Plane Portrait anterior-posterior (PPPAP), Mean Velocity (MV), Planar Deviation (PD), Sway Area (SA), Total Phase Plane Portrait (TPPP). All values in **bold** are significant (***P*** < 0.05).

Sway	MSAML	MSAAP	SDSAML	SDSAAP	MRML	MRAP	SDVML	SDVAP	PPPML	PPPAP	MV	PD	SA	TPPP
MAML	1.00													
MAAP	**0.87**	1.00												
SDSAML	**0.84**	0.53	1.00											
SDSAAP	0.56	**0.72**	**0.69**	1.00										
MRML	**0.73**	**0.68**	**0.83**	0.61	1.00									
MRAP	**0.54**	**0.81**	0.59	**0.85**	**0.54**	1.00								
SDVML	**0.74**	0.58	**0.60**	**0.59**	**0.59**	0.49	1.00							
SDVAP	0.47	**0.59**	**0.73**	**0.78**	0.51	**0.76**	**0.62**	1.00						
PPPML	0.46	**0.76**	**0.58**	**0.69**	0.52	0.35	**0.32**	0.47	1.00					
PPPAP	0.09	**0.64**	**0.52**	**0.60**	0.21	0.46	**0.44**	0.53	0.69	1.00				
MV	**0.79**	**0.67**	**0.87**	**0.86**	**0.49**	0.57	**0.94**	**0.91**	0.27	0.32	1.00			
PD	**0.67**	**0.64**	**0.71**	**0.69**	**0.53**	**0.69**	0.36	0.38	0.44	0.54	**0.73**	1.00		
SA	**0.58**	**0.63**	**0.64**	**0.74**	0.61	**0.59**	**0.79**	0.72	0.36	0.48	**0.89**	**0.63**	1.00	
TPPP	0.37	**0.50**	**0.78**	**0.71**	0.36	0.39	0.47	**0.55**	**0.78**	**0.81**	0.57	0.41	0.39	1.00

3.4 Discussion

In this study, the participants carried out a listening task under different conditions that challenged their postural control. Both their performance in the listening task and postural control were assessed. The results of this dual-task paradigm show that performance on both tasks deteriorated and were most affected when measured in conjunction with the most difficult postural tasks.

The study's first goal was to examine the effects of a listening task on postural control. Results of the MV measure demonstrated increased postural sway while listening. Deteriorations in the performance of the listening task was also evident. The change in body sway, increased, when responding to a listening task was significant and of a large effect size suggesting a true effect of listening on posture control. Considering that the listening task used in this experiment is a form of a cognitive task, findings from this experiment are consistent with previous research concerning posture and cognition in general which reported a change in either the postural task or the cognitive task or both of them (Kerr, Condon and McDonald 1985; Maylor and Wing 1996; Shumway-Cook et al. 1997).

The study's second goal was to determine the most suitable sway measure to detect the effect of the listening task. MV was the only measure that showed a listening main effect on postural control tasks of varying difficulties. MV has been reported in previous literature as a sensitive body sway measures able to reflect modifications in posture control either due to the type and difficulty of the posture position and condition (Doyle et al. 2008) or due to the addition of a cognitive task (Moghadam et al. 2011). However, to the knowledge of the researcher, MV has not been reported in previous literature to be specifically sensitive to changes caused by listening. The novelty of this finding, MV is sensitive to postural listening effects, will give insights into where to take this research further and also have clinical implications for clinicians who are quantifying posture performance especially with individuals having hearing difficulty.

It was also noted that by increasing the difficulty of the postural task, the number of errors in the listening task were also increased, especially in the most difficult posture stance condition, that is RS. This can be justified, as introducing a more challenging stance position were dissimilar enough to produce changes in attentional demands (Andersson, Yardley and Luxon 1998). Another factor that might also have played a role in increasing postural sway is articulation, as the listening task in this study required the participants to respond verbally to what they had heard. Articulation has been reported to increase postural sway (Yardley et al. 1999; Dault, Yardley and Frank 2003). Articulation causes changes in respiration (Conrad and Schönle 1979) which was reported to affect posture (Jeong 1991).

To conclude, nine postural sway measures, five of them in the AP and ML directions, were calculated and analysed, only MV showed an increased body sway in dual-task conditions where the participants were performing the postural task and the listening task simultaneously. This can be justified as MV is the only measure that is calculated based on changes of displacement over time in both directions. The other measures, such as MSA, MR, MSASD (either in ML or AP directions) are calculated from displacement only. SDV might have not been sufficient enough to show the effect of the listening task on posture, because it was calculated from each direction, AP and ML, separately. Although the effect size of the MV measure was less compared to the posture and vision effect sizes, it is still large enough to support the significance level indicated.

The findings from this chapter have raised new research questions around the effect of listening whilst maintaining an upright posture. Why does a postural sway measure, MV, detect this effect while others do not? The ability of MV to detect this effect of listening on posture indicates that there is a time element involved, as it is calculated based on changes over time in both AP and ML directions. Is there a more sophisticated way or measure that can be used to explore this suggested relationship between listening and posture? What are the underpinning strategies and mechanisms that individuals use to maintain an erect posture while listening? These queries directed the research towards the second experiment covered in the coming chapter, Chapter Four. This chapter employed a more sophisticated postural sway analysis method called stabilogram diffusion analysis (SDA).

4 Chapter Four:

Studying the Influence of Listening on Posture Control Using Stabilogram Diffusion Analysis

4.1 Introduction

In the previous chapter, nine body sway measures were calculated from raw COP data in order to investigate the effect of listening on posture control in healthy adults with normal hearing. Results revealed that mean velocity was the only index that showed a significant main effect of listening on posture.

It was proposed in Chapter Three that mean velocity was successful in detecting the effect of listening on posture control, maybe because it is calculated based on the change in displacement in AP and ML directions over time, while the rest of the COP measures are mainly summary statistics of displacement or velocity in one direction only. To examine this observation and in order to understand the underpinning strategies that individuals use to maintain their posture whilst listening, the experiment in this chapter aimed to investigate the matter, using a different method to analyse COP data.

To achieve this, posture behaviour whilst listening in healthy individuals with normal hearing will be studied using a more sophisticated sway parameter; Diffusion Plots (DP) will be calculated using Stabilogram Diffusion Analysis (SDA). This conceptual and theoretical framework for studying the postural control system in this way is based on the assumption that maintaining upright posture can be viewed as a stochastic process. Maintenance of upright posture can be viewed as a stochastic process, because of the continuous and random variations of COP trajectories recorded during quiet stance. This framework was first described by

Collins and De Luca (1993) and proposes that open-loop and closed-loop mechanisms are involved in maintaining erect posture during quiet stance.

Their suggestion of the involvement of an open-loop challenged the concept that upright posture is only controlled by feedback mechanisms. In their model, the feedback time delay connected to the delay in neural transmission, and processing sensory and motor signals, was described as open loop. However, their model does not exclude closed feedback mechanisms; it rather utilises them over long time intervals (more than one second), while open-loop mechanisms are involved over short time intervals (less than one second).

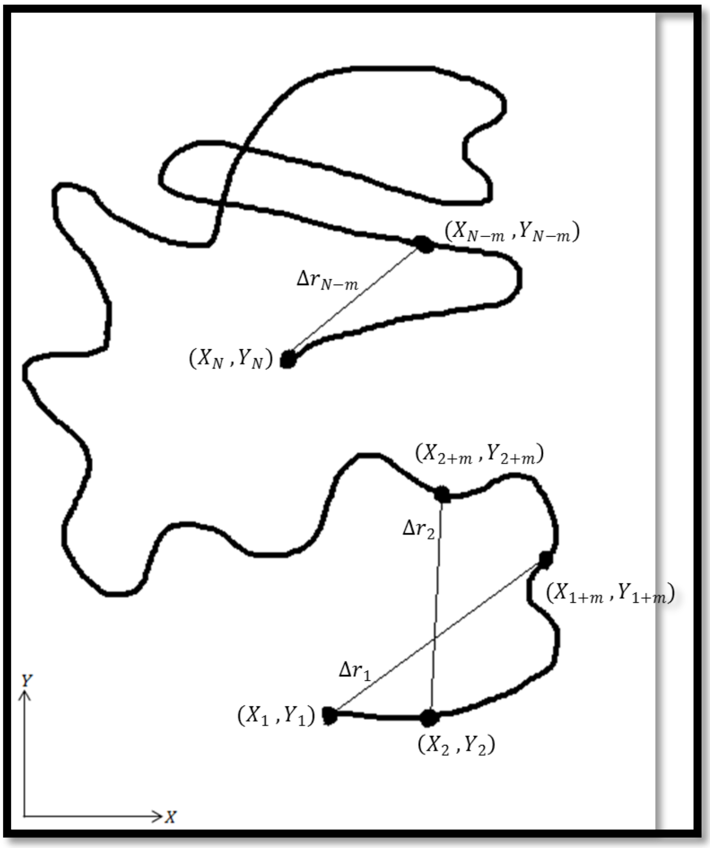
These open-loop and closed-loop strategies suggest that sway velocity might change during the recording period, over time, and there is an indication of this from mean sway. Therefore, SDA is an analysis system that allows exploration of the nature of that change, which may relate to strategies used in postural control. Thus, SDA is used in this experimental chapter to study how listening affects these postural control strategies. This chapter will provide an overview of SDA and its measures, followed by the experimental protocol and procedure, analysis, results and discussion.

4.1.1 Stabilogram diffusion analysis

In 1993, Collins and De Luca noticed that the continuous and random variations of COP behaviour recorded during quiet stance may have some characteristics that look similar to a Brownian Motion. This led them to assume that maintaining upright posture can be viewed as a stochastic process. In the dictionary, a stochastic process is defined as “Having a random probability distribution or pattern that may be analysed statistically but may not be predicted precisely” (OxfordDictionary 2014). In other words, a stochastic process can be defined as a random process changing or evolving or developing over time. This means that even if it is known how a system is currently performing and how it has performed in the past, it is not possible to predict precisely its performance in the future. However, the probabilities of its performance can be estimated.

This stochastic process concept has been used in different disciplines to study numerous phenomena where the behaviour of interest varies in time in an erratic way. These phenomena have been studied by developing mathematical models that can estimate the probability distributions of possible outcomes calculated from random variation of inputs over time (Ibe 2013). From this point of view, Collins and De Luca (1993) established mathematical techniques, SDA, to describe the random behaviour of the human postural control.

Collins and De Luca analysed COP trajectories as one-dimensional and two-dimensional random walks using SDA. Random walks, for the purpose of posture control here, can be defined as the movement of an object or change in a variable that is unpredictable. In this case the object/variable is COP. The COP coordinates “walk” across the force plate (constrained by the force plate dimensions) in a random manner (see Figure 4-1). This method involves calculating changes in COP for different increments of time.



$$< \Delta r^2 >_{\Delta t} = \frac{\sum_{i=1}^{N-m} (\Delta r_i)^2}{(N-m)} \qquad \text{(Equation 1)}$$

Figure 4-1 The method used for calculating mean square planar displacement $< \boldsymbol{\Delta r^2} >$ as a function of time interval <Δt> for a COP trajectory. Source (Collins and De Luca 1993).

The square of the displacements of the COP trajectories are calculated between all pairs of points separated by a specific time interval $\Delta\mathbf{t}$**,** and then averaged. Equation one illustrates that the mean square planar displacement <Δr^2> was quantified as a function of time interval, Δt, for a COP trail made up of several, N, data points $(x_1, y_1;\ x_2, y_2;\ \ldots\ x_N, y_N)$ over increasing data intervals, donated as *m* in equation one. This process is repeated for increasing values of $\Delta\mathbf{t}$. The mean square COP displacement is then plotted against $\Delta\mathbf{t}$ for each trial and averaged overall trials to obtain a resultant stabilogram-diffusion plot for a particular subject-condition combination (see Figure 4-2). The resultant plots for each participant on each test condition was used for the stabilogram diffusion analysis.

The stabilogram-diffusion analysis involves the extraction of three sets of posturographic parameters: diffusion coefficients, scaling exponents and critical point coordinates (Collins and De Luca 1993; Collins and De Luca 1994; Collins and De Luca 1995a; Collins and De Luca 1995b; Collins et al. 1995). The diffusion coefficient is an average measure of the stochastic activity of a random movement. This brings back the observation of Collins and De Luca who noticed that COP behaviour have some characteristics that looks similar to a Brownian Motion. Einstein studied the simplest form of this motion, one-dimensional random walk, and indicated that its mean square displacement <Δx^2> was related to the time interval Δt by the following expression:

$$<\Delta r^2> = 2\,D\Delta t, \quad \text{(Equation 2)}$$

Where: *D* is the diffusion coefficient (Einstein 1905).

So diffusion coefficient is directly related to its amplitude and can be thought of as an indicator of the relative stability of the system (Doyle et al. 2008; Martin and Clark 2012). The short-term and long-term COP diffusion coefficients characterise the stochastic activity of the open-loop (non-feedback) and closed-loop (feedback) postural control mechanisms, respectively (Collins and De Luca 1995a; Collins et al. 1995).

Diffusion coefficients are calculated from the slopes of the resultant linear-linear plots of mean square COP displacement versus the change in time (Collins et al., 1995). The long-term diffusion coefficients are usually lower than the respective short-term diffusion coefficients, which reflects the increased level of stochastic activity over the short-term time series compared to the long-term time series (Collins and De Luca 1993).

The quantification of the correlation between the step increments that make up an experimental time series is the second posturographic parameter used in SDA and these are termed “scaling exponents” (Collins and De Luca 1993). To enable calculations of this parameter, equation two has to be modified to account for random walks in a plane/multi-dimensional rather than one-dimension. So equation two which describes a simple Brownian motion has to be extended to account for fractional Brownian motion (Mandelbrot and Van Ness 1968). So the relation given by Einstein to describe mean square displacement, equation one, was generalized to a scaling law as follows:

$$<\Delta r^2> = 2D\ (\Delta t)^{2H} \text{ (Equation 3)}$$

Where: H can be any value in the range $0 < H < 1$.

Hence, one can say $H = 0.5$ for simple classical Brownian motion.

Scaling exponents are calculated from the slopes of the resultant log-log plots of mean square COP displacement versus the change in time. This measure can be thought of as providing an indication of whether the motion of the COP is more or less likely to continue moving in the same direction that it is currently moving in (Doyle et al. 2008). As previously mentioned, scaling exponents may assume a value in the range of 0 to 1. If the scaling exponents are equal to 0.5, then the increments in COP displacements are statistically independent. If the scaling exponent value is greater than 0.5, then past and future increments are positively correlated, i.e., future displacement increments tend to move in the same direction as the current displacement value (termed persistent behaviour). However, If scaling exponents are less than 0.5, then the stochastic activity is negatively correlated, i.e., increasing/decreasing trends in the past imply decreasing/increasing trends in the future (termed anti-persistent behaviour) (Collins and De Luca 1993; Collins et al. 1995; Peterka 2000). From a physiological standpoint, SDA scaling exponents quantify the correlated behaviour of the respective postural control mechanisms, i.e., short-term scaling exponents characterise the drift-like dynamics of the open-loop postural control mechanisms, whereas the long-term scaling exponents characterise the antidrift-like dynamics of the closed-loop postural control mechanisms (Collins et al. 1995).

The critical point coordinates approximate the transition region that separates the short-term and long-term regions. The estimation of the critical point coordinates is determined by the intersection point of the straight lines fitted to the two regions of the linear-linear version of the resultant stabilogram-diffusion plot. The transition points occur at relatively small time intervals (0.33 to 1.67 s) and small mean square displacement (1.10 mm^2 to 29.37 mm^2) (Collins and De Luca 1993; Collins and De Luca 1994; Collins and De Luca 1995b). These coordinates approximate the temporal and spatial characteristics of the region over which the physiological postural control system switches from open-loop control to closed-loop control.

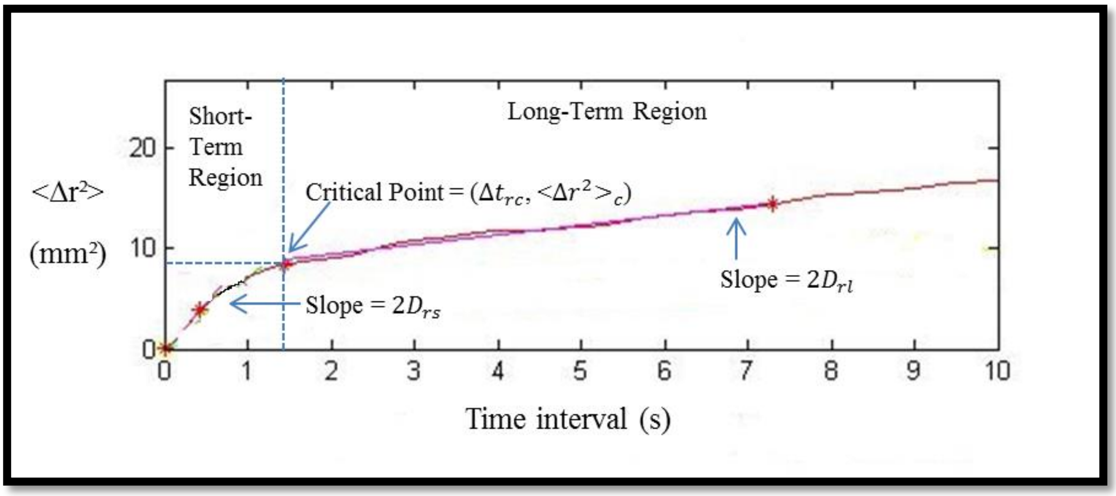


Figure 4-2 A resultant planar stabilogram-diffusion plot (<Δr2> Vs Δt) generated from COP time series according to the method illustrated in Figure 4-1. The diffusion coefficients Drs and Drl are calculated from the slopes of the regression lines fitted from the short-term and long-term regions, respectively. The critical point = (Δt_{rc}, $<\Delta r^2>_c$), is defined by the intersection of the two lines fitted to the two regions of the plot. The scaling exponents H_{rs} and H_{rl} are computed from the slopes of the log-log plots of the short-term and long-term regions, respectively.

The plot can be divided into two regions, which indicate two types of behaviour. Displacements were positively correlated (that is called persistence), over short time intervals (Δt <1s), and displacements were negatively correlated (anti-persistence), over longer time intervals (Δt >1s). The transition point, with coordinates (Δt_{rc}, $<\Delta r^2>_c$), marks the intersection of the short-term and long-term regions within the stabilogram diffusion plot. Diffusion coefficients and scaling exponents were estimated for the short-term and long-term regions for AP, ML and planar dimensions, as described by Collins and De Luca (Collins and De Luca 1993).

Their model, as in the plot above, implied that the postural control system used open-loop control systems (non-feedback) over short-term time intervals, where COP drifted away from a stable position, and closed-loop control mechanisms (feedback) over long-term time intervals, where COP returned towards equilibrium. This suggests that the feedback mechanisms are used to control COP after some time (< 1s) when sensory information detects that COP has moved beyond a threshold. Peterka (2000) stated that the time required to recruit the closed-loop system depends upon the time taken to sense the movement, transmit the impulses, process the information and activate the muscles.

In their study, Collins and De Luca (1993) calculated the SDA of 25 male adults by plotting mean square displacement against time interval. A typical time series plot demonstrates two distinct parts – the short-term region and the long-term region. In the short-term region, the COP trajectory shows persistence, that is, the COP tends to drift away from a relative equilibrium point. This is interpreted as the short-term region corresponding to the open-loop system of balance. The long-term analysis shows anti-persistence behaviour. That is, the COP tends to return back to the relative point of equilibrium during this phase. The long-term region is thought to correspond with the closed-loop system of balance (Collins and De Luca 1993; Nashner 1976; Nashner, Granit and Pompeiano 1979; Peterka 2000; Yim-Chiplis and Talbot 2000). Furthermore, the analysis revealed that the two phases are separated by a point known as the critical point. The point corresponds to the time and distance after which COP displacement is detected, and the person switches from the open-loop system to the closed-loop system (Collins and De Luca 1993).

A number of subsequent studies have used SDA to examine the effect of different test conditions on open-loop and closed-loop control mechanisms during quiet stance in different populations. For example, Collins et al. (1995) demonstrated that older adults utilise open-loop systems for a greater duration and switch to the closed-loop after a greater displacement. This study also demonstrated that traditional body sway measures were not sensitive enough to detect these age-related changes (Collins et al. 1995). Riley et al. (1997a) studied the effect of external perturbation in the form of tilting, and the effect of vision on the parameters of postural stability in normal adults (Riley et al. 1997a). Riley et al. (1997b) also studied the effect of conflicting inputs of touch and vision on the same parameters. In both studies the researchers found that the two parts of the SDA, that is the short-term phase and the long-term phase, are preserved under all conditions (Riley et al. 1997a; Riley et al. 1997b).

From the review provided above, SDA may provide more information and help to give greater insight into the postural control system compared to traditional body sway measures. SDA considers the involvement of open-loop mechanisms that might have a role in maintaining erect posture. This is different to many postural control studies (Diener et al. 1991; Diener, Horak and Nashner 1988; Dietz et al. 1991; Horak, Nashner and Diener 1990; Moore et al. 1988; Woollacott, Von Hosten and Rösblad 1988) that investigated body sway under different perturbations and assume postural control systems are controlled by feedback mechanisms only which may explain the lack of inconsistency in their findings.

In this thesis it is hypothesised that introducing a listening task whilst maintaining posture under different conditions will cause a greater time delay and a larger displacements before shifting from open-loop (non-feedback) mechanisms to closed-loop (feedback) mechanisms. It is expected to see a larger body sway, higher stochastic activity, in the short term region and a delayed critical points before participants change posture strategies during dual-task conditions when compared to baseline conditions. It is expected that using this analysis will lead to a greater understanding of the strategies utilised by the postural control system to maintain upright posture while concurrently listening.

4.2 Methods

4.2.1 Ethical approval

An ethical approval from the School of Healthcare Research Ethics Committee (SHREC) was obtained for this study (SHREC/RP/296) (see Appendix Nine).

4.2.2 Inclusion and exclusion criteria

As already noted before, the inclusion criteria were: age between 18 and 60 years; normal middle ear function (ear pressure from -500 to +500 pa, compliance between 0.3 to 1.6 cm^3 and ear canal volume between 0.6 to 1.5 cm^3); normal hearing bilaterally (hearing thresholds better than or equal to 20 dB HL across octave frequencies between 250 and 8 kHz); no personal history of balance problems; and English as the first language. Any participant who did not meet the inclusion criteria, reported neurological disorders, motor problems, severe visual impairments or were unable to understand and follow verbal instructions, was excluded from the study. Participants were advised to visit their GPs if abnormal hearing thresholds were noted.

4.2.3 Participants

Thirty-two participants was indicated to be required by Cohen's tables for a study with 85% power, alpha = 0.05 and r = 0.50 (Cohen 1988). 32 normal healthy participants, 17 males and 15 females, aged 20 – 57 years old (mean age 36.5 SD± 9.4) met the inclusion criteria and were included in the study; none of them had participated in the previous experiment. The participants were mainly students and staff recruited from the university community, but participants from around Leeds city were also recruited. The study was advertised using posters inside the university campus and around Leeds city, by circulating emails to university staff and students, and by the snowball strategy. Written informed consent (see Appendix Three) was obtained from all participants after oral information, printed information (see Appendix Ten) and the opportunity to ask questions had been given. Participants were given vouchers to recompense them for their time.

4.2.4 Equipment and listening task

4.2.4.1 Screening equipment

A calibrated Interacoustics MT10 middle ear analyser was used to perform tympanometry testing using a 226 Hz probe tone to establish middle ear function and a calibrated FONIX Hearing evaluator, FA-12 Digital Audiometer was used to conduct audiometry to evaluate hearing thresholds.

4.2.4.2 Experimental equipment

In order to evaluate posture, COP data were collected using a Kistler force plate type 9286BA and an amplifier or a data Acquisition Box (Kistler DAQ Type 5691A) controlled by a SAMSUNG laptop running proprietary software (BioWare v5.1.1.0). Using the laptop and the analysis software, BioWare, these data were controlled and processed to give visual and statistical data of ground reaction forces, moments and most importantly, COP (see section 3.2.4.2 for details).

4.2.4.2.1.1 SDA measures

Diffusion coefficients, scaling exponents and critical points were calculated from the COP data using the SDA technique, as described by (Collins and De Luca 1993). This calculation was executed in MatLab (version 8.1.0.604) using a code written by Andrea Stamp, which she made available online (see Appendix 11). Stamp stated that the software produces average stabilogram diffusion plots from multiple trials, as defined and utilised by Collins in many papers.

4.2.4.3 Listening stimuli

The listening stimuli used in this experiment was a validated competing words test-free recall which is a sub-test of the SCAN-3. The task used here is similar to the task used in Experiment One in Chapter Three; for further details see 3.2.4.3.

4.2.5 Procedure

The procedure conducted in this experiment followed the procedure in Experiment One in Chapter Three. However, in the dual-task experiment, the participants were asked to complete five 30-seconds trials recorded with a sampling rate of 100 Hz. The tests were carried out in the audiology laboratory at the School of Healthcare at the University of Leeds. Every participant underwent a screening, which involved obtaining medical history (see Appendix Seven) followed by a routine clinical audiological assessment. After screening, the participants who met inclusion criteria were tested in the dual-task experiment (see section 3.2.5 for the detailed procedure).

4.3 Data and Results

Diffusion coefficients, scaling exponents and critical time points data and results in the planar, which is the vector sum of anterior-posterior and mediolateral components of the analysis, will be presented and discussed here. As SDA measures were calculated for the short-term and long-term regions, subscripts *S* and *L* will denote the short-term and long-term regions, respectively.

4.3.1 Data

4.3.1.1 Diffusion coefficients

Looking into this component of SDA analysis revealed a number of observations. First, the short-term diffusion coefficients were larger than their respective long-term diffusion coefficients under all test conditions. Secondly, diffusion coefficients in the short-term region were generally greater in conditions NSOA, NSCA, RSOA and RSCA, where the participants were performing a posture task concurrently with a listening task, when compared to the conditions, NSO, NSC, RSO and RSC, where they were performing a posture task. Figure 4-3 illustrates this trend in test conditions completed under NS and RS with eyes open and eyes closed with and without a listening task.

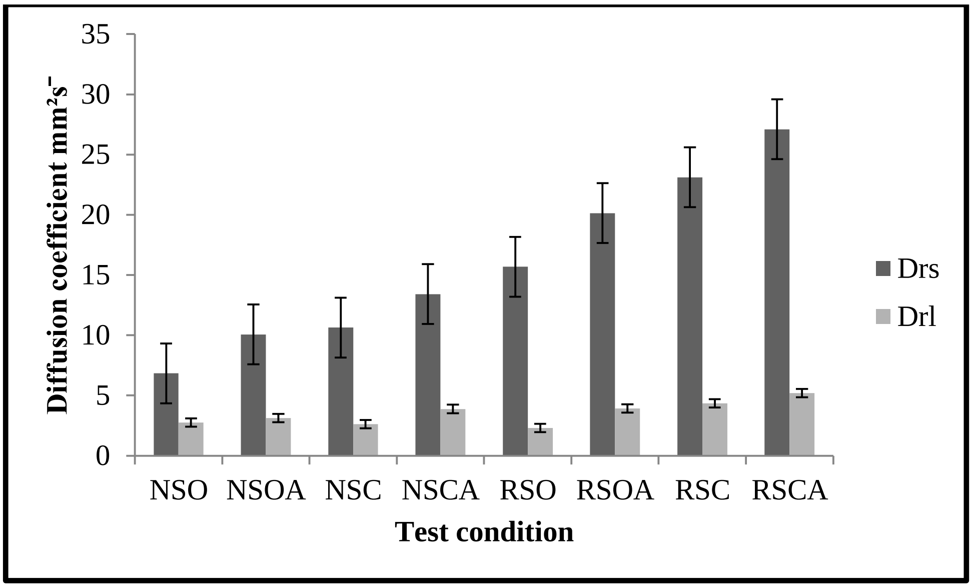


Figure 4-3 Mean and standard error of diffusion coefficients during normal stance and Romberg stance conditions with eyes open and closed with and without a listening task.

It was observed that some participants showed in general smaller diffusion coefficients compared to other participants and specifically at long-term diffusion coefficients where cases were recorded near zero. This was justified by (Collins and De Luca 1993) by the fact that after some small time, the COP no longer moved past a threshold. This threshold is limited to the area of support defined by a participant's feet, and hence the COP will be limited to that area, which means that the stabilogram would also saturate to that systematic value or threshold.

4.3.1.2 Scaling exponents

Scaling exponents in short-term regions were always greater than 0.5 under all test conditions, indicating persistence strategies and revealing a positively correlated posture behaviour. On the other hand, this SDA component was less than 0.5 in the long-term region in all test conditions, indicating anti-persistence strategies and revealing a negatively correlated posture behaviour (see Figure 4-4).

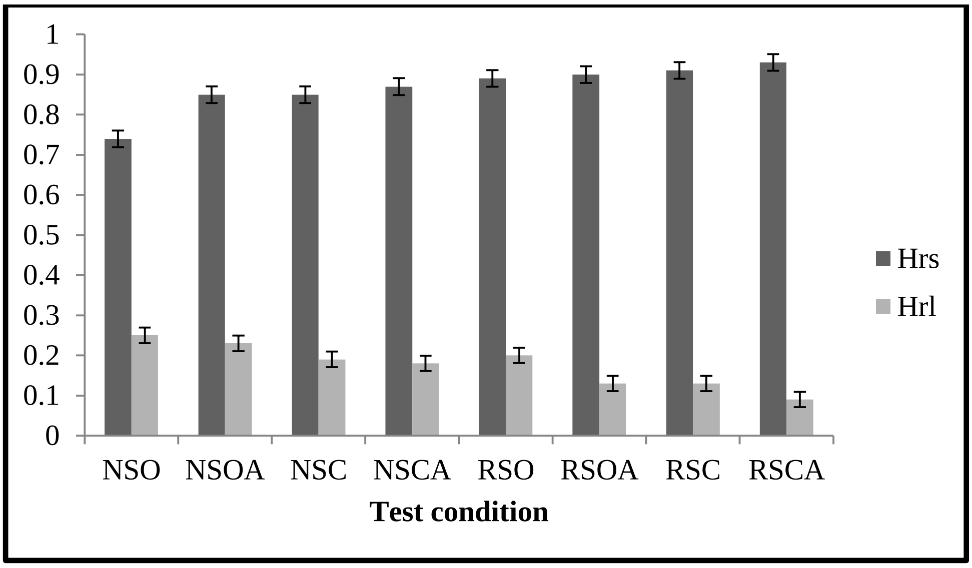


Figure 4-4 Scaling exponents under all test conditions for planar indicating Hrs always above 0.5 and Hrl below 0.5.

4.3.1.3 Critical point coordinates

Critical points, as indicated earlier, are defined by time intervals and mean square displacements. These critical points are of much interest here as they indicate the transition point when the participants shifted from open-loop strategies to closed-loop strategies. These points are, as indicated earlier, occurred after short times (within 1 second) and they provide valuable information regarding the strategies used to maintain upright posture over a certain period of time. Here, the critical points time intervals increased in dual-task conditions compared to the baselines, as shown in Figure 4-5 below.

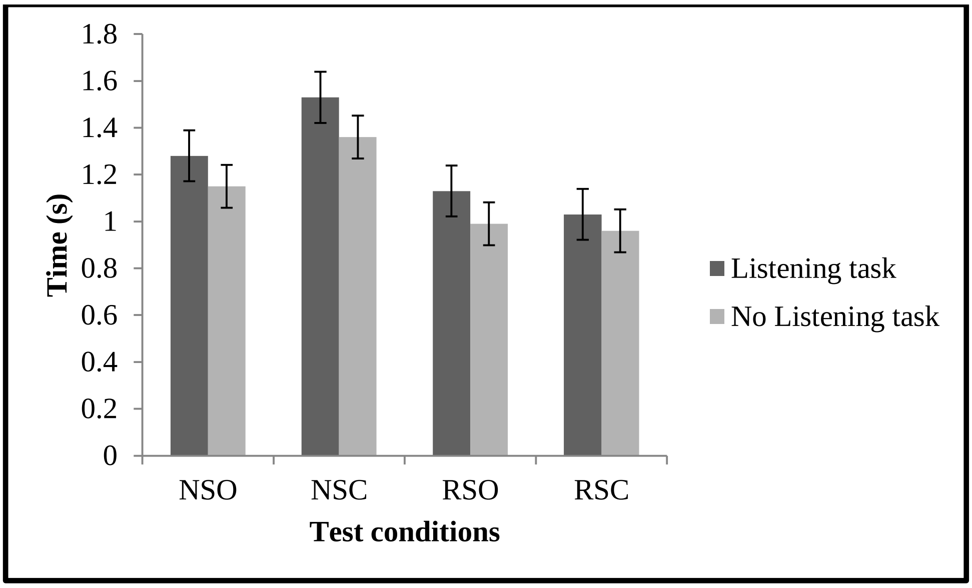


Figure 4-5 Critical time coordinates time intervals (s) for planar in all test conditions. Time intervals increased in dual-task conditions compared to the baselines.

Critical points mean square displacements showed the same behaviour. These mean square displacements were always larger during dual-task conditions Figure 4-6.

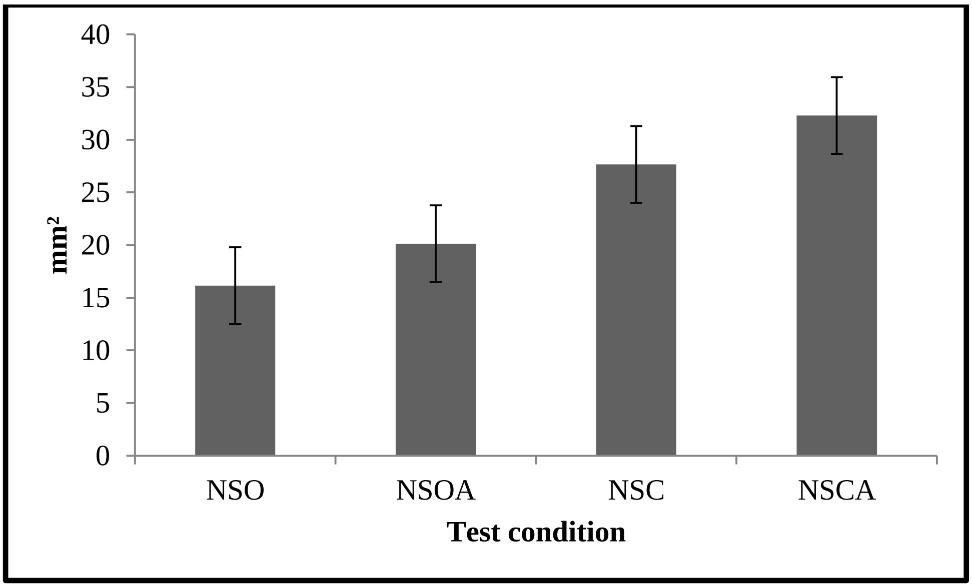


Figure 4-6 Critical points mean square displacements (mean and SE) in NS conditions

4.3.1.4 Listening task

For the listening task, average scores in dual-task conditions differ from their baselines, however, participants were able to score at least 71% correct in the most difficult dual task conditions. See Figure 4-7 where listening task average scores under the different posture task conditions are plotted.

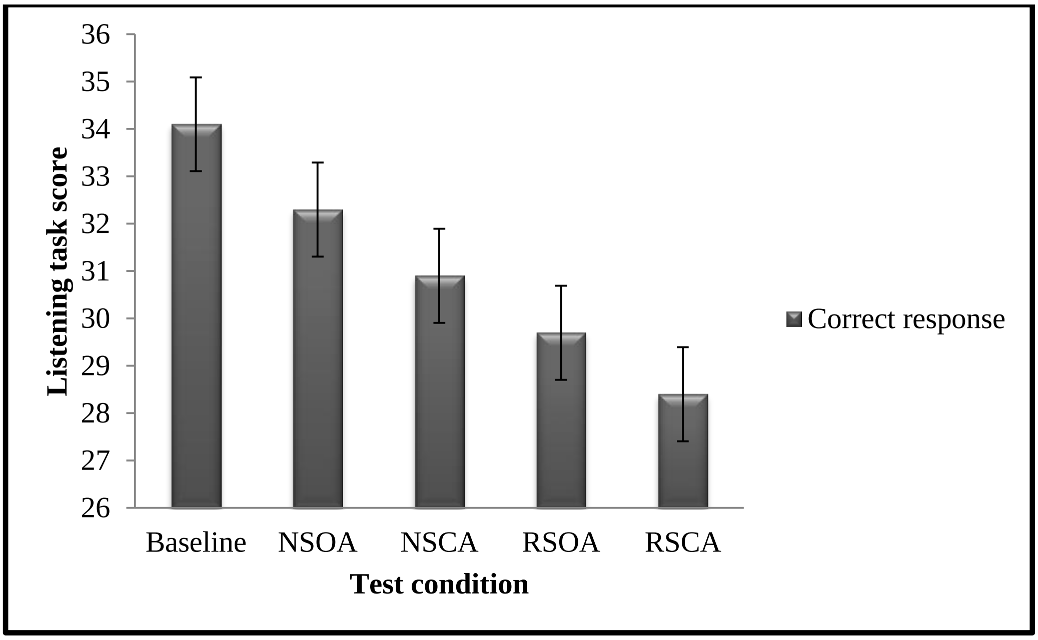


Figure 4-7 Listening task average scores under different test conditions.

The performance of the listening task deteriorated as the posture task difficulty increased, and this may be explained by a reallocation of attentional resource with increasing posture task complexity. Listening task scores under NSCA, RSOA and RSCA differed significantly from the baseline. Listening task scores in conditions under the same type of stance did not differ significantly from each other regardless eyes were either open or closed.

4.3.2 Analysis and results

In order to further examine these observations, a two-way repeated measures analysis of variance (ANOVA) with a 2×2×2 design was applied to the SDA data. The model was built based on two levels of the posture task (NS and RS), two levels of vision status (eyes open and eyes closed) and two levels of the listening task (no listening task and listening task). Test assumptions, i.e. normality and homogeneity of variance, were met.

Overall, there were main effects of listening, stance and vision for all diffusion plot measures but only *Hrs* showed an interaction between listening and posture and listening and vision. Significance was reported in (Table 4-1) and simple effects associated with it were examined using pairwise comparisons, which were conducted using a Bonferroni protected alpha level of $p = 0.05$. As there are three main measures, namely; the diffusion coefficient, scaling exponent and critical point coordinates, and each has short-term and long-term sub-measures for planar, the simple effects will be explored in separate sections.

Table (4-1): Summary of ANOVA results on the effect of listening on posture control for all calculated SDA measures showed main effects. F = F value, P = significance, MSE = mean square error and η_P^2 = partial eta squared. Effect degree of freedom = 1 and error degree of freedom = 31 for SDA measures. Alpha at 0.05.

Factor	**statistics**	**Drs**	**Drl**	**Hrl**	**Δtrc**	**Δr²c**
Posture	***F***	1304.72	6.36	18.22	61.08	429.40
	MSE	0.04	0.24	0.15	0.01	0.06
	P	**< 0.001**	**0.017**	**< 0.001**	**< 0.001**	**< 0.001**
	η_p^2	0.98	0.17	0.37	0.66	0.93
Vision	***F***	438.83	1.18	19.18	3.90	269.96
	MSE	0.025	0.26	0.12	0.01	0.05
	P	**< 0.001**	0.27	**< 0.001**	0.06	**< 0.001**
	η_p^2	0.94	0.04	0.38	0.11	0.90
Listening	***F***	29.52	9.27	11.81	9.96	12.49
	MSE	0.03	0.22	0.17	0.01	0.06
	P	**< 0.001**	**0.01**	**0.002**	**0.004**	**0.001**
	η_p^2	0.49	0.23	0.28	0.24	0.29

4.3.2.1 Planar short-term region diffusion coefficient (Drs)

The listening simple effects associated with the listening and posture and listening and vision main effects were analysed and presented in Table 4-2. Pairwise comparisons revealed that there is a significant difference between Drs measured during dual task conditions (performing a posture task concurrently with responding to a listening task) in both NS and RS positions with eyes either open or closed when compared to the baseline conditions (performing the posture task only). Drs always increased in dual task conditions.

Table (4-2) Short term diffusion coefficient (Drs). Simple effects associated with Listening main effects with posture and vision. NS = normal stance, RS = Romberg stance, EO = eyes open, EC = eyes closed, SE = standard error, F = F value, P-value = significance, η_P^2 = Partial eta squared

Factor	**Condition**	**Listening task**	**Drs Mean**	**Mean difference (2-1)**	**SE**	***F***	***P***	η_P^2
Posture	**NS**	**No listening task (1)**	1.14	0.11	0.03	17.15	**<0.001**	0.36
		Listening task (2)	1.25					
	RS	**No listening task (1)**	1.99	0.11	0.03	12.99	**0.001**	0.21
		Listening task (2)	2.11					
Vision	**EO**	**No listening task (1)**	1.36	0.10	0.04	7.85	**0.01**	0.20
		Listening task (2)	1.47					
	EC	**No listening task (1)**	1.77	0.12	0.03	14.64	**0.001**	0.32
		Listening task (2)	1.89					

4.3.2.2 Planar long-term region diffusion coefficient (Drl)

Listening simple effects are presented in Table 4-3. Comparisons revealed that Drl decreased significantly whilst dual tasking in the RS position and in eyes open conditions. This diffusion coefficient decrement indicates a less stochastic activity by the postural system under dual task conditions compared to their baseline conditions in the long term region when in tandem stance with eyes open.

Table (4-3) Long term diffusion coefficient (Drl) simple effects associated with Listening main effects with posture and vision. NS = normal stance, RS = Romberg stance, EO = eyes open, EC = eyes closed, SE = standard error, F = F value, P-value = significance, η_P^2 = partial eta squared

Factor	**Condition**	**Listening task**	**Drl Mean**	**Mean difference (2-1)**	**SE**	***F***	***P***	η_P^2
Posture	**NS**	**No listening task (1)**	0.34	-0.11	0.1	1.13	0.30	0.04
		Listening task (2)	0.23					
	RS	**No listening task (1)**	0.56	-0.25	0.01	0.25	**0.003**	0.25
		Listening task (2)	0.31					
Vision	**EO**	**No listening task (1)**	0.42	-0.11	0.08	6.39	**0.02**	0.17
		Listening task (2)	0.23					
	EC	**No listening task (1)**	0.47	-0.15	0.09	3.09	0.09	0.09
		Listening task (2)	0.32					

4.3.2.3 Planar short-term region scaling exponent (Hrs)

This measure showed a significant interaction between posture and listening and vision and listening Table 4-4.

Table (4-4): Summary of ANOVA results on the interaction between listening and posture and listening and vision for Hrs. F = F value, P = significance, MSE = mean square error and η_p^2 = partial eta squared. Effect degree of freedom = 1 and error degree of freedom = 31. Alpha at 0.05.

Interactions	**statistics**	**Hrs**
Listening*Posture	***F***	11.40
	MSE	0.002
	P	**0.002**
	η_p^2	0.27
Listening*Vision	***F***	13.99
	MSE	0.001
	P	**0.001**
	η_p^2	0.31

The simple effects associated with the interactions reported here are presented in Table 4-5. Further investigations to this interaction revealed that the presence of a listening task which requires participants to respond to, affects different stance positions differently. The interaction here indicates that Hrs in normal stance position differ significantly during dual task conditions and that 29.1% of the variance is accounted for by the listening task. Similarly the interaction between vision and listening suggests that the listening task affects Hrs differently depending on the availability of visual information. Here 39.2% of the variance is accounted for by listening during eyes open conditions.

Table (4-5) Short term scaling exponent (Hrs) simple effects associated with Listening main effects with posture and vision. NS = normal stance, RS = Romberg stance, EO = eyes open, EC = eyes closed, SE = standard error, F = F value, P-value = significance, η_P^2 = Eta partial squared

Factor	**Condition**	**Listening task**	**Hrs Mean**	**Mean difference (2-1)**	**SE**	***F***	***P***	η_P^2
Posture	**NS**	**No listening task (1)**	-0.10	0.04	0.01	12.72	**0.001**	0.29
		Listening task (2)	-0.07					
	RS	**No listening task (1)**	-0.04	-0.004	0.01	0.416	0.52	0.01
		Listening task (2)	-0.05					
Vision	**EO**	**No listening task (1)**	-0.09	0.03	0.01	20.02	**<0.001**	0.39
		Listening task (2)	-0.06					
	EC	**No listening task (1)**	-0.05	-0.002	0.01	0.046	0.083	0.001
		Listening task (2)	-0.05					

4.3.2.4 Planar long-term region scaling exponent (Hrl)

Simple effects of listening was explored to enable better understanding of the main effects Hrl revealed. These simple effects are presented in table (4-6). Hrl decreased significantly under dual task conditions when a posture task was concurrently performed with a listening task requiring response. This Hrl decrements were evident in RS position regardless eyes were open or close.

Table (4-6) long term scaling exponent (Hrl) simple effects associated with Listening main effects with posture and vision. NS = normal stance, RS = Romberg stance, EO = eyes open, EC = eyes closed, SE = standard error, F = F value, P-value = significance, η_P^2 = partial Eta squared

Factor	Condition	Listening task	Hrl Mean	Mean difference (2-1)	SE	*F*	*P*	η_P^2
Posture	**NS**	**No listening task (1)**	-0.75	-0.09	0.06	2.42	0.13	0.07
		Listening task (2)	-0.85					
	RS	**No listening task (1)**	-.88	-0.26	0.07	13.80	**0.001**	0.31
		Listening task (2)	-1.14					
Vision	**EO**	**No listening task (1)**	-0.72	-0.19	0.06	9.67	**0.004**	0.24
		Listening task (2)	-0.90					
	EC	**No listening task (1)**	-0.92	-0.16	0.08	4.32	**0.046**	0.12
		Listening task (2)	-1.08					

4.3.2.5 Planar time interval (Δtrc)

Simple effects were investigated here again to gain more insights about the significant change in planar time interval revealed by the listening main effects Table 4-7. Dual task conditions increased significantly Δtrc in NS positions with eyes open and closed. In normal stance conditions, 15.7% of the variance is accounted for by listening.

Table (4-7) planar time interval (Δtrc) simple effects associated with listening main effects with posture and vision. NS = normal stance, RS = Romberg stance, EO = eyes open, EC = eyes closed, SE = standard error, F = F value, P-value = significance, η_P^2 = partial Eta squared

Factor	**Condition**	**Listening task**	**Δtrc Mean**	**Mean difference (2-1)**	**SE**	***F***	***P***	η_P^2
Posture	**NS**	**No listening task (1)**	0.09	0.05	0.02	5.78	**0.02**	0.16
		Listening task (2)	0.13					
	RS	**No listening task (1)**	-0.01	0.03	0.02	3.06	0.09	0.09
		Listening task (2)	0.05					
Vision	**EO**	**No listening task (1)**	0.02	0.05	0.02	8.51	**0.007**	0.22
		Listening task (2)	0.07					
	EC	**No listening task (1)**	0.05	0.03	0.02	4.49	**0.042**	0.13
		Listening task (2)	0.08					

4.3.2.6 Planar mean square displacement < Δr^2c>

The main effect of listening and posture was investigated and results are provided in Table 4-8 below. Δr^2c increased in dual task conditions for both stance positions employed in this research either with eyes open or eyes closed. Actually this measure showed a steady increase as the postural task gets more challenging.

Table (4-8) planar mean square displacement (Δr^2c) simple effects associated with Listening main effects with posture and vision. NS = normal stance, RS = Romberg stance, EO = eyes open, EC = eyes closed, SE = standard error, F = F value, P-value = significance, η_P^2 = Eta partial squared

Factor	Condition	Listening task	Δr^2c Mean	Mean difference (2-1)	SE	*F*	*P*	η_P^2
Posture	**NS**	**No listening task (1)**	1.44	0.13	0.05	8.36	**0.007**	0.21
		Listening task (2)	1.57					
	RS	**No listening task (1)**	2.08	0.08	0.03	8.51	**0.007**	0.22
		Listening task (2)	2.16					
Vision	**EO**	**No listening task (1)**	1.54	0.09	0.04	6.41	**0.017**	0.17
		Listening task (2)	1.63					
	EC	**No listening task (1)**	1.97	0.12	0.04	8.14	**0.008**	0.21
		Listening task (2)	2.09					

4.3.3 Reliability of measures

As to the variation noticed between the behaviour of these SDA measures and also due to the fact that that the effect sizes calculated for them was not always of a satisfactory level, a reliability assessment of the measures using the data collected for this program of research is plausible in order to compare the reliability results calculated here with existing published SDA reliability literature in order to validate that what have been reported and concluded about the measures using the ANOVAs in the previous section.

In order to assess the reliability of the SDA measures, all measurements of every condition for every participant has to be retrieved. This is because, at this point of time, this research can only validate reliability by assessing reliability between the recordings of the same conditions. Intra-class correlation was performed for the NSO conditions recordings for the total number of subjects. The NSO, normal stance with eyes open, condition was chosen to facilitate comparisons with published literature. The coefficients obtained (Table 4-9) revealed comparable results with previously published work (Collins and De Luca 1993). The following table provide correlations coefficients of the SDA measures involved in this research.

Table 4-9 Correlation coefficients for all recoded trails under the NSO condition for all subjects.

SDA measure	Drs	Drl	Hrs	Hrl	Δtrc	Δr^2c
Correlation coefficient	0.87	0.59	0.91	0.88	0.71	0.91

4.4 Discussion

Stabilogram-diffusion analysis (SDA) is based on the assumption that the movement of the COP represents the combined output of co-existing deterministic and stochastic mechanisms. The stabilogram-diffusion analysis involves the extraction of three sets of posturographic parameters: diffusion coefficients, scaling exponents and critical point coordinates (Collins and De Luca 1993). The diffusion coefficient is an average measure of the stochastic activity of a random walker, i.e. it is directly related to its jump frequency and/or amplitude, and can be thought of as an indicator of the relative stability of the system (Doyle et al. 2008).

Quantification of the correlation between the step increments that make up an experimental time series is termed “scaling exponents” (Collins et al. 1995). This measure can provide an indication of whether the motion of the COP is more or less likely to continue moving in the same direction that it is currently moving in (Martin and Clark 2012). From a physiological standpoint, SDA scaling exponents quantify the correlated behaviour of the respective postural control mechanisms, i.e., short-term scaling exponents characterise the drift-like dynamics of the open-loop postural control mechanisms, whereas the long-term scaling exponents characterise the antidrift-like dynamics of the closed-loop postural control mechanisms (Collins and De Luca 1995b).

The critical point coordinates approximate the transition region that separates the short-term and long-term regions. The estimation of the critical point coordinates is determined by the intersection point of the straight lines fitted to the two regions of the linear-linear version of the resultant stabilogram-diffusion plot (Collins and De Luca 1993; Collins and De Luca 1995a; Collins et al. 1995). These coordinates approximate the temporal and spatial characteristics of the region over which the physiological postural control system switches from open-loop control to closed-loop control.

In the current investigation there were differences between the test conditions involving responding to a listening task, compared to conditions with no listening task, in critical point measures in all standing conditions. These results indicate that there is difference in the time point location when the postural control system switches from open-loop to closed-loop between listening and no listening conditions. In relation to the critical point coordination measures, the time point when the intersection occurred was progressively delayed as the complexity of the task being executed increased. For example, the critical point for the RSCA occurred at a longer time intervals when compared to the NSOA, NSCA and RSOA, respectively.

This time delay was interpreted as a result of introducing the listening task, which was then exacerbated by narrowing the base of support i.e. in RS conditions or EC conditions (Paulus, Straube and Brandt 1984a; Paulus et al. 1989; Paulus, Straube and Brandt 1984b; Ring, Nayak and Isaacs 1989) or both. This indicates that there is a significant increase in the stochastic activity of the COP whilst listening.

Although Collins and De Luca (1995a) investigated the effect of a different sensory input (visual input) on open-loop and closed-loop control mechanisms, the critical point coordinate findings of the current study are consistent with their results. Those authors found that there is a visual input status-related increase in critical point coordinate components, i.e. time interval and mean square displacement. They interpreted these findings as evidence of the visual-related modification in the temporal interaction of the open-loop and closed-loop control mechanisms. Thus according to the results of the current study, participants while dual tasking and responding to a listening stimuli may inadvertently allow the body segments to drift over larger displacements, this cannot be done for longer time periods before corrective feedback mechanisms are utilised. However, caution should be taken when comparing the findings of the current study with those of Collins and De Luca since the current study manipulated a sensory input in a different modality.

Generally there was a significantly higher diffusion coefficient in the dual-task conditions compared to baseline conditions. This difference between test conditions was mainly due to the increase in short-term diffusion coefficients seen in dual-task conditions when the participants were responding to the auditory stimuli. It may be hypothesised that due to responding to a listening task there may be an increased time delay in the sensing, transmission and processing, which may increase the average frequency of COP movement. Differences between dual-task conditions and baseline conditions were also noticed when considering scaling exponents measures. This result suggests that the behaviour of the open-loop postural control mechanisms while standing and concurrently responding to a listening task is different to standing only. This perhaps can be justified as there is a difference when responding to an auditory input, as there is a tendency to drift away from a relative equilibrium point over the short term, leading to higher scaling exponents.

The scaling exponents in the long-term region were also different whilst dual tasking; however they were always smaller than the short-term scaling exponents. This maybe suggests that the behaviour of the closed-loop postural control mechanism is more stable due to the more negatively correlated data, i.e. an increased probability that any movement away from a relative equilibrium point will be offset by corrective adjustments back towards the equilibrium position (Collins and De Luca 1995a). This tightening of the postural control system may be interpreted as a tightening of the system to offset the effects of the increased tendency to drift during the short-term period.

To conclude, all three parameters (diffusion coefficient, scaling exponent and critical time coordinates) over the short-term and the long-term regions were successful in detecting the effect of listening on posture control. The findings from this experiment support the findings from the previous chapter. It is now clear that in order to detect or examine the effect of listening on posture control, it is best to use a sway measure which is time dependant and calculated from both anterior-posterior and mediolateral directions i.e. mean velocity or diffusion plots.

5 Chapter Five:

Hearing Loss Simulation

5.1 Introduction

In this chapter, a technique is described in which the deliberate control of peripheral hearing loss can help concentrate on a certain level of hearing loss, and specifically identify whether sound distortion caused by peripheral auditory structure damage – relating to posture, in this case – is the reason for reduced performance in individuals with hearing impairments. A listening task was created using a simulation of SNHL, and was designed for use in normally hearing individuals who also took part in a posture task during the same period of research. A short overview of the methods available for use when simulating hearing loss is presented in this chapter, followed by a discussion of the advantages and disadvantages of each technique. A description of the simulation method selected for the study will be given, and then analysis of the method from subjective and objective standpoints will also be carried out in order to determine the appropriateness of the simulation for the experiment outlined in Chapter Six.

5.2 Hearing loss simulation

SNHL causes various supra-threshold effects within the auditory system, some of which are set out in Chapter Two (Section 2.4, ‘Perceptual consequences of hearing loss’). The predominant effects are reduced temporal resolution, reduced frequency selectivity, and loudness recruitment. All of these phenomena can happen at the same time along with reduced sensitivity in the auditory system, which consequently leads to higher absolute thresholds. A number of studies have sought to investigate supra-threshold complications obtained through hearing impairment induced by

SimHL (Moore and Glasberg 1993). The use of a simulation such as SimHL is helpful for the study of the psychoacoustic effects of SNHL; as long as the simulation is accurate, the effect can be observed in normal hearing individuals as each psychoacoustic factor related to hearing loss can be manipulated and studied separately (Baer and Moore 1993).

5.2.1 Advantages of using hearing loss simulation

There are several benefits associated with the simulation of hearing loss, including being able to customise the degree of hearing loss, controlling for cognitive aspects, and the ability to use between-subjects methodology which increases study validity. Although these factors are advantageous, it is worth bearing in mind that there are a number of limitations to simulation as well. Acknowledgement of such limitations is important, and the results obtained from studies using SimHL must also take this into account. The first drawback is that SimHL takes away the ecological validity of studies that apply it. This means that SimHL cannot capture any behavioural adaptations that a hearing impaired individual may have developed due to their hearing loss. Moreover, it should also be recognised that SimHL cannot capture all aspects of SNHL. It has been suggested that while the SNHL phenomena that SimHL is used to evaluate are the most important for speech understanding, it is important to consider the influence of those that are not covered (Moore 2007), 2007). The simulation of hearing loss does not emulate reduced temporal resolution precisely, and it cannot reproduce decrements in central auditory processing. These two factors can be problematic in terms of task performance, therefore any interpretation of the results must consider these points.

To this end, as long as the limitations are acknowledged and accounted for in the analysis of results drawn from studies using SimHL, this technique is considered the most appropriate for the continued work outlined in the current thesis. The rest of this chapter discusses the way in which simulation is accomplished, as well as the precise method selected for application in this study. The testing carried out on the simulation to ensure its suitable functioning is also discussed.

5.2.2 Procedure for simulating hearing loss

There have historically been two main procedures used in simulating the effects of particular types of hearing loss. The first is to use a filtered noise masker – this procedure masks the absolute thresholds of a normal ear to match those of an unmasked audiogram of a hearing impaired ear (Humes et al. 1987; Zurek and Delhorne 1987; Dubno and Schaefer 1992). The second procedure involves the application of Digital Signal Processing (DSP) techniques, which alter sound in such a way that it simulates how sound is perceived by an individual with a hearing impairment (Ter Keurs, Festen and Plomp 1992; Ter Keurs, Festen and Plomp 1993; Baer and Moore 1993; Baer and Moore 1994; Moore et al. 1997; Moore and Glasberg 1993; Nejime and Moore 1997).

5.2.2.1 Filtered noise masker

The application of a filtered noise masker masks the absolute thresholds of a normal ear to match those of an unmasked audiogram of a hearing impaired ear. According to Moore and Glasberg (1993), findings taken from normal ears using a noise masker are typically very similar to results obtained from unfiltered stimuli in impaired ears. Occasionally, however, individuals with actual SNHL demonstrate poorer results than those measured in the simulation. The authors suggest that this type of simulation can cause a loudness recruitment effect, but that this effect differs from that which occurs in SNHL due to it being of central rather than peripheral origin (Phillips 1987); moreover, the loudness recruitment remains confined to the smaller range of sound levels around the masked threshold (Stevens and Guirao 1967). Thus, this kind of simulation allows frequency-specific threshold elevation to be emulated, yet fails to effectively reproduce loudness recruitment or frequency selectivity (due to broadened auditory filters). Further, only mild to moderate hearing impairment can be simulated using a filtered noise masker, because simulating a more severe impairment would require the masking noise to be excessively loud, which could potentially damage a normal hearing ear as well as being uncomfortable for the participant (Baer and Moore 1993; Moore and Glasberg 1993).

5.2.2.2 Digital Signal Processing techniques

Digital signal processing (DSP) is commonly used to change sounds so that they more closely match those heard by individuals with hearing impairment. Various

DSP methods can be used to simulate particular effects associated with cochlear hearing impairment, namely loudness recruitment/threshold elevation and frequency selectivity. A summary of these techniques are provided in the following sections.

5.2.2.2.1 Frequency selectivity

Frequency selectivity is the ability to resolve all parts of a complex sound – an ability which is reduced in those with SNHL. The representation of sound in these individuals is less distinct, and this causes problems with auditory perception such as speech understanding (Moore 1996a). Reduced frequency selectivity is typically emulated by smoothing or smearing the frequency spectra of particular stimuli continually in order to mimic the experience of a hearing impaired ear in the cochlear excitation pattern of a normally hearing ear (Moore 2007). Several studies have used this approach to simulation in the assessment of the supra-threshold effects of reduced frequency selectivity, and have found that the results closely resemble those predicted from individuals with real hearing impairments (Ter Keurs, Festen and Plomp 1992; Ter Keurs, Festen and Plomp 1993; Baer and Moore 1993; Baer and Moore 1994).

The speech recognition thresholds (SRTs) were measured in participants exposed to smeared stimuli along with various kinds of masking noise (Ter Keurs, Festen and Plomp 1992; Ter Keurs, Festen and Plomp 1993). The results demonstrated that SRTs were higher using smeared speech when the simulated filter bandwidth was doubled or more, implying that the smearing paradigm affected speech intelligibility considerably. The results also showed that the increase in SRT persisted when the sound was a single competing speaker, but not for the speech-shaped sounds of processed stimuli, including SimHL. For normally hearing conditions, or unprocessed stimuli, if the masker was a single competing speaker the SRTs were reduced. This result is to be expected, because hearing impaired individuals are more affected by masking using a single speaker than those with normal hearing (Festen and Plomp 1990; Moore, Glasberg and Stone 1991; Peters, Moore and Baer 1998).

A more advanced version of the smearing technique was used by Baer and (Baer and Moore 1993), which allowed alterations to be made to the degree of asymmetry and broadening of auditory filter shapes. The findings showed that SRTs were progressively reduced through widening of the auditory filters when using a speech-shaped noise masker at low signal to noise ratios. A later study (Baer and Moore 1994) found the same results, which were also in accordance with findings from research by Ter Keurs, Festen and Plomp (1993). This study demonstrated that there was a smaller difference between speech intelligibility in single speaker and noise conditions for SimHL compared to unprocessed stimuli. Thus, it is reasonable to suggest that DSP provides a good approximation of the loss of frequency selectivity experienced by individuals with SNHL.

5.2.2.2.2 Loudness recruitment

Loudness recruitment is the abnormal increase in loudness that occurs with increased intensity of stimulation. In people with SNHL, this results in a reduced dynamic range. This means that the range between a sound being just audible and excessively loud is smaller for these individuals with SNHL than for those with normal hearing (Moore 1996b). DSP has been used to simulate loudness recruitment along with threshold elevation (Moore and Glasberg 1993; Duchnowski and Zurek 1995; Moore, Glasberg and Vickers 1995; Moore et al. 1997). In this method, a signal is split into a number of frequency bands that correlate with the basilar membrane auditory filters. Within each band, the range of levels is widened, and the bands are put together again to create a processed waveform. A more extensive range of hearing loss severities can be modelled using this technique with an accuracy of approximately 90 dB SPL (Moore and Glasberg 1993) (Moore and Glasberg, 1993), and the method does not require the use of masking sound which is therefore more comfortable for the participant.

An accurate loudness recruitment simulation is conducted through the use of a loudness model. The perceived loudness (in dB) for an individual with SNHL can be ascertained by asking a participant with unilateral hearing loss to match the loudness of a tone heard in their normal ear with a tone heard in their impaired ear. From this, a loudness model can be developed that can be applied to a normal hearing ear, which recreates the sensations of loudness experienced in an impaired ear (Moore 2007).

A previous study supports the validity of this method in simulating threshold elevation and loudness recruitment in the normal hearing ear. DSP was used in research carried out by Duchnowski and Zurek (1995) to evaluate how loudness recruitment and threshold elevation affect the perception of syllables heard in speech and in quiet. They set up the algorithm to correspond with the characteristics of individuals with hearing impairment who were tested in an earlier study (Zurek and Delhorne 1987), and the simulations were applied to normally hearing participants. Testing revealed that the pattern results from the normally hearing participants matched those of the hearing impaired participants from the earlier study. Thus, the implication is that the simulation reflects the subjective experiences of individuals with hearing impairment to a significant degree.

In another study, participants with unilateral moderately-severe sensorineural hearing loss were recruited and simulations of cochlear impairment were applied to their normally hearing ear (Moore et al. 1997). The majority of the participants agreed that the loudness and dynamics of the stimuli were appropriate. However, the participants indicated that the speech heard using the simulated stimuli was significantly clearer than for their impaired ear, and this was reflected in the results whereby performance in speech recognition was poorer when listening to the simulation in the impaired ear.

While part of the inconsistency in results can be attributed to the phenomenon of neglect, it is also the case that threshold elevation and loudness recruitment are not the only factors involved when hearing impaired individuals experience difficulties understanding speech. Instead, speech recognition is affected by a number of different factors associated with cochlear hearing loss. According to Moore (2007), existing data shows that audibility, frequency selectivity, and (to a lesser extent) loudness recruitment are the factors affecting speech recognition the most in individuals with moderate, severe and profound hearing losses. This means that if only one of these phenomena is simulated, the effect on the results is likely to be less than if all phenomena are applied at the same time.

5.2.2.2.3 SNHL simulation by Digital Signal Processing

A single DSP paradigm in one particular study made efforts to simulate the various elements involved in cochlear hearing loss - loudness recruitment, threshold elevation, diminished selectivity of frequency - and to determine how the ability to comprehend speech when noise is present (Nejime and Moore 1997). In this research, the simulation paradigm that was employed by the same research group was a concatenation of methodologies used in previous recreations of hearing loss (Baer and Moore 1993; Moore and Glasberg 1993). Because it acknowledges that the process of widening auditory filters is regulated by frequency (Faulkner, Rosen and Moore 1990), and that alterations in the selection of frequencies amongst people with impaired hearing are not as sharp, it is an advancement on the emulations that have taken place (Stelmachowicz et al. 1987; Murnane and Turner 1991). If the outcomes of a recreated condition of limited hearing loss are contrasted with a control condition, Nejime and Moore (1997) study can be considered the benchmark. Their findings echoed what might be predicted in a real-life instance of SNHL; to date, theirs is the only work that has recreated every factor of cochlear hearing loss concurrently.

5.2.2.3 Conclusion

To sum up, even though the outcomes from experiments involving DSP and noise-masking echo what may be found from an actual reduction in hearing, DSP seems to be a better method, as it can replicate some elements of cochlear hearing loss to a better level of accuracy than a noise-masker is able to, particularly in terms of selectivity in frequencies and loudness recruitment. DSP can also be applied across a greater breadth of degrees of hearing loss, from moderate to severe.

DSP's method of condition recreation has been deputed in the concurrent recreation of multiple elements of cochlear hearing loss, thereby imparting a more genuine depiction of SNHL than a stand-alone psychoacoustic event. However, it must be recorded that SimHL can only recreate SNHL to an extent, as there are some elements of impairment that it is not capable of emulating, such as the reduction in central auditory processing and temporal resolution. Conditions resulting from the auditory system being harmed, like the firing patterns of auditory nerves, are not able to be recreated by the use of DSP alone.

Although SimHL does not allow for these elements, there are features of SNHL that the methodology targets, such as decreased selectivity of frequency, elevation of thresholds and loudness recruitment, which are thought to be crucial in comprehending speech (Moore 2007); consequently, they are more probably the reason for the rise in the effort of listening needed amongst hearing impaired people. Such psychoacoustic properties can be reproduced extremely accurately, therefore if experiments involving simulation are conducted with people of normal hearing, the results will probably be very indicative of tests involving people who actually have hearing impairment. It must be recorded, however, that if conclusions are to be made from methodologies that use SimHL, the previously observed limitations must be taken into consideration.

5.3 The employed simulator

An account of the particular simulation method used in the work that this thesis describes is contained in this section. Amongst efforts to achieve a better way of recreating a total hearing loss, DSP displays better results than can be attained by using a filtered masking noise. Therefore, Nejime and Moore (1997) DSP approach was adopted through the use of MATLAB (2010). A concise overall picture is given in the next section of the simulation approach employed by Nejime and Moore (1997), followed by the testing took place to make sure that its validity and accuracy were relayed correctly.

5.3.1 Overview

SimHL's fundamental nature is its application of different DSP methods to an acoustic waveform, from which an audio file is obtained. This file is illustrative of how a person with SNHL would perceive the original form. The next section explains this process's essential stages, but to create a representation with the necessary level and form, there must be accurate specification of the different input parameters, including: 1) An audiogram of the hearing loss that is subject to simulation; 2) the calibration dB SPL, so as to introduce a loudness model that is founded on the input sound level; and 3) the output file's target dB SPL; it is up to the user as to whether the output file is to be subject to amplification or attenuation (apart from effects from hearing loss simulation) in connection with the input file's level.

5.3.2 How it functions

Three elements of SNHL are attempted to be emulated through this simulation approach: (1) threshold elevation, (2) loudness recruitment, and (3) decreased frequency selectivity. Two wide-ranging processing steps need to be taken in order to combine psychoacoustic phenomena: the first one uses a loudness model, thereby recreating loudness recruitment and elevating thresholds, while the second one engages a smearing application so as to reproduce the reduction of the selectivity of frequencies. To begin with, though, there must be appropriate preparation of the input file before there can be application of the two steps.

1. Preparation of sound file

Measurement of the RMS sound value inside the input file takes place; this happens primarily to offer a true picture of the intensity of the waveform across its lifespan, taking away the consequences of normal pressure fluctuations over a period of time (Rosen and Howell 2011). The input file's RMS value is then given a random dB value, denoted as the calibration dB SPL value. To the beginning of the file are added two calibration sounds: (1) a 520 Hz tone, and (2) a short bust of noise that is representative of speech frequencies (ANSI 1997). Production of each calibration noise is through the RMS value as the one deduced from the input file, resulting in both noises equalling the same level. In due course, these will be used so that the participants receive stimuli at the right level.

The next stage for the input file is to pass through a filter that recreates a sound's altered frequency attributes when it moves from an unrestricted field to the cochlea; the outer and middle ear elements involve the attenuation of sound at particular frequencies in relation to others prior to it arriving at the cochlea (Yost 2001). Such adjustments have been recorded by previous studies and are labelled as transfer functions; two of them are exemplified as: 1) A transfer function related to the head; this identifies a sound's acoustic alterations emanating from its being propagated during its passage through air, into the ear canal, and ending at the eardrum (Shaw 1974); (2) A transfer function of the middle ear, relating a sound's acoustic alterations that happen from its travel from the middle ear to the oval window that is positioned at the cochlea entrance (Killion 1978). When these two transfer functions are amalgamated, lower frequency attenuation occurs, while at mid-range frequencies there is a small increase.

2. Spectral smearing

Once there has been addition of the calibration sounds and application of the transfer functions' frequency stamp, smearing takes place, so as to recreate a lessening of the selectivity of frequencies. In order to work out the level of smearing that needs to be administered, the hearing thresholds between 2kHz and 8kHz are averaged. Any values equal to or above 57 dB HL are denoted as severely smeared, with moderate smearing referring to the range of 36 to 56 dB HL, mild smearing in the range of 16-35 dB HL, while anything below 15 dB HL is considered as not subject to smearing. The extent to which smearing occurs controls the level to which auditory filters widen: the higher extent of impairment, the larger the degree of widening, therefore the bigger the measure of smearing. There is separation of the input file into segments, with transference of every segment from the domain of time (pressure fluctuations over a period of time) to the domain of frequency (the proportional strength of the frequencies embedded in the waveform). For frames that progressively overlap, the simulation extracts the determined excitation pattern on the basilar membrane of whichever ear is subject to simulation. A definition of the excitation pattern may be stated as the auditory filters' output as an element of the frequency of the filter centre (Moore and Glasberg 1993).

As increasing frequency levels raise the bandwidths of auditory filters, the excitation patterns tilt upwards; there is a bigger excitation pattern at higher frequencies because when the filters are broader, they record more energy than filters that are narrower and encompass a smaller bandwidth. Removal of the tilt can be achieved through obtaining an estimate of power per hertz for each band of frequency. The reason for this to be done is that the simulated signal that ensues will move through a stack of auditory filters in experiments on people with normal hearing, which will singularly cause the excitation pattern to tilt.

After the higher-frequency tilt is removed, the excitation patterns that follow are subsequently divided into separate elements of power and phase; a smearing application is used to convolve the power, upon which the spectral elements are substituted with a weighted sum of the components surrounding it. The outcome of this is that the frequency spectrum is smoothed or smeared, thereby lowering the contrast between the heights and depths. There then follows an amalgamation of the original phase spectrum and the smeared spectrum, and a reversion of the signal from the frequency to the domain of time. Overlapping of the frames then occurs, and they are reconfigured to make a signal of identical length to the input signal, albeit with a spectrum that is now smeared.

3. Loudness model application

The input signal having been smeared, application of the recreated threshold elevation and loudness recruitment takes place. Passage of stimuli through a widened auditory filter bank, configured by the configurations and thresholds taken from measurements from people suffering hearing loss from moderate to severe, occurs. So as to generate minimum impact on sound’s spectral elements, (the spectral smearing algorithm having previously addressed and modified them), the sound’s envelope element is modified, as opposed to the entire waveform. The conveyance of the sound envelope over a period of time is by amplitude modulations, while the temporal fine structure concerns oscillations of extreme speed, whose rate is near to the auditory band's middle frequency (Moore 2008), gradually conveying frequency modulations (Loughlin and Tacer 1996; Stickney et al. 2004). In a general sense, the envelope contains data on a signal's amplitude/power, while data on the signal's frequency content is offered by the temporal fine structure. When reproducing loudness recruitment, therefore, modifications in the sound envelope are preferred.

The next emulation is loudness growth, which is achieved by increasing the signal's envelope for every auditory filter to the power N, which serves to magnify shifts in the envelope. If separate values of N for various filters are applied, loudness growth can be recreated in a way that is frequency-dependent, making allowances for precise recreation of loudness growth for those types of hearing loss that alternate over frequencies. Moore and Glasberg (1993), for instance, experimented with alternating values of N in separate frequency bands for hearing losses that were thus sloping and of high frequency.

An impaired ear's perception of loudness normally coincides with a normal ear at some point, which is normally thought of as 90 - 100 dB (Moore 2007). At points above this, loudness growth is not dissimilar between people with hearing impairment and normal hearing respectively. The waveform amplitude is therefore scaled to match the unprocessed sound's amplitude when the output level is the same as or larger than the catch-up point of loudness perception. If this scaling was not present, loudness would multiply to the same degree past the catch-up level. After the envelope has been processed, multiplication of loudness by each channel's fine structure occurs, resulting in a waveform that recreates senses of loudness associated with SNHL.

4. Output

The final step is to reverse both transfer functions that had been applied to the input file; this takes away the frequency effects. This is done so that the transfer functions can be applied when subjects listen to the stimuli. The simulation process is thus complete and the resulting waveform, together with the appended calibration sounds, reflects the psychoacoustic phenomenon of loudness recruitment, threshold elevation, and reduced frequency selectivity. In theory, this waveform represents to a person with normal hearing how a person with SNHL would perceive a sound.

5.4 Validation of the simulation

If any significant conclusions are to be drawn from the results of this study, based on this simulation as the source of hearing loss, it is essential that the validity and accuracy of the simulation are evaluated. The following section describes the processes of objectively and subjectively testing the simulation so that its suitability for the research can be determined.

5.4.1 Objective analysis

Two types of objective analysis were conducted: (1) making sure that the loudness model of the simulation was an accurate reflection of real hearing loss, and (2) making sure that independent levels of loudness recruitment and threshold elevation were utilised across the various frequency bands.

5.4.1.1 Obtaining the Loudness model

The reduction in level of a two-second burst of white noise, processed using the hearing loss simulation, in relation to a reference sound, was determined for various input levels (0–120 dB in 5 dB increments), together with varying levels of flat hearing loss (the same Pure Tone Audiometry (PTA) thresholds across all frequencies; 10–90 dB in 10 dB increments). The following formula was used to calculate the reduction in level from the reference file:

$$dB = 20\ log10\ (\frac{V}{V\ ref})$$

where V = the RMS absolute voltage of the sound processed with the hearing loss

and Vref = an RMS reference voltage, for comparison with V. Vref was calculated using the RMS value of the white noise processed with a SimHL of 0 dB HL for each of the employed frequencies. For instance, if the white noise was used in the simulation for a normal hearing and hearing loss condition with an input SPL of 60 dB, and RMS voltage values of 0.8 and 0.4, the alteration in dB relating to the waveform in the hearing loss condition is as follows:

$$20\ log10\ \left(\frac{0.4}{0.8}\right) = \ -6.02\ dB$$

The dB level in the hearing loss condition is therefore 54 dB (60 dB - 6.02 dB).

Similar calculations were carried out for several different severities of hearing loss across a wide range of input sound intensities. This allowed the researcher to produce a loudness model, which was examined to determine whether abnormal loudness growth had been represented accurately in the simulation for different intensities of input.

Expected results were obtained from the simulation with regard to loudness recruitment and threshold elevation, over different severities of hearing loss (see Figure 5-1). At the threshold level, the dB values produced were accurate to within ±3 dB. For all of the different levels of hearing loss, when sounds were close to the threshold they crossed the X-axis at 0 dB – this signified that there was no sound in the wave file, so it could not be heard. For example, with a flat hearing loss of 20 dB, an input sound with an intensity of 20 dB could not be heard as the output sound file was 0 dB.

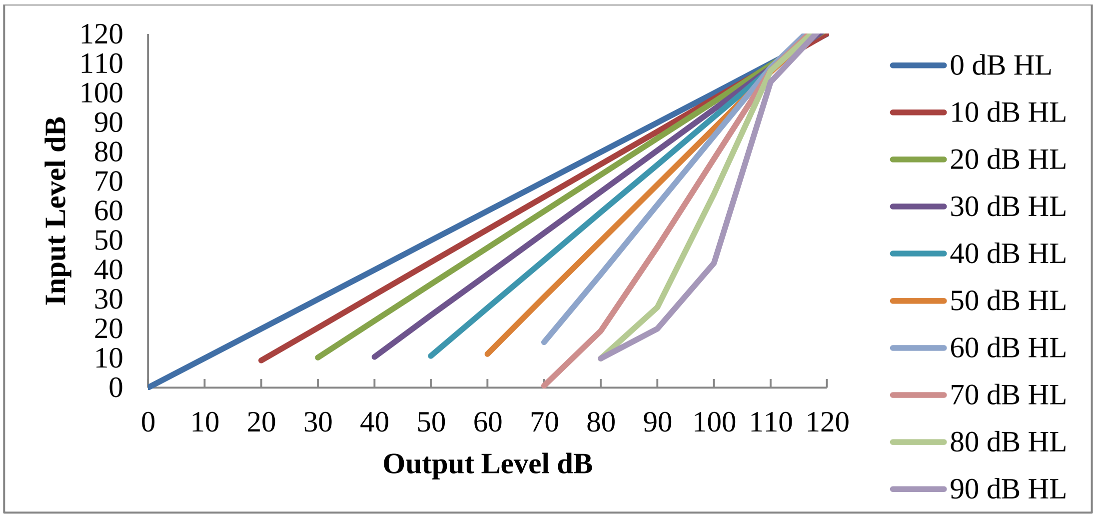


Figure 5-1 Estimation of the loudness growth model used by the simulation software for varying degrees of hearing loss. Each line represents the loudness growth for a given flat hearing loss across all frequencies.

At levels over the absolute threshold point, loudness grew abnormally, then it reached a convergence point, before growing normally. For most levels of hearing loss, this convergence point was 100 dB, although for very severe hearing loss the pattern was not so predictable. In fact, with the most severe level of simulated hearing loss (flat 90 dB) there was no linear loudness growth. Furthermore, inaccuracies were seen in the representation of loudness perceptions at levels close to the threshold. It was therefore decided to exclude hearing losses above 80 dB from any of the experiments where this simulation method was employed.

The gradient of the loudness growth slopes in this research were consistent with those reported by Moore and Glasberg (1993). For instance, for a hearing loss of 50 dB, the loudness growth gradient was 2, signifying that loudness grew twice as quickly as the reference condition, i.e. normal hearing. For mild hearing loss, the gradient was less, and it was more for severe hearing loss; this is predicted by models of loudness perception for SNHL individuals (Moore and Glasberg 1993). The objective testing therefore demonstrated that as far as loudness recruitment and threshold elevation are concerned, the simulation was highly accurate for a wide range of hearing losses and input sound pressure levels.

5.4.1.2 Different thresholds across frequencies

Despite the fact that the results of the simulation were as expected with regard to loudness recruitment and threshold elevation, this was the case with simple flat hearing losses across a range of frequencies. However, the validation tests have not clarified whether the SimHL method is capable of accurately simulating threshold elevations in the case of hearing loss with different absolute thresholds across frequencies. In order to evaluate this element of the hearing loss simulation, a broadband noise was produced under various hearing loss conditions.

The test stimulus was two seconds of white noise, with an equal amount of energy at each of the frequencies. The input level and desired output level of the created sound was set to 80 dB SPL. Two conditions were created to simulate hypothetical hearing losses, correlating to different levels of sensory decline. For each condition, the test signal was run through the simulation, and the output was converted to the frequency domain. The magnitude spectra were examined to determine the degree to which they accurately reflected the simulated hearing losses.

It is worth noting that it was not expected that there would be an exact match between the frequency magnitude of the processed signals and the audiogram; rather, when the input levels were near the threshold it was expected that the perceived loudness would be approximately 0 dB. It was also predicted that at higher sensation levels loudness would grow abnormally, therefore the simulation output levels should have exceeded the levels predicted by a linear relationship with the hearing loss magnitude. This expectation was indeed fulfilled. The frequency spectra that were obtained were accurate reflections of the changes in frequency thresholds observed in each respective audiogram. The results also revealed the effect of loudness recruitment – at frequencies where thresholds were under 80 dB SPL level of the input sound, the frequency spectrum followed the audiogram, but their gradients were different.

This analysis shows that the simulation method is accurate for sloped hearing losses. As expected, in every case where the 80 dB input sound attained the threshold at higher frequencies, the relative size of the frequency response was zero or very near to it. Where thresholds at higher frequencies were reduced, but did not exceed the input sound’s level, the relative magnitude of the frequency response curve did not reach zero. This signals that at these frequencies the sound should still be audible, but at a reduced loudness. Furthermore, the different gradients of the frequency response and the absolute hearing thresholds illustrates that abnormal loudness growth occurred in different frequency bands. Together with the results from flat hearing losses, these results confirm that the simulation was accurate with regard to its ability to accurately reproduce the modelled effects of loudness recruitment and threshold elevation occasioned by SNHL, independent of the configuration of the hearing loss.

5.4.2 Subjective analysis

Although it is important to determine the simulation’s accuracy theoretically, the subjective experience of the SimHL must also be representative of real SNHL. Speech testing was therefore conducted so that the ecological validity of the output from the simulation could be assessed. With simulated hearing loss, a person with normal hearing should produce a speech audiogram that is similar to that expected from someone with real SNHL. Speech testing can therefore prove or disprove the subjective validity of the research simulation. This will fulfil two purposes: providing insight into the ecological validity of simulated hearing loss, and providing further data to determine whether a simulation behaves as expected with regard to threshold elevation – in order to be heard, sounds will have to be louder. In addition, this study will provide information on how successfully the simulation can smear stimuli in an attempt to copy the lower frequency selectivity that comes with SNHL. The smearing paradigm may distort sounds to such an extent that they cannot be understood by some people with a certain level of hearing loss, even when they are sufficiently loud to be heard.

Speech testing is commonly employed to determine how well an individual can comprehend conversational speech (Bess and Humes 2003). Words or phrases are played to the subjects at a certain sound level, and then the subjects repeat them.

Each stimulus is awarded a mark based on the accuracy of the repetition. These marks are added together and calculated as a percentage correct score. This is then plotted as a function of the stimulus presentation level, which results in a speech audiogram. The audiogram is typically S-shaped, although the type of speech material utilised will determine the exact form of the shape (Figure 5-2).

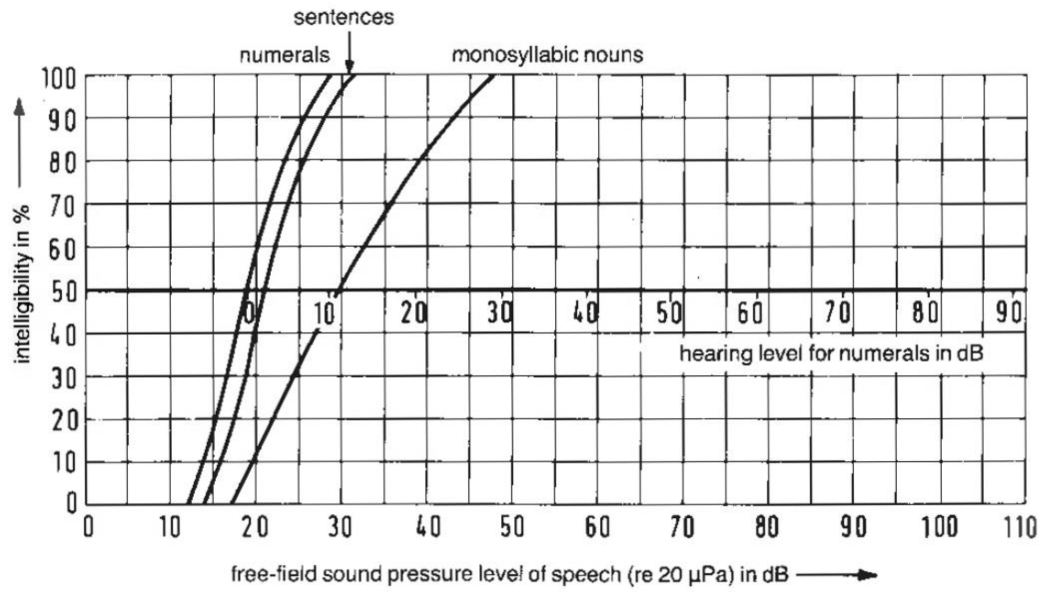


Figure 5-2 Typical speech audiogram results obtained from using different types of speech stimuli. Source: (Martin and Clark 1997).

As hearing impairment increases, the S shape tends to move positively along the X-axis. At some levels of SNHL, the end of the S will tail-off, giving the impression that it is not possible to achieve a 100% speech recognition (Graham and Baguley, 2009). For conductive hearing loss this is not so, as the distortion caused by cochlear damage is not present (Figure 5-3).

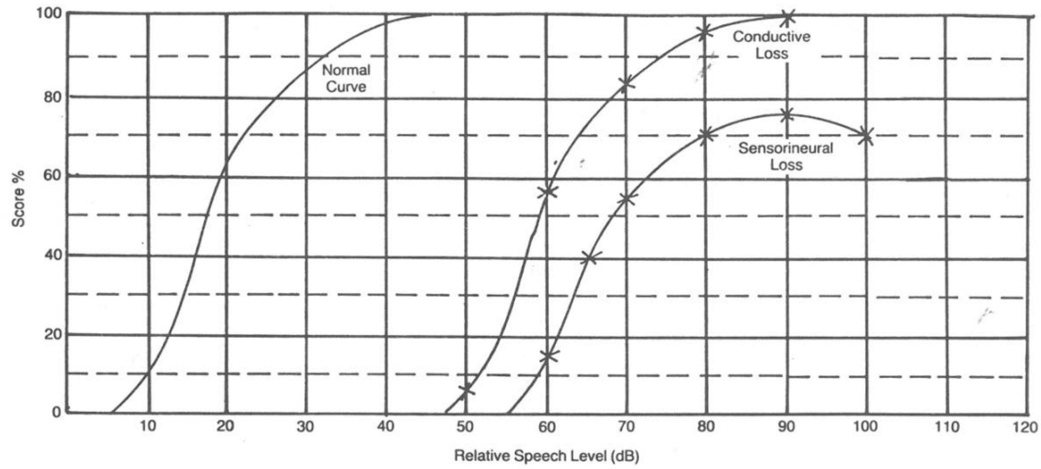


Figure 5-3 Speech audiogram results using sentences as stimuli. Source: (Martin and Clark 1997).

5.4.2.1 Method

5.4.2.1.1 Participants

Subjects for speech testing were English speakers (6 females and 6 males). They were aged between 20 and 28 years old (M = 23.08, S.D. = 2.54). Pure tone audiometry testing was carried out for all participants (as per the British Society of Audiology, 2011), and they were confirmed to have bilateral normal hearing. All of their single absolute thresholds were below 15 dB HL.

5.4.2.1.2 Materials

Arthur Boothroyd word lists (Boothroyd 1968) were utilised, and they were presented to the subjects through Tele-phonics TDH-39P headphones. Three auditory conditions were used: (1) a reference condition with no hearing loss; (2) a simulated mild to moderate hearing loss condition; and (3) a simulated moderate to severe hearing loss condition. The latter two conditions reflected cases of increasing loss of sensitivity with frequency (see table 5-1 for thresholds).

All of the stimuli used were passed through the simulation with different calibration dB SPL values, corresponding to the level at which they would be presented during the speech test. In this way, it was ensured that the equipment was correctly calibrated and that the levels of stimuli would be accurately presented. Stimuli levels were altered relative to a reference sound file, ensuring that the loudness relationships were preserved. Instead of just moving the dial setting on the audiometer, stimuli level changes were pre-empted and made in the simulation environment. This helped to ensure the accuracy of the loudness growth model and the level of smearing.

Table 5-1 levels of the simulated mild to moderate and moderate to severe hearing losses.

Frequency	**Mild to moderate HL (dB)**	**Moderate to severe HL (dB0**
125 Hz	15	30
250 Hz	20	30
500 Hz	20	30
750 Hz	30	40
1 kHz	35	50
1.5 kHz	40	60
2 kHz	40	70
3 kHz	45	70
4 kHz	45	80
6 kHz	55	80
8 kHz	60	80
10 kHz	60	80
12 kHz	60	80
14 kHz	60	80
16 kHz	60	80

5.4.2.1.3 Procedure

The subjects were placed in a sound-proofed booth to take all of the speech tests. The order in which the conditions were presented was varied so that all of the possible permutations occurred equally. The participants repeated all of the words they were able to hear, even if they were unsure what the word was, or had only partially heard the word – in that case they were asked to say what they thought the word was.

For each input intensity, the word lists were randomly varied, in order to make sure that running the same word lists at specific presentation levels had no impact. Each word contained three phonemes and each correct phoneme scored one point – there were ten words so the maximum possible score was thirty at each level. The score was transformed to a percentage correct figure and plotted in dB as a function of presentation level.

5.4.2.2 Results

The patterns observed for each participant were similar to those expected to be seen in individuals with a hearing impairment. One way of assessing the success of a simulation is whether the test results move positively along the X-axis as hearing impairment increases. If this is the case, it confirms that the simulation is able to accurately copy threshold elevation – the words need to be made louder for the subjects to hear them clearly. In fact, for the majority of the participants, the amount by which the graph moved along the X-axis was close to the simulated hearing loss when this was 500–1000 Hz. Therefore we can say that the simulation gave all of the participants a similar subjective experience of SNHL, regardless of the small variations in their baseline absolute thresholds across different frequencies (Figure 5-4).

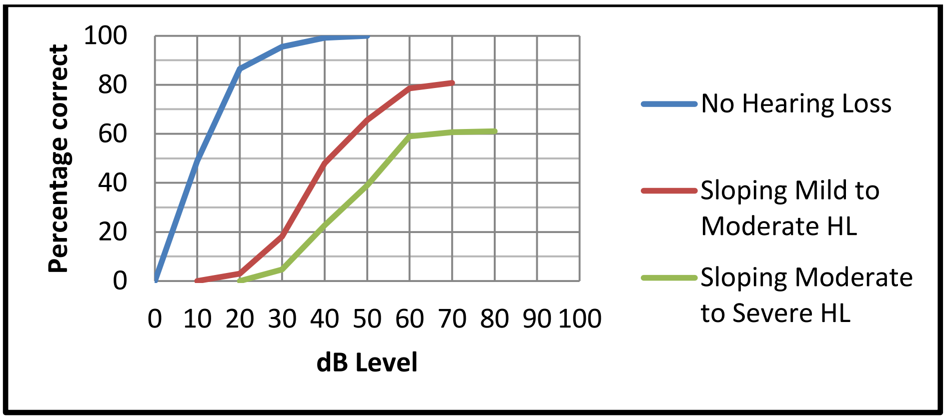


Figure 4-3 Speech audiogram illustrating average speech recognition scores at different hearing loss levels.

Another aspect that indicated the success of the simulator in subjectively emulating SNHL was that it reduced the recognition of speech, even when the loudness was uncomfortably high. None of the participants could tolerate stimuli that were presented to them at more than 90 dB HL, and with simulated moderate hearing loss, none of them scored more than 90% for any of the word lists at any level. It was also observed that for some of the participants at high dial settings, the percentage correct scores rolled off for the mild and moderate conditions. This occurs in actual test data and it is believed to be due to the distortion caused by hearing loss at loud input levels (Martin and Clark 1997).

The distortion apparent in SimHL also affected the gradient of the speech test curve. For example, each time the dial setting was increased by 10 dB HL, performance improved more for the normal condition than for the mild and moderate conditions. When the speech test results were averaged, the approximate gradients of the speech test curves decreased with increasing hearing loss. Other studies have reported the same phenomenon (Martin and Clark 1997), and it implies that not just the attenuation of sound but other aspects of hearing loss influence people's ability to recognise speech. Again, this advocates the successful emulation of reduced frequency selectivity through spectral smearing, which, together with loudness recruitment and threshold elevation was the only aspect of SNHL simulated in this research.

The ecological validity of the simulation has been supported by these subjective results, which also show the accuracy of the simulation in reproducing the perceptual consequences of SNHL. The results also confirm that the DSP produced stimuli subjectively similar to the conditions experienced by someone with actual cochlear hearing impairment.

5.5 Summary

This research, in particular the work presented in Chapter 6, supports the use of hearing loss simulation, while highlighting its relative advantages and disadvantages. The testing of the simulation method used in this study shows that it is highly accurate and valid. It can accurately reproduce threshold elevation for various hearing loss severities (up to 90 dB HL) related to specific frequencies. In a manner akin to real loudness perception models, the simulation method used here illustrated loudness recruitment across a variation of input intensities. The loudness recruitment model is correlated reliably with the severity of hearing impairment, and changes in loudness recruitment can be applied to particular frequency bands. Subjectively, the simulation method ensures that speech stimuli have to be made louder by a relative level before they are perceived. Moreover, for both the normal hearing and simulated hearing loss conditions, a given increase in level does not result in a comparable increase in intelligibility. Speech is not entirely intelligible, even at clearly audible levels, as a result of the frequency smearing applied to the stimuli.

The simulation is therefore objectively sound, and it is capable of producing results in people with normal hearing that correlate to the results that would be obtained from people with actual SNHL. The benefits of this study's method include the rise in experimental power and control, together with its ability to take away extraneous variables such as cognitive capabilities. The following chapter examines how the SimHL employed in this research was used to examine the impact of hearing loss on posture. This research will provide valuable information on the extent to which peripheral hearing loss increases the amount of effort required for listening, and it should therefore be able to be applied to other tasks that are being performed concurrently.

6 Chapter Six:

Simulated Hearing Loss Affects Posture

6.1 Introduction

In Chapter Three it was demonstrated that when healthy adults with normal hearing actively carried out a listening task with a concurrent postural task, body sway increased. This finding was further explored by studying the changes in stochastic processes involved in posture control in Chapter Four. This work indicated higher posture control stochastic processes when a posture task and a listening task are executed simultaneously. Results from both chapters indicated that listening has a destabilising effect on posture control. The previous chapters gave some insight into the cognitive processes that may be involved in exacting a dual-task cost and these have been explained in the terms of a cognitive framework previously used to describe dual-task performance in other sensory domains, derived from Baddeley's WM (Baddeley 2012). The research presented in this chapter extends these ideas to explore the hypothesis that hearing-impaired people will find an auditory task more effortful than their hearing counterparts and will use a larger proportion of available cognitive/WM resources, leaving less processing resources available for a posture task.

It was hypothesised that if listening compromised the attentional resources available to execute postural tasks in healthy adults with normal hearing, this effect is expected to be more prominent in cases of hearing loss. This assumption was made based on the idea that listening requires attention and demands mental processing (Mishra et al. 2010; Rönnberg et al. 2013a; Rudner et al. 2011), and the idea that listening in more adverse situations, such as listening in background noise (Mattys et al. 2012; Bronkhorst 2015), with hearing loss (Stenfelt and Rönnberg 2009) or when the cognitive demands of listening are increased (Mattys et al. 2012), will require even greater attention and hence more cognitive processing demands.

As mentioned previously in this thesis, the aim of this project was to set the scene for this concept or observation of the attentional cost of performing a listening task on posture control in adults with normal hearing and with hearing loss. It was thought, however, that examining this observation using simulated hearing loss at this stage instead of recruiting adults with real hearing loss would be more beneficial in understanding the matter (See chapter Five). By investigating this matter in normal adults with simulated hearing loss, the project will be in control of other confounding factors such as ageing, reduced cognitive skills, decreased central processing efficiency, emotional state i.e. depression and fatigue, impaired memory or ability to learn new tasks which are reported to be caused by or correlated to hearing loss. Using simulated hearing loss will also enable the ability to set specific hearing loss levels and guarantee that all participants are exposed to the same levels. Simulated hearing loss will ensure that the only component of the hearing loss is peripheral. This approach will help validate whether the effect is also true for this specific group and assist in setting protocols for future research aiming to recruit individuals with actual hearing loss.

In this chapter, it is hypothesised that the evident destabilising effect of listening on posture mediated by attention, seen in Chapter Three and Chapter Four, will be larger in the presence of hearing loss. It is thought that the attentional resources required to maintain erect posture will be compromised by the increased attentional demand to respond to a listening stimulus when combined with hearing loss.

6.2 Methods

6.2.1 Ethical approval

An ethical approval from the School of Healthcare Research Ethics Committee (SHREC) was obtained for this study (SHREC/RP/296), (Appendix 12).

6.2.2 Inclusion and exclusion criteria

As already noted before, the inclusion criteria were: age between 18 and 60 years, normal middle ear function (ear pressure from -500 to +500 pa, compliance between 0.3 to 1.6 cm^3 and ear canal volume between 0.6 to 1.5 cm^3), normal hearing bilaterally (hearing thresholds better than or equal to 20 dB HL across octave frequencies between 250 and 8 kHz), no personal history of balance problems, and English as the first language. Any participant who did not meet the inclusion criteria, reported neurological disorders, motor problems, severe visual impairments or unable to understand and follow verbal instructions was excluded from the study. Participants were advised to visit their GPs if abnormal hearing thresholds were noted.

6.2.3 Participants

For a power level of 85%, alpha = 0.05 and r = 0.50, Cohen's tables indicated the need of Thirty-two participants (Cohen 1988). 32 Thirty-two normal healthy participants, 14 males and 18 females aged 21 – 56 years old (mean age 37.6 SD± 14.2) met the inclusion criteria and were included in the study, none of them participated in the previous experiments. Participants were students and staff recruited from the university community, and local residents in Leeds. The study was advertised using posters inside the university campus and around Leeds city, by circulating emails to university staff and students, and by snowball strategy. Written informed consent was obtained from all participants after oral information, printed information and the opportunity to ask questions had been given. Participants were given gift vouchers to recompense them for their time.

6.2.4 Equipment and listening task

6.2.4.1 Screening equipment

A calibrated Interacoustics MT10 middle ear analyser was used to perform tympanometry testing using a 226 Hz probe tone to establish middle ear function and a calibrated FONIX Hearing evaluator, FA-12 Digital Audiometer was used to conduct audiometry to evaluate hearing thresholds.

6.2.4.2 Experimental equipment

For balance evaluation, COP data were collected using a Kistler force plate type 9286BA (600x400x35mm) and a data Acquisition Box, amplifier, (Kistler DAQ Type 5691A) controlled by a SAMSUNG laptop running proprietary software (BioWare v5.1.1.0). Using the laptop and the analysis software, BioWare, these data are controlled and processed to give visual and statistical data of ground reaction forces, moments and most importantly, COP (see section 2.2.4.2 for details).

6.2.4.3 Listening stimuli

A listening task with simulated hearing losses of different levels was used in this experiment. These stimuli were developed as described in Chapter Five.

6.2.5 Procedure

The procedure conducted in this experiment followed the procedure in Experiment One in Chapter Two, apart from using a different listening stimuli. The tests were carried out in the audiology laboratory at the School of Healthcare at the University of Leeds. Every participant underwent a screening which involved obtaining medical history (see Appendix Seven) followed by a routine clinical audiological assessment. After screening, subjects who met inclusion criteria were tested in the dual-task experiment. Participants were asked to complete three 30-second trials recorded with a sampling rate of 100 Hz. Mean velocity and diffusion plots measures were calculated from the raw COP data (see section 3.2.5 for the detailed procedure).

6.3 Data and Results

6.3.1 Data

6.3.1.1 Mean velocity

Mean velocity showed an increased body sway with increasing the difficulty of the listening task and the difficulty of the balance task. Mean sway scores for MV in each test condition are provided in Table 6-1.

Table 6-1 Mean velocity (mm/s) mean scores for each standing condition (NSO normal stance eyes open, NSC normal stance eyes closed, RSO Romberg stance eyes open and RSC Romberg stance eyes closed) under different listening situations (No task no listening "baseline", No HL listening task presented with no simulated hearing loss, MM HL listening task presented with mild to moderate hearing loss and MS HL listening task presented with moderate to severe hearing loss).

MV mean scores	**Listening Condition**			
Standing Condition	**No Task**	**No HL**	**MM HL**	**MS HL**
NSO	12.1	12.9	13.5	13.9
NSC	14.8	15.3	15.7	16.9
RSO	27.1	27.9	28.6	29.6
RSC	52.1	52.9	53.9	56.3

From the table provided above it is noted that the easiest standing condition, NSO, showed the least body sway mean velocity, and the hardest standing condition, RSC, showed the greatest body sway mean velocity, with the NSC and RSO conditions recording body sway mean velocity in between. It is also evident that body sway mean velocity increased as the difficulty of the listening task increased from the no listening task condition to the listening task condition simulated with moderate to severe hearing loss.

6.3.1.2 SDA measures

For the diffusion plots measures, diffusion coefficients in the short-term region were larger than their respective long-term diffusion coefficients for all test conditions. Diffusion coefficients in the short-term region were generally greater in dual-task conditions (NSOA, NSCA, RSOA and RSCA) where participants were performing a posture task concurrently with a listening task when compared to the their baseline conditions (NSO, NSC, RSO and RSC), where they were performing a posture task alone. Generally the harder the listening task was, the larger the diffusion coefficients were.

Scaling exponents in short-term regions were always greater than 0.5 under all test conditions, indicating persistence strategies revealing a positively correlated posture behaviour. On the other hand, this SDA component was less than 0.5 in the long-term region in all test conditions, indicating anti-persistence strategies revealing a negatively correlated posture behaviour. It was also noted that, every time the listening task was introduced, the scaling exponent in the short-term region increased when compared to the baseline conditions, and that the scaling exponent increased with increasing posture and listening task difficulty.

The critical point coordinates: time intervals and mean square displacements, which indicate the transition point where participants shifted from open-loop strategies to closed-loop strategies, increased in dual-task conditions compared to their baselines.

6.3.2 Results

To further examine the observations reported above, mean velocity and diffusion plots measures data were analysed using repeated measures analysis of variance (ANOVA) with 2×2×4 design. The model was built based on two levels of the balance task (NS and RS), two levels of vision (eyes open and eyes closed) and four levels of the listening task (no listening task, listening task with no simulated hearing loss, listening task with a simulated mild to moderate hearing loss and listening task with a simulated moderate to severe hearing loss). Test assumptions related to normality, homogeneity of variance and sphericity were met.

6.3.2.1 Mean velocity

Mean velocity showed a significant interaction between listening and vision only. No interaction between listening and posture was noted, $\boldsymbol{F}_{(3,\ 285)} = 0.862$, MSE = 0.003 $\boldsymbol{p} = 0.461$, $\eta_p^2 = 0.026$.

Table 6 -2: Summary of ANOVA results on the interaction between listening and vision for MV. F = F value, P = significance, MSE = mean square error and η_P^2 = partial eta squared. Effect degree of freedom = 1 and error degree of freedom = 285. Alpha at 0.05.

Interactions	**statistics**	**Hrs**
Listening*Vision	***F***	4.78
	MSE	0.003
	P	**0.003**
	$\boldsymbol{\eta_p^2}$	0.129

Inspection of the interaction between listening and vision revealed that under eyes open conditions, the way the hearing loss affects MV sway measure was significantly dependent on the hearing loss level in four out of six comparisons. However, under eyes closed conditions this was evident in only one case were the listening task processed with no hearing loss was compared to the task processed with moderate to severe hearing loss. Table 6-3 below illustrates these comparisons in more details.

Table 6-3 Summary of ANOVA results on the interaction between listening and vision for MV. F = F value, P = significance, MSE = mean square error and η_P^2 = partial eta squared. Effect degree of freedom = 1 and error degree of freedom = 93. Alpha at 0.05.

Factor	**Condition**	**Listening task**	**MV Mean**	**Mean difference (2-1)**	**SE**	***F***	***P***	η_P^2
Vision	**EO**	**No listening task (1)**	1.26	0.04	0.04	16.43	**<0.001**	0.35
		Listening task with no loss (2)	1.29					
	EC	**No listening task (1)**	1.45	0.01	0.01	5.99	1.00	0.16
		Listening task with no loss (2)	1.46					
	EO	**No listening task (1)**	1.26	0.02	0.01	16.43	**0.004**	0.35
		Listening task with MMHL (2)	1.27					
	EC	**No listening task (1)**	1.45	-0.00	0.01	5.99	1.00	0.16
		Listening task with MMHL (2)	1.44					
	EO	**No listening task (1)**	1.26	0.01	0.01	16.43	0.36	0.35
		Listening task with MSHL (2)	1.27					
	EC	**No listening task (1)**	1.45	-0.01	0.001	5.99	0.2	0.16
		Listening task	1.44					

		with MSHL (2)						
	EO	**Listening task with no loss (1)**	1.29	-0.02	0.01	16.43	**0.002**	0.35
		Listening task with MMHL (2)	1.27					
	EC	**Listening task with no loss (1)**	1.46	-0.01	0.01	5.99	0.17	0.16
		Listening task with MMHL (2)	1.44					
	EO	**Listening task with no loss (1)**	1.29	-0.03	0.01	16.43	**<0.001**	0.35
		Listening task with MSHL (2)	1.27					
	EC	**Listening task with no loss (1)**	1.46	-0.02	0.01	5.99	**0.001**	0.16
		Listening task with MSHL (2)	1.44					
	EO	**Listening task with MMHL (1)**	1.27	-0.01	0.00	16.43	0.48	0.35
		Listening task with MSHL (2)	1.27					
	EC	**Listening task with MMHL (1)**	1.44	-0.01	0.01	5.99	0.55	0.16
		Listening task with MSHL (2)	1.44					

6.3.2.2 SDA results

There were main effects of listening, stance and vision for diffusion plot measures (Table 6-4). Simple effects associated with listening main effects were examined using pairwise comparisons, which were conducted using a Bonferroni protected alpha level of $p = 0.05$. To facilitate the follow of this section, simple effects will be explored in separate sections.

Table (6-4): Summary of ANOVA results on the effect of listening on posture control for all calculated SDA measures showed main effects. F = F value, P = significance, MSE = mean square error and η_p^2 = partial eta squared. Effect degree of freedom = 1 and error degree of freedom = 95 for SDA measures. Alpha at 0.05.

Factor	statistics	Drs	Drl	Hrs	Hrl	Δtrc	Δr²c
Posture	***F***	1294.21	6.46	62.67	16.22	59.14	433.51
	MSE	0.04	0.27	0.03	0.14	0.02	0.06
	P	**< 0.001**	**0.014**	**< 0.001**	**< 0.001**	**< 0.001**	**0.016**
	η_p^2	0.97	0.18	0.65	0.39	0.68	0.93
Vision	***F***	432.63	1.19	30.81	20.18	3.88	273.10
	MSE	0.03	0.28	0.21	0.15	0.01	0.01
	P	**< 0.001**	0.28	**< 0.001**	**< 0.001**	**< 0.001**	0.27
	η_p^2	0.94	0.03	0.03	0.36	0.14	0.97
Listening	***F***	30.52	9.37	6.51	12.81	10.21	13.19
	MSE	0.03	0.25	0.01	0.16	0.01	0.06
	P	**< 0.001**	**0.004**	**0.013**	**0.003**	**< 0.001**	**0.003**
	η_p^2	0.49	0.26	0.31	0.26	0.26	0.29

6.3.2.2.1 Planar short-term region diffusion coefficient (Drs)

The listening simple effects associated with the listening and posture and listening and vision main effects were analysed and presented in Table 6-5. Pairwise comparisons revealed that there is a significant difference between Drs measured during dual task conditions (performing a posture task concurrently with responding to a listening task) in both NS and RS positions with eyes either open or closed when compared to the baseline conditions (performing the posture task only). Drs always increased in dual task conditions.

Table (6-5) Short term diffusion coefficient (Drs). Simple effects associated with Listening main effects with posture and vision. NS = normal stance, RS = Romberg stance, EO = eyes open, EC = eyes closed, SE = standard error, F = F value, P-value = significance, η_P^2 = Partial eta squared

Listening task	Mean difference (2-1)	SE	*F*	*P*	η_P^2
No task (1) Vs No SimHL (2)	0.02	0.003	10.13	1.00	0.16
No task (1) Vs SimMMHL (2)	0.11	0.03	17.54	**<0.001**	0.29
No task (1) Vs SimMSHL (2)	0.11	0.03	11.85	**<0.001**	0.30
No SimHL (1) Vs SimMMHL (2)	0.01	0.003	9.64	0.74	0.14
No SimHL (1) Vs SimMSHL (2)	0.02	0.02	13.21	**<0.001**	0.29
SimMMHL (1) Vs SimMSHL (2)	0.02	0.01	12.27	**<0.001**	0.32

6.3.2.2.2 Planar long-term region diffusion coefficient (Drl)

This measure did not show any significance under different listening conditions when standing in the NS. However, when standing in the RS Drl showed significance difference under some listening conditions. Listening simple effects are presented in Table 6-6. This diffusion coefficient indicates a less stochastic activity by the postural system under dual task conditions compared to their baseline conditions in the long term region when in tandem stance.

Table (6-6) Long term diffusion coefficient (Drl) simple effects associated with Listening main effects with posture and vision. NS = normal stance, RS = Romberg stance, EO = eyes open, EC = eyes closed, SE = standard error, F = F value, P-value = significance, η_P^2 = partial eta squared

Listening task	Mean difference (2-1)	SE	*F*	*P*	η_P^2
No task (1) Vs No SimHL (2)	0.01	0.004	9.54	0.08	0.11
No task (1) Vs SimMMHL (2)	0.25	0.08	12.99	**0.002**	0.28
No task (1) Vs SimMSHL (2)	0.31	0.09	13.83	**<0.001**	0.36
No SimHL (1) Vs SimMMHL (2)	0.01	0.01	8.67	0.07	0.09
No SimHL (1) Vs SimMSHL (2)	0.01	0.01	10.21	1.00	0.12
SimMMHL (1) Vs SimMSHL (2)	0.09	0.003	11.34	0.06	0.17

6.3.2.2.3 Planar short-term region scaling exponent (Hrs)

Investigations of simple effects for Hrs revealed that the varying levels of the listening task affects different stance positions and conditions differently according to the difficulty of both the listening task and the posture task. The simple effects associated with listening main effect are presented in Table 6-7.

Table (6-7) Short term scaling exponent (Hrs) simple effects associated with Listening main effects with posture and vision. NS = normal stance, RS = Romberg stance, EO = eyes open, EC = eyes closed, SE = standard error, F = F value, P-value = significance, η_P^2 = Eta partial squared

Listening task	Mean difference (2-1)	SE	*F*	*P*	η_P^2
No task (1) Vs No SimHL (2)	0.004	0.01	7.15	0.07	0.36
No task (1) Vs SimMMHL (2)	0.04	0.01	14.57	**<0.001**	0.32
No task (1) Vs SimMSHL (2)	0.05	0.01	13.81	**0.008**	0.34
No SimHL (1) Vs SimMMHL (2)	0.01	0.002	6.61	0.12	0.02
No SimHL (1) Vs SimMSHL (2)	0.02	0.01	12.97	**<0.001**	0.29
SimMMHL (1) Vs SimMSHL (2)	0.02	0.01	13.02	**<0.001**	0.26

6.3.2.2.4 Planar long-term region scaling exponent (Hrl)

Hrl significantly differ under dual task conditions when a posture task was concurrently performed with a listening task of varying HL levels. Simple effects of listening was explored and are presented in table (6-8) below. Exploring this measure revealed that when in NS postural performance changed under dual task conditions but was not statically significant, however, in RS changes in postural performance were statistically significant regardless eyes were open or closed when a listening task of varying HL levels were being executed concurrently with a postural task.

Table (6-8) long term scaling exponent (Hrl) simple effects associated with Listening main effects with posture and vision. NS = normal stance, RS = Romberg stance, EO = eyes open, EC = eyes closed, SE = standard error, F = F value, P-value = significance, η_P^2 = partial Eta squared

Listening task	**Mean difference (2-1)**	**SE**	***F***	***P***	η_P^2
No task (1) Vs No SimHL (2)	0.01	0.02	8.91	0.10	0.14
No task (1) Vs SimMMHL (2)	0.24	0.07	27.87	**<0.001**	0.36
No task (1) Vs SimMSHL (2)	0.29	0.09	29.17	**<0.001**	0.39
No SimHL (1) Vs SimMMHL (2)	0.02	0.01	10.51	0.06	0.13
No SimHL (1) Vs SimMSHL (2)	0.17	0.01	16.19	**<0.001**	0.02
SimMMHL (1) Vs SimMSHL (2)	0.27	0.02	21.13	**0.02**	0.02

6.3.2.2.5 Planar time interval (Δtrc)

Simple effects were investigated to gain more insights about the significant change in planar time interval revealed by the listening main effects, Table 6-9. Dual task conditions increased Δtrc significantly regardless of the nature of the stance position and the difficulty of the listening task.

Table (6-9) planar time interval (Δtrc) simple effects associated with listening main effects with posture and vision. NS = normal stance, RS = Romberg stance, EO = eyes open, EC = eyes closed, SE = standard error, F = F value, P-value = significance, η_P^2 = partial Eta squared

Listening task	Mean difference (2-1)	SE	*F*	*P*	η_P^2
No task (1) Vs No SimHL (2)	0.04	0.02	5.58	**0.03**	0.12
No task (1) Vs SimMMHL (2)	0.05	0.02	9.56	**0.003**	0.22
No task (1) Vs SimMSHL (2)	0.05	0.03	6.85	**0.02**	0.16
No SimHL (1) Vs SimMMHL (2)	0.03	0.02	3.01	0.09	0.06
No SimHL (1) Vs SimMSHL (2)	0.05	0.03	8.14	**0.001**	0.27
SimMMHL (1) Vs SimMSHL (2)	0.02	0.01	4.74	0.09	0.04

6.3.2.2.6 Planar mean square displacement < Δr^2c>

The simple effects associated with the main effects of listening and posture was investigated and results are provided in Table 4-8. Δr^2c increased in dual task conditions for both stance positions employed in this research either with eyes open or eyes closed. Actually this measure showed a steady increase as the postural task gets more challenging.

Table (6-10) planar mean square displacement (Δr^2c) simple effects associated with Listening main effects with posture and vision. NS = normal stance, RS = Romberg stance, EO = eyes open, EC = eyes closed, SE = standard error, F = F value, P-value = significance, η_P^2 = Eta partial squared

Listening task	**Mean difference (2-1)**	**SE**	***F***	***P***	η_P^2
No task (1) Vs No SimHL (2)	0.10	0.09	6.34	0.29	0.04
No task (1) Vs SimMMHL (2)	0.23	0.004	8.99	**0.04**	0.30
No task (1) Vs SimMSHL (2)	0.25	0.08	9.85	**<0.001**	0.36
No SimHL (1) Vs SimMMHL (2)	0.09	0.04	7.01	**0.09**	0.17
No SimHL (1) Vs SimMSHL (2)	0.17	0.05	9.21	**0.004**	0.22
SimMMHL (1) Vs SimMSHL (2)	0.13	0.05	9.61	**0.005**	0.29

6.4 Discussion

This study investigated the extent to which carrying out a listening task under simulated hearing loss conditions impacted on postural control performance whilst adopting NS and RS with eyes open and eyes closed. The present study proposes that the increased listening effort required to perform the auditory task in the presence of a simulated hearing loss will lead to a concurrent decrease in performance on the posture task. If this is the case, it would suggest that aspects of posture attention are disturbed under some listening conditions.

In the present study, the addition of a listening task resulted in minimal changes in body sway when standing in a well-learned position such as NS with eyes open, indicating that this stance condition may only require minimal amount of attention. In the same type of stance but with the eyes closed, it was noted that the addition of the listening task resulted in changes in body sway. Furthermore, when the posture stance task difficulty was increased by introducing the RS with eyes open and eyes closed respectively, the addition of the listening task resulted in greater body sway.

The body sway mean velocity was again able to show the effect of listening on posture control as it did in Experiment One in Chapter Three of this thesis. In this experiment, mean velocity was not only able to detect the effect of listening on posture, but it was also able to reveal this effect at varying levels of hearing loss. This finding validate the results from experiment one regarding MV and also added that this measure is sensitive to detect the effect of listening on posture under varying level of simulated hearing losses. In the literature, mean velocity was reported to be a sensitive body sway measure, able to reflect modifications in posture control either due to the type and difficulty of the posture position and condition (Doyle et al. 2008), or due to the addition of a cognitive task (Moghadam et al. 2011), i.e. listening in this study. Although MV was reported to be sensitive, the novelty in this line of research is that it was identified out of many to be sensitive to measure the effect of listening under normal conditions on posture and also has the power to detect this effect under different levels of simulated hearing loss. MV was also able to distinguish between MMHL and MSHL.

The diffusion plots components calculated in the present study, critical point coordinates, diffusion coefficient and scaling exponents, revealed differences between test conditions of varying levels of task difficulty, either in listening or posture. Critical point measures showed differences between test conditions involved responding to a listening task simulated with different hearing loss levels when compared to conditions with no listening task or task with no simulated hearing loss. These results indicate that there is difference in the time point location when the postural control system switches from open-loop to closed-loop. In relation to the critical point coordination measures, the time point where the intersection occurred was progressively delayed as the complexity of the task being executed increased. For example, the critical point for the conditions RSC with MSHL occurred at a longer time intervals when compared to the rest of the conditions.

This progressive time delay of the critical point is thought to be the consequence of performing a gradually difficult listening task. This also reflects the increasing difficulty of the posture task, which exacerbated this delay, as the task gets harder by narrowing the base of support, i.e. in RS conditions or EC conditions or both (Paulus, Straube and Brandt 1984a; Paulus et al. 1989; Paulus, Straube and Brandt 1984b; Ring, Nayak and Isaacs 1989). This indicates that there is a significant increase in the stochastic activity of the COP whilst listening under simulated hearing loss conditions. According to the results of the current study, participants while dual tasking and responding to a listening stimuli with simulated hearing loss may allow larger body sway displacements before corrective feedback mechanisms are utilised.

Generally there was a significantly higher diffusion coefficient in the dual-task conditions compared to baseline conditions. This difference between test conditions was mainly due to the increase in short-term diffusion coefficient seen in dual-task conditions when participants were responding to the listening task with simulated moderate to severe hearing loss (MSHL). It may be hypothesised that due to the increased difficulty of the listening task there may be an increased time delay in sensing, transmission and processing, which may increase the average frequency of COP movement.

Differences between dual-task conditions and baseline conditions, especially under the MMHL and MSHL conditions, were detected when considering scaling exponents measures. This result suggests that the behaviour of the open-loop postural control mechanisms, while standing and concurrently responding to a listening task simulated with MMHL or MSHL, is different to standing only without listening and also different to standing and responding to a listening stimuli with no simulated hearing loss. The scaling exponents in the long-term region were also different whilst dual tasking; however, they were always smaller than the short-term scaling exponents.

It is also worth mentioning that although generally body sway increased with increasing strength of hearing loss but this was not evident for every condition. Actually there was some conditions where body sway decreased with stronger levels of simulated hearing loss. Based on the research executed here, the relationship between the strength of hearing loss and the amplitude of body sway cannot be judged to be linear.

To conclude, mean velocity and all the three diffusion plots parameters, diffusion coefficient, scaling exponent and critical time coordinates, were successful in detecting the effect of listening on posture control. Not only that, but they were also able to detect this effect at varying levels of simulated hearing loss. The findings from this experiment support the findings from the previous chapters, as it is clear now that in order to detect or examine the effect of listening on posture control, it is best to use a time-dependant sway measure, i.e. mean velocity or diffusion plots.

7 Chapter Seven:

General Discussion and Conclusions

7.1 General Discussion

The aim of this thesis was to explore the effect of listening on posture control, to understand how individuals modify or change the strategies they use to maintain an erect posture while listening, and finally to explore whether the presence of simulated hearing impairment would make a difference to this relationship between listening and posture. This thesis, unlike previous research, has indirectly looked at the cognitive cost of listening, not simply hearing, on posture control. As noted earlier, listening involves more processing as it requires attending to the task, comprehending and responding. Hearing on the other hand is a passive process and requires much less attention and effort. In this section, the results from the experimental chapters will be discussed in order to answer the overarching question of this thesis and also the questions these results have raised will be reported.

The findings from the first experiment, reported in Chapter Three, demonstrated that listening has a destabilising effect on posture control and that MV was able to detect this effect. This influence was revealed by studying the participants' postural sway while dual tasking, compared to their baselines. In this experiment, nine body sway measures were employed to evaluate posture performance. Existing studies in the literature have used these body sway measures to evaluate posture control under different test conditions (Doyle et al. 2008; Jeka et al. 2004; Moghadam et al. 2011; Gray, Ivanova and Garland 2014; Santos et al. 2008; Karlsson and Frykberg 2000). This large number of body sway measures was calculated because it was difficult for this study to decide which measure might detect the effect of listening on posture, due to the vast number of body sway measures used to investigate posture control in the previous literature and also due to discrepancies in previous results (Bauer et al. 2008; Demura, Kitabayashi and Aoki 2008; Doyle, Newton and Burnett 2005; Mancini et al. 2012; Raymakers, Samson and Verhaar 2005). Moreover, the research into the relationship between listening and posture is under researched and that there is no evidence of which parameter(s) are likely to be most sensitive. Therefore, these measures were calculated, with the aim of representing as far as possible the different postural sway characteristics, such as body sway amplitude, velocity, path and area.

Out of these measures, only one was sensitive to the effect of listening on posture control, and that was mean velocity (MV). It might have been the only sway measure sensitive to postural sway changes in the presence of a simultaneous listening task due to the nature of the way it is calculated. MV is calculated based on the changes observed in sway displacement over time in the AP and ML directions. This is a novel finding as, to the knowledge of the researcher, MV was not precisely reported to be linked to listening or that it is able to detect its effect on posture. This finding is relevant both for future research concerning posture and listening and for the direction of the subsequent experiments carried out in this thesis. Now we know which postural sway measure to calculate, for how long, and for how many trials.

The second experiment presented in Chapter Four shed light onto what is happening when participants are concurrently listening and maintaining an upright posture. The results of this experiment revealed that the participants adopted different postural strategies during dual tasking, compared to their single task performance, and hence they provide some evidence that listening alters the strategy to maintain stability. The ability of the SDA to provide insights regarding the relationship between listening/sound and posture in this line of research is supported by previous literature which reported SDA analysis beneficial in studying posture under different circumstances or when studying different populations and/or health conditions (Collins and De Luca 1993; Collins and De Luca 1994; Collins and De Luca 1995b). In this research, and when the listening task was introduced, participants utilised open-loop strategies for an extended time, compared to the baseline when they were performing a single task. The time that they required to shift from open-loop strategies to closed-loop strategies was longer, indicating that they needed longer to process the new information before shifting to closed-loop mechanisms. Although they have examined a different type of input, i.e. visual, the results from this study are similar to the results of Collins and De Luca (1995a) when they examined the influence of a visual input on open-loop and closed-loop postural control mechanisms.

The findings in this second experiment also suggest an extended listening effort in order to attend to the listening stimuli. It was proposed that increased listening effort required more processing resources, which compromised the resources available to execute the postural tasks, especially as the postural task difficulty increased. The results supported the notion that increased body sway reflects a shift of the attention resources towards the listening task. The errors on the listening task on the other hand, especially in the difficult standing positions, reflect that as the postural task difficulty increased, the attention resources were reallocated to accommodate the posture task, and as a result the listening task performance deteriorated. This suggests that posture comes first over listening, when standing in unstable positions that might lead to falls (Bloem et al. 2001; Grabiner and Troy 2005; Yogev-Seligmann, Hausdorff and Giladi 2008).

These findings are consistent with the literature, as a large body of research reports that posture comes first in risky standing positions, especially in older adults (Lindenberger, Marsiske and Baltes 2000). However, some research has revealed that young adults show a ‘cognition first’ behaviour (Bernard-Demanze et al. 2009); this might have been different in this study as the participants’ mean age was higher (36.5 years) compared to the 28 years in Bernard-Demanze et al. (2009) study. Moreover, the Romberg position with eyes closed introduces a more challenging posture position (Black et al. 1982) which might force the participant to adopt ‘posture first’ strategies. This might also be because the listening task was difficult, with competing words, so even if they wanted to do much better on it, they essentially were unable to because they had an overwhelming urge to avoid losing their ability to maintain their upright posture, which would lead to falling.

Chapter six presented a third experiment, the aim of which was to evaluate how simulated hearing loss would influence postural control. The results obtained under conditions of simulated hearing loss were compared to those obtained under normal hearing conditions. It was thought that using simulated hearing loss would control for confounding factors such as ageing (Davis 1995) and cognitive decline (Baldwin and Ash 2011; Foo et al. 2007; Humes 2007; Lunner and Sundewall-Thorén 2007), and would also control for the hearing impairment being only peripheral. Using a simulated hearing loss would indicate that the effect was purely due to peripheral impairment, and if that were true, it was expected that individuals with a true/real hearing loss would show at least similar results, if not worse, as a result of other consequences of hearing loss such as poor processing efficiency (Lunner and Sundewall-Thorén 2007). The results show that the performance of a simultaneous listening task of varying difficulty degraded performance on posture tasks and that the more complex the posture task became, the greater the effect of the listening task. Moreover, as the level of the simulated hearing loss increased, the more the performance deteriorated on both the posture task and the listening task.

In terms of the exacerbating effect of simulated hearing loss, the effect of the concurrent auditory task on posture progressively increased as the thresholds of the simulated hearing loss elevated and as the difficulty of the posture task increased. The deterioration in postural performance during easy posture positions and listening task conditions might be explained by the idea that attentional resources devoted to the execution of the posture task were withdrawn to perform the listening task. However, as the posture task became more difficult and the simulated hearing thresholds increased, the posture task complexity affected the listening task scores. Concurrent engagement in the listening task resulted in significant differences in postural control performance. Since there was a trend for decreased posture control performance during simulated hearing loss conditions, the results underline the possibility that hearing loss may cause individuals to find challenging posture conditions more difficult while dual tasking.

Postural sway performance during dual task conditions with simulated hearing loss was always worse when compared to conditions where there were no hearing loss, both in this experiment and the experiment reported in Chapter Three. Although the results from both experiments cannot be explicitly compared, as different listening stimuli were used, it is clear that hearing loss introduced a more prominent effect on postural control.

Although thought useful to included older adults whom are medically free in this line of research for purposes of understanding the relation between listening and posture and setting the scene for future research, it is also important to emphasize that in all three experiments there were large age range which might affected the results. Older adults have been reported to have a higher postural sway in the AP and ML directions (Mitra, Knight and Munn 2013) and ageing is reported to reduce general postural stability (Rubenstein 2006). In dual task studies older adults have been also reported to prioritize the postural task over the cognitive task (Boisgontier et al. 2013; Dault and Frank 2004). So it is important to interpret the results reported here with caution to the effects of having a large age range.

Although the focus of this thesis were to investigate the attentional demands of listening on posture, mediated by attention, and review the findings from a cognitive psychology domain, it will also be reviewed and discussed in light of the previous literature concerning sound and posture which is more sensory in nature. In previous literature and as described in section 2.4 of this thesis, sound has been reported by some authors to have a stabilizing effect and by some others to have a destabilizing effect.

In these studies sound is usually found helpful in reducing body sway when sound is delivered through speakers usually on both sides of the subject, in front or behind them or around them at different angles. Different sounds were used like tonal stimuli, noise or meaningful messages and the sound source were either stationary or moving. However, this is not the case in the current line of research as the listening task was 1) meaningful and requires immediate verbal response 2) was delivered through headphones 3) participants were not instructed to focus on one task over the other. Verbally responding, articulating, has been reported to alter posture as it involves changes in respiration and using headphones takes away the spatial information that the speakers convey and hence they cannot be used as landmarks to help decrease body sway. When using headphones the sound source is not independent of the participant’s movement and hence cannot provide any spatial or acoustical information. The third point is also vital as most of the studies instruct participants to focus on the source of the sound, count the number of the tones, count the number of laps completed by sound source in moving sound sources studies. Instruction are reported to effect subjects' postural responses (Fraizer and Mitra 2008; Mitra and Fraizer 2004).

These differences between the studies conducted in this research and previous research reporting a stabilizing effect of sound might be the reasons why this line of research found the different listening tasks used in this research are causing increased postural sway. The findings here suggests that the nature of the auditory stimuli i.e. nature, source, moving or not plays a huge role in the sound and posture relation and might cause a decrease or increase in postural sway. To conclude, spatial cues, forcing attention to sound, spatial attributes of sound, morphological features of sound and motion perception all plays a role in making a sound source to make a stabilizing or destabilizing effect.

From a cognitive psychology domain, this line of research revealed a consistency of the idea that there is a limited cognitive resource (Baddeley 2012; Cowan 2005). The results presented here could be explained by, but not limited to, the work of Baddeley (2012) who suggested a general resource pool (the central executive), two sub-slave systems (the visuo-spatial loop and phonological loop) and an episodic buffer. In terms of Baddeley's multi-component model, it can be suggested that the posture tasks in this research accessed resources from the visuo-spatial loop, while the listening tasks demanded resources from the phonological loop. As the posture task and/or the listening task becomes harder, the central executive, which manages the two slave systems, reallocates attentional resources based on the difficulty of the tasks being performed. As the task difficulty increases and no further attentional resources can be allocated for execution, the performance deteriorates. This scenario can explain the results presented in this research.

It is also important to recognise that, although it is hypothesised that resources are from different modalities, the episodic buffer will, at some point, bind information from both modalities and also recall related previous knowledge from the LTM for the purpose of execution (Allen, Hitch and Baddeley 2009; Delogu, Nijboer and Postma 2012; Allen et al. 2012; Baddeley, Hitch and Allen 2009). Results can also be explained by other WM models that look at WM as a unitary resource pool or unit, with a central executive to allocate resources between tasks being performed based on demands (Cowan 2005; Cowan 2012).

Hence, it seems that performing a listening task is effortful and requires attention in a way that might compromise the performance of other concurrent tasks, i.e. posture in this study. This finding is consistent with the existing literature, which reported listening to be effortful; this effort was reported to be even higher in adverse listening conditions, such as listening in noisy conditions (Mattys et al. 2012), with hearing loss (Stenfelt and Rönnberg 2009; Gatehouse, Naylor and Elberling 2006a; Zekveld, Kramer and Festen 2011) or when performing challenging cognitive tasks (Mattys et al. 2012).

Findings from this research have been discussed in light of the sensory relation between sound and posture and also in light of the attentional demand that sound might have on posture. The results found here can be justified by both approaches and highlights the importance of investing more research in understanding this relation as understanding auditory perception through both approaches may help better understand health conditions i.e. sensory substitution and improve current rehabilitation services.

7.2 Implication of current work and Contribution to the field

The primary aim of this research is to understand how listening might influence posture control in healthy individuals with normal hearing and with simulated hearing loss to improve our understanding of this relation. A better understanding of the mechanism that underpins the relationship between listening and posture control may contribute to improved health by reducing negative consequences. This was the main motivation for my research. The experimental results presented in this thesis suggest that listening in normal hearing individuals can influence posture control and that when simulated hearing loss was introduced this influence was more prominent. This suggests that peripheral hearing loss does affect ones' postural control, and it may also suggest that posture control for individuals with actual hearing loss may be more vulnerable to listening due to other co-existing factors like ageing, cognitive decline and low memory abilities.

The implication of the current work is that listening influences posture in adults and, that this influence can be detected by a global body sway measure, and that the mechanisms individuals use to maintain erect posture can be studied using a structural body sway measure. In addition, the suggestion that as normal individuals exhibited this effect, other coexisting factors must be considered beside actual hearing loss when evaluating individuals with real hearing impairment, in order to measure postural performance changes.

The identification of MV and SDA as sensitive body sway measures to the effect of listening on posture is the highlight contribution of this research to the field. These findings will facilitate future research by focusing on using MV as a tool to detect this effect over other global body sway measures and using SDA parameters not only to detect this effect side by side with MV but also to understand the underpinning strategies individuals use to maintain posture under different conditions. MV and SDA parameters are valuable tools as they are both able to detect the effect of listening on posture. However, SDA parameters have the benefit to give more insight and provide richer information regarding posture strategies adopted by individuals to maintain erect posture. MV on the other hand is a simple sway measure which can be easily calculated either by the tester or through available commercial balance system's software sold with force plates. MV is an easy validated tool which can be easily transferred to the clinical practice and help assess patients with hearing impairments and/or hearing devices to evaluate their posture and monitor their progress. In other words, both measure are valuable, SDA has the benefit of better posture performance understanding while MV can be used as a marker of the influence that listening might have on posture. For example if the person carrying the assessment noticed this influence using MV and wants to understand the individual under investigation posture behaviour, posture performance can be then analysed using SDA to 1) confirm the effect noticed using MV and also to 2) understand how that individual changed posture strategies in order to account for that influence.

The current research showed that simulated hearing loss affects posture control under different test conditions of varying difficulty indicating that peripheral auditory impairment has the ability to affect posture performance. This suggests that when the effect of listening on posture control is investigated in individuals with actual hearing loss, other existing factors, i.e. cognitive decline or low cognitive abilities, which are known to be associated with hearing loss, may exaggerate this effect, suggesting the involvement of higher cognitive processing. If this is the case, these higher processes cannot be estimated by objective sensory sensitivity measurements only.

Findings of this research might help better understand and give reasoning to some observations like, but not limited to, the low hearings aids uptake, the cognition first phenomenon and explains why might some individuals cannot carry out a motor activity from as simple as standing to running and a conversation at the same time.

7.3 Further Research

Results from this research have raised more questions for further research to continue investigating this topic. There are a number of research questions that remain to be explored in this novel area, which have arisen as a result of the work presented here. One of them is to investigate the effect of listening on posture control under a variety of different auditory tasks. The listening tasks used in this project were words only and investigating the matter using other materials such as sentences may reveal a clearer picture, more pronounced effect or even different results. It is also important to investigating the effect of listening on posture control under a variety of postural tasks. This can take the shape of walking, for example, which might reflect a more precise picture as such a task is similar to day-to-day real life activity. From the research conducted here, the researcher can conclude that, for standing balance, the RS especially in the eyes closed condition was too difficult and extreme. I would suggest to avoid it in future research as the information it added were minimal and the same conclusions could have been drawn without doing this difficult condition.

The research carried here revealed the power of using simulated hearing loss. Investigating the effect of listening on posture control under different levels and configurations of the simulated hearing loss should inform how different levels and shapes of hearing loss might influence posture and help to understand more specific postural behaviour of specific simulated audiograms. After understanding different configurations of hearing loss and their relation to posture using the simulation, this relation should be investigated in individuals with actual hearing impairment. Although the simulated hearing loss is reported to imitate real hearing loss, it cannot reflect precisely how a real impaired ear functions. Hence there is a risk that the simulation misses aspects of a real hearing loss. It would be very useful to conduct this research on individuals with hearing impairment in order to compare the results with those obtained from this project with hearing loss simulation, and also to inform future research.

Another future research venue is to investigate if critical time points of diffusion analysis could account for the varying difficulty of the listening conditions. It is thought that the harder the task the longer the delay, where the individual will switch from open-loop mechanisms to closed-loop mechanisms. In my view, I think this measure can be a very useful tool to indicate how individuals are coping with what they are listening to, especially individuals with hearing impairment. If enough research is conducted on this measure, I think it can be used as a sign of risk of falling and it can be used as performance indicator in rehabilitation plans. Future research concerning this measure should aim for a larger sample size, less testing conditions and every condition to be repeated at least 10 times so the averaged recordings can reflect a more reliable and precise picture of posture performance using this measure.

7.4 Conclusion

The results from the current research support the hypothesis that if listening makes sufficient cognitive processing demands, then the processing capacity required to maintain posture is compromised, as executing both tasks concurrently makes them compete for cognitive resources. The work in this thesis has also demonstrated that simulated hearing loss demands extra effort and hence posture performance is prone to larger deterioration when compared to maintaining posture and listening simultaneously with no simulated hearing loss.

The work described here has shown that peripheral hearing impairment only, i.e. normal individuals with simulated hearing loss, influenced maintaining posture and caused the body to sway more under dual-task conditions. This can suggest the idea that if individuals with real hearing impairment were recruited, the observed effect would be expected to be even larger and to find out further research is required. This effect would be expected to be larger because hearing loss is associated with cognitive decline, low memory abilities and old age. This initial conclusion about how individuals with real hearing impairment might perform should direct future research, which may facilitate the development of this concept and allow a better understanding of how hearing impairment might affect posture control, which may provide a link to the risk of falls and low hearing aid uptake.

In conclusion, the hypothesis of the overarching question of this program of research is true and listening has a destabilizing effect on posture in healthy adults with normal hearing and simulated hearing losses. This effect can be detected by body sway mean velocity and the mechanisms underpins this relationship between listening and posture can be studied using stabilogram diffusion analysis.

Publications

The work in this thesis has been accepted and presented at local conferences such as the Annual University of Leeds Postgraduate Research Conference, the School of Healthcare conference and also in national and international conferences:

Alshamrani, M., Thyer, N.J. & Brooke, R. 2013. The Effect of Listening on Posture Control in Healthy Adults with Normal Hearing: Preliminary data. *The 4th Annual Conference of the British Society of Audiology, 4th-6th September 2013, Stock-on-Trent, England.*

Alshamrani, M., Thyer, N.J. & Brooke, R. 2015. Does Listening Affects Posture Control? A Study Using Global and Structural Body Sway Measures. *Third International Conference on Cognitive Hearing Science for Communication, 14th-17th June 2015, Linkoping, Sweden.*

Alshamrani, M., Thyer, N.J. & Brooke, R. 2016. Simulated Hearing Loss Affects Postural Control in Healthy Adults. *33rd World Congress of Audiology, 18th-21st September 2016, Vancouver, Canada.*

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Appendix 1: Ethical approval for experiment one

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UNIVERSITY OF LEEDS

01 March 2013

Dr Ruth E Brooke PhD
Lecturer in Audiology
School of Healthcare
Baines Wing
University of Leeds
LS2 9UT

Dear Ruth

Research Project Amendment for Ethical Approval (SHREC/RP/296)

Thank you for submitting the amendment to your research proposal *'The effects of listening on postural control'*.

This has been reviewed and I can confirm on behalf of the School of Healthcare Research Ethics Committee (SHREC) that ethical approval is granted and the amendment may be implemented based on the documentation received at date of this letter.

Document	*Version*	*Date*
Ethics Application Form	1	16/10/12
Amendment Form	1	22/02/13
Recruitment Poster	3	22/02/13
Consent Form	3	22/02/13

Ethical approval does not infer you have the right of access to any member of staff or student or documents and the premises of the University of Leeds. Nor does it imply any right of access to the premises of any other organisation, including clinical areas. SHREC takes no responsibility for you gaining access to staff, students and/or premises prior to, during or following your research activities.

Please note: You are expected to keep a record of all your approved documentation, as well as documents such as sample consent forms, and other documents relating to the study. This should be kept in your study file, and may be subject to an audit inspection. If your project is to be audited, you will be given at least 2 weeks notice.

It is our policy to remind everyone that it is your responsibility to comply with Health and Safety, Data Protection and any other legal and/or professional guidelines there may be.

The committee wishes you continued success with your project.

Yours sincerely

Kuldip Kaur Bharj

Dr Kuldip Bharj, OBE
Acting Chair, School of Healthcare Research Ethics Committee

Appendix 2: Study advertisement poster text

Study title: The Effect of listening on Posture

Participants are needed for a study investigating the effect of listening on posture.

I am looking to recruit individuals aged 18-60, who speak English as their first language and who have no history of hearing or balance problems.

The commitment would be to attend one test session, arranged during normal working hours at a time that is convenient for you, lasting no more than 90 minutes. Testing will take place within the School of Healthcare at the University of Leeds. During the session you will be asked to perform some short listening and balance tasks whilst standing on a low platform which will measure the body position. At the beginning of the session, prior to any testing taking place, you will be asked to sign a consent form agreeing to take part in the study.

If you would like to take part, would like more information or have any questions about the study please contact:

Mohammad Alshamrani

3.35 PhD suite

Baines Wing

School of Healthcare

Faculty of Medicine and Health

University of Leeds

LS2 9UT

Tel: 0113 3431374

Email: hcmaa@leeds.ac.uk

Ethical approval has been obtained for this study from the School of Healthcare Research Ethics Committee (SHREC/RP/296)

Thank you for taking the time to read this poster.

Appendix 3: Consent form

Participant Consent Form

Title of Study: The effects of listening on posture

	Please confirm agreement to the statements by putting your initials in the box below
I have read and understood the participant information sheet	
I have had the opportunity to ask questions and discuss this study	
I have received satisfactory answers to all of my questions	
I have received enough information about the study	
I understand that I am free to withdraw from the study:- 1 At any time (prior to or during the test session) 2 Without having to give a reason I understand that it will not be possible to withdraw my data after the test session as all data collected will be saved and stored anonymously.	
I understand that any information I provide, including personal details, will be kept confidential, stored securely and only accessed by those carrying out the study.	
I understand that data collected may be included in published documents but all information will be anonymised.	
I give permission to regulatory authorities to access the research data for auditing purposes.	
I agree to take part in this study	
Participant Signature ………………………………………………………… Date	
Name of Participant	
Researcher Signature …………………………………………………….. Date	
Name of Researcher	

Thank you for agreeing to take part in this study

Appendix 4: Experiment one participant information sheet

Study title: Exploring the influence of listening on posture in healthy adults with normal hearing

I would like to invite you to volunteer to take part in the above named research study but before you decide, please read the following information. It is important that you understand why this research study is being done and what it will involve before you volunteer to take part.

Please ask if anything is not clear or if you would like more information.

What is the purpose of this study?

Posture and balance control are often thought to be automatically controlled, requiring little active attention. However, recent research suggests that this may not be the case and that postural control may be affected in situations when attention is divided, for example, when multi-tasking. The aim of this research project is to investigate this further by investigating the effect of performing a listening task whilst performing a simultaneous postural control task.

Who is doing the study?

This study is being done by Mohammad Alshamrani, a PhD student based in School of Healthcare, University of Leeds. This work is being supervised by Drs Nick Thyer and Ruth Brooke, lecturers in Audiology within the School.

Who is being asked to participate?

We are asking individuals who are aged 18-60, who speak English as their first language and who have no history of hearing or balance problems to take part.

What will be involved if I take part in this study?

If you take part you will be required to attend a single test session which will last a maximum of 120 minutes. The session will be arranged during normal working hours at a time that is suitable to you and will take place within the Audiology Suite in the School of Healthcare, at the University of Leeds.

At the beginning of the test session you will be asked to complete a short hearing test which will involve listening to a series of beeps through headphones and telling the researcher whether or not you heard them. If any problems with your hearing are identified, or you report any balance problems, you will unfortunately not be able to take part in the study and you will be advised to visit your GP.

Following the hearing screen, you will be asked to complete short experiments (each last about 30 seconds and repeated 3 times) which will be performed in a random order. Experiment 1 will involve listening to different words delivered through headphones to opposing ears at the same time and telling the researcher what you heard. For experiment 2 you will be asked to stand on a low platform, also called a force plate, with eyes open and eyes closed. For experiment 3 you will be asked to complete experiment 1 and 2 simultaneously. In experiment 4 you will be asked to stand on a force plate with one foot in front of the other (heel to toe) with eyes open and eyes closed. For experiment 5 you will be asked to complete experiment 1 and 4 at the same time.

You will be provided with comfort breaks between each experiment.

What are the advantages and disadvantages of taking part?

You will need to give up some of your time to take part. By taking part you will help us gain an understanding of how sound and listening can affect our ability to maintain posture and balance. At the end of the session we will give you a £20 gift voucher to thank you for your time.

Can I withdraw from the study at any time?

Yes. Your participation is voluntary and so you can change your mind about taking part and withdraw your participation at any time (prior to and during the test session) without giving a reason. If you withdraw part way through the test session any data already collected will be deleted. However, it will not be possible to withdraw your data after the test session as all data will be saved and stored anonymously.

Will the data obtained in the study be confidential?

Yes, all information provided and data collected will be kept confidential. Your name and contact details, and the signed consent forms, will be kept confidential, stored in a secure place and only accessed by the researchers involved in the study. Personal information will not be linked to the test results in any way. Gender and age (not birth dates) will be recorded however; this will not be linked to any names. Test data will be anonymised and kept for 3 years.

What will happen to the results of the study?

The results of the experiments will be recorded, analysed and kept for 3 years prior to deletion. The results of the study will be written up in the form of a PhD thesis and may be written up and submitted for publication in an academic journal. All data used will be anonymised.

What if there is a problem?

If you have any complaints or concerns about the study you can speak to the clinic that referred you to the research team and they will do their best to answer your questions. Should you have a complaint about the way the study is being conducted, please contact my supervisor Dr Nicholas Thyer (Tel: 0113 343 1238; email: N.J.Thyer@leeds.ac.uk) or the Faculty Head of Research and Innovation Support Ms Clare Skinner (Tel: 0113 343 4897; email: c.e.skinner@leeds.ac.uk).

Who has reviewed this study?

Ethical approval has been granted by the School of Healthcare Research Ethics Committee (SHREC/RP/296).

If you agree to take part, would like more information or have any questions about the study please contact:

Mohammad Alshamrani

3.35 PhD suite

Baines Wing

School of Healthcare

Faculty of Medicine and Health

University of Leeds

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Thank you for taking the time to read this information sheet.

Appendix 5: The R software code developed to calculate sway measures for expermnit one

```
#recall the data
ff=read.table(file.choose())
fn=names(table(c(ff)))
getwd()
setwd("C:/Users/hcmaa/Desktop/Exp1")
P=list()
for(i in 1:length(fn)){
P=c(P,list(read.table(fn[i],skip=13)))
}
str(P)
x=matrix(ncol=length(fn),nrow=3000)
y=matrix(ncol=length(fn),nrow=3000)
for(i in 1:length(fn)){
 x[,i]=P[[i]]$V2
 y[,i]=P[[i]]$V3
}
str(x)
dim(x)
### looking into  the data (general stats)
summary(x)
summary(y)
### COP measures (SD of amplitude, Planar deviation, SD of Velocity,
## SD of amplitude
MLam=c()
APam=c()
for(i in 1:length(fn)){
MLam[i]=sd(x[,i])
APam[i]=sd(y[,i])
}
## planar deviation
pldev=sqrt(MLam^2+APam^2)
## SD of velocity
vx=matrix(ncol=length(fn),nrow=2999)
for(j in 1:length(fn)){
```

```
for(i in 1:2999){
vx[i,j]=x[i+1,j]-x[i,j]/(0.01)
}
su=matrix(ncol=length(fn),nrow=2999)
for(j in 1:length(fn)){
for (i in 1:2999){
su[i,j]=((vx[i,j]-mean(vx[,j]))^2)
}
MLvelocity=c()
for (j in 1 : length(fn)){
MLvelocity[j]=sqrt(sum(su[,j])/2998)
}
vy=matrix(ncol=length(fn),nrow=2999)
for(j in 1:length(fn)){
for(i in 1:2999){
vy[i,j]=y[i+1,j]-y[i,j]/(0.01)
}
su=matrix(ncol=length(fn),nrow=2999)
for(j in 1:length(fn)){
for (i in 1:2999){
su[i,j]=((vy[i,j]-mean(vy[,j]))^2)
}
APvelocity=c()
for (j in 1 : length(fn)){
APvelocity[j]=sqrt(sum(su[,j])/2998)
}
##### mean velocity
vx=matrix(ncol=length(fn),nrow=2999)
for(j in 1:length(fn)){
for(i in 1:2999){
vx[i,j]=x[i+1,j]-x[i,j]
}
vy=matrix(ncol=length(fn),nrow=2999)
for(j in 1:length(fn)){
for(i in 1:2999){
vy[i,j]=y[i+1,j]-y[i,j]
}
VD=(sqrt((vx)^2+(vy)^2))/0.01
```

```
dim(VD)
VM=c()
for(i in 1:792){
VM[i]=sum(VD[,i])/2999
}
#mean velocity of condition 1
rep=(1:33)*24+1
rep2=(1:33)*24+2
rep3=(1:33)*24+3
REP1=c(1:3,rep[1:32],rep2[1:32],rep3[1:32])
VM1=VM[REP1]
GVM1=sum(VM1)/99 # mean(VM1)
SDVM1=sd(VM1)
# mean velocity of condition 2
rep=(1:33)*24+4
rep2=(1:33)*24+5
rep3=(1:33)*24+6
REP2=c(4:6,rep[1:32],rep2[1:32],rep3[1:32])
VM2=VM[REP2]
GVM2=sum(VM2)/99 # mean(VM2)
SDVM2=sd(VM2)
# mean velocity of condition 3
rep=(1:33)*24+7
rep2=(1:33)*24+8
rep3=(1:33)*24+9
REP3=c(7:9,rep[1:32],rep2[1:32],rep3[1:32])
VM3=VM[REP3]
GVM3=sum(VM3)/99 # mean(VM3)
SDVM3=sd(VM3)
# mean velocity of condition 4
rep=(1:33)*24+10
rep2=(1:33)*24+11
rep3=(1:33)*24+12
REP4=c(10:12,rep[1:32],rep2[1:32],rep3[1:32])
VM4=VM[REP4]
GVM4=sum(VM4)/99 # mean(VM4)
SDVM4=sd(VM4)
```

```
# mean velocity of condition 5
rep=(1:33)*24+13
rep2=(1:33)*24+14
rep3=(1:33)*24+15
REP5=c(13:15,rep[1:32],rep2[1:32],rep3[1:32])
VM5=VM[REP5]
GVM5=sum(VM5)/99 # mean(VM5)
SDVM5=sd(VM5)
# mean velocity of condition 6
rep=(1:33)*24+16
rep2=(1:33)*24+17
rep3=(1:33)*24+18
REP6=c(16:18,rep[1:32],rep2[1:32],rep3[1:32])
VM6=VM[REP6]
GVM6=sum(VM6)/99 # mean(VM6)
SDVM6=sd(VM6)
# mean velocity of condition 7
rep=(1:33)*24+19
rep2=(1:33)*24+20
rep3=(1:33)*24+21
REP7=c(19:21,rep[1:32],rep2[1:32],rep3[1:32])
VM7=VM[REP7]
GVM7=sum(VM7)/99 # mean(VM1)
SDVM7=sd(VM7)
# mean velocity of condition 8
rep=(1:33)*24+22
rep2=(1:33)*24+23
rep3=(1:33)*24+24
REP8=c(22:24,rep[1:32],rep2[1:32],rep3[1:32])
VM8=VM[REP8]
GVM8=sum(VM8)/99 # mean(VM1)
SDVM8=sd(VM8)
## ppp (phase plane portrait)
MLppp=sqrt(MLam^2+MLvelocity^2)
APppp=sqrt(APam^2+APvelocity^2)
totppp=sqrt(APppp^2+MLppp^2)
save("totppp",file="totppp.RData")
# Sway Area
```

```
sxy=matrix(ncol=length(fn),nrow=3000)
for(j in 1:length(fn)){
for(i in 1:3000){
sxy[i,j]=(x[i,j]-mean(x[,j]))*(y[i,j]-mean(y[,j]))
}
sdxy=c()
for(j in 1:length(fn)){
sdxy[j]=(sum(sxy[,j])/2999)
}
#Area
AR=c()
for(j in 1:length(fn)){
AR[j]=2*pi*pf(0.05,2, 2998)*sqrt(APam[j]^2*MLam[j]^2+(sdxy[j]^2))
}
#AR condition 1
AR1=AR[REP1]
GMAR1=mean(AR1)
SDAR1=sd(AR1)
#AR condition 2
AR2=AR[REP2]
GMAR2=mean(AR2)
SDAR2=sd(AR2)
#AR condition 3
AR3=AR[REP3]
GMAR3=mean(AR3)
SDAR3=sd(AR3)
#AR condition 4
AR4=AR[REP4]
GMAR4=mean(AR4)
SDAR4=sd(AR4)
#AR condition 5
AR5=AR[REP5]
GMAR5=mean(AR5)
SDAR5=sd(AR5)
#AR condition 6
AR6=AR[REP6]
GMAR6=mean(AR6)
SDAR6=sd(AR6)
```

```
#AR condition 7
AR7=AR[REP7]
GMAR7=mean(AR7)
SDAR7=sd(AR7)
#AR condition 8
AR8=AR[REP8]
GMAR8=mean(AR8)
SDAR8=sd(AR8)
```

Appendix 6: Competing words list and scoring sheet

Competing Words–Free Recall (CW–FR)

Circle + for words that are repeated correctly. Circle – for incorrect responses or no response.

Tracks

10 Test Directions
11 Practice Items a & b
12 Test Items

Right Ear			Left Ear		
Practice Items					
a. dig			barn		
b. yard			hill		
Test Items					
1. fell	+	–	pink	+	–
2. coat	+	–	tent	+	–
3. glove	+	–	milk	+	–
4. road	+	–	bake	+	–
5. hope	+	–	room	+	–
6. lake	+	–	ant	+	–
7. tell	+	–	woke	+	–
8. wet	+	–	hand	+	–
9. feel	+	–	sheep	+	–
10. bad	+	–	read	+	–
11. care	+	–	bed	+	–
12. fruit	+	–	bus	+	–
13. nest	+	–	mud	+	–
14. chip	+	–	night	+	–
15. gave	+	–	some	+	–
16. five	+	–	gone	+	–
17. ice	+	–	move	+	–
18. ran	+	–	gold	+	–
19. frog	+	–	mean	+	–
20. soft	+	–	last	+	–

RE Score [ ] + LE Score [ ] = [ ]

CW–FR Total Score

Appendix 7: Medical history screening questionnaire

Screening questionnaire

Name: …………………………………..........

Age: ……………………………………...........

Sex: …………………………………………

Educational attainment: ……………..........

Date: ………………………………………..

1. Do you have any hearing problems? Yes No

If yes please specify...…………………………………………...………

2. Do you have any balance problems? Yes No

If yes, please specify.……………………………………………..……...

3. Do you suffer from tinnitus (Ringing in the ear)? Yes No

If yes please specify…………………………………………………..

4. Do you have any major health problems? Yes No

If yes, please specify…………………………………………………….

………………………………………………………………………..

5. Use the blank below if you want to add any related information about your hearing or your general health situation……………………………

……………………………………………………………………………

……………………………………………………………………………

Thank you

Appendix 8: Example Audiogram

FONIX® HEARING EVALUATOR™
Digital Audiometer

NAME ______________________ AGE ______ DATE ____________

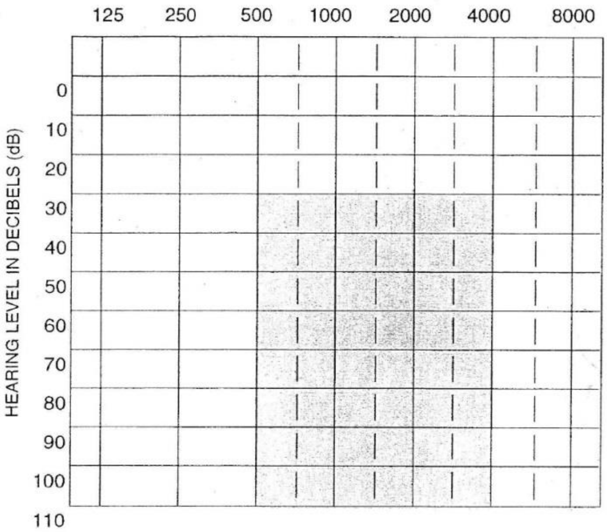
ADDRESS ______________________ PHONE ____________

CITY, STATE & ZIP ______________________

EXAMINER ______________________ AUD. SERIAL NO. ____________

COMMENTS ______________________

AUDIOGRAM



AUDIOGRAM KEY

		RIGHT	LEFT
AIR	UNMASKED	○	×
	MASKED	△	□
BONE MASTOID	UNMASKED	<	>
	MASKED	[	]

EXAMPLES OF NO RESPONSE SYMBOLS

	SRT	Speech Discrimination	MCL	UCL
R		% @ dB		
L		% @ dB		

HEARING AID SIMULATOR RESULTS

	dB/Octv.	MCL/GAIN	UCL/SPL (MPO)
R		dB	dB
L		dB	dB

COMMENTS ______________________

PTA (dB HL)

	500 Hz	1000 Hz	2000 Hz
R	dB	dB	dB
L	dB	dB	dB

TARGET 2cc - FOG

Printout(s) Enclosed ❑ Yes ❑ No

	250	500	1K	2K	3K	4K
R						
L						

FRYE

FRYE ELECTRONICS, INC.
P.O. Box 23391 • Tigard, OR 97281-3391 • USA
(503) 620-2722 • (800) 547-8209
Fax: (503) 639-0128
www.frye.com • e-mail: sales@frye.com

REF: MISC. DOCUMENTS / FORMS / FA-10 FORM 7/04

Appendix 9: Ethical approval for experiment two

Faculty of Medicine and Health

Research Office

University of Leeds
Worsley Building
Clarendon Way
Leeds LS2 9NL
United Kingdom

+44 (0) 113 343 4361

UNIVERSITY OF LEEDS

11 November 2013

Mohammad Alshamrani
3.35 PhD Suite
School of Healthcare
University of Leeds

Dear Mohammad

Ref no: SHREC/RP/296 – Amendment_2

Title: The effects of listening on postural control

Thank you for submitting the amendment to the above named project.

This has been reviewed and I can confirm on behalf of the School of Healthcare Research Ethics Committee (SHREC) that ethical approval is granted and the amendment may be implemented based on the documentation received at date of this letter.

Document	*Version*	*Date submitted*
SHREC Amendment_form	1	05.11.13
Participant Information sheet (V1.2)	1	05.11.13

Ethical approval does not infer you have the right of access to any member of staff or student or documents and the premises of the University of Leeds. Nor does it imply any right of access to the premises of any other organisation, including clinical areas. SHREC takes no responsibility for you gaining access to staff, students and/or premises prior to, during or following your research activities.

Please note: You are expected to keep a record of all your approved documentation, as well as documents such as sample consent forms, and other documents relating to the study. This should be kept in your study file, and may be subject to an audit inspection. If your project is to be audited, you will be given at least 2 weeks notice.

It is our policy to remind everyone that it is your responsibility to comply with Health and Safety, Data Protection and any other legal and/or professional guidelines there may be.

The committee wishes you continued success with your project.

Yours sincerely

Kuldip Kaur Bharj

Dr Kuldip Bharj, OBE
Acting Chair, School of Healthcare Research Ethics Committee

Appendix 10: Experiment two participant information sheet

Study title: Studying the influence of listening on posture control using stabilogram diffusion analysis

I would like to invite you to volunteer to take part in the above named research study but before you decide, please read the following information. It is important that you understand why this research study is being done and what it will involve before you volunteer to take part.

Please ask if anything is not clear or if you would like more information.

What is the purpose of this study?

Posture and balance control are often thought to be automatically controlled, requiring little active attention. However, recent research suggests that this may not be the case and that postural control may be affected in situations when attention is divided, for example, when multi-tasking. The aim of this research project is to investigate this further by investigating the effect of performing a listening task whilst performing a simultaneous postural control task.

Who is doing the study?

This study is being done by Mohammad Alshamrani, a PhD student based in School of Healthcare, University of Leeds. This work is being supervised by Drs Nick Thyer and Ruth Brooke, lecturers in Audiology within the School.

Who is being asked to participate?

We are asking individuals who are aged 18-60, who speak English as their first language and who have no history of hearing or balance problems to take part.

What will be involved if I take part in this study?

If you take part you will be required to attend a single test session which will last a maximum of 120 minutes. The session will be arranged during normal working hours at a time that is suitable to you and will take place within the Audiology Suite in the School of Healthcare, at the University of Leeds.

At the beginning of the test session you will be asked to complete a short hearing test which will involve listening to a series of beeps through headphones and telling the researcher whether or not you heard them. If any problems with your hearing are identified, or you report any balance problems, you will unfortunately not be able to take part in the study and you will be advised to visit your GP.

Following the hearing screen, you will be asked to complete short experiments (each last about 30 seconds and repeated 5 times) which will be performed in a random order. Experiment 1 will involve listening to different words delivered through headphones to opposing ears at the same time and telling the researcher what you heard. For experiment 2 you will be asked to stand on a low platform, also called a force plate, with eyes open and eyes closed. For experiment 3 you will be asked to complete experiment 1 and 2 simultaneously. In experiment 4 you will be asked to stand on a force plate with one foot in front of the other (heel to toe) with eyes open and eyes closed. For experiment 5 you will be asked to complete experiment 1 and 4 at the same time.

You will be provided with comfort breaks between each experiment.

What are the advantages and disadvantages of taking part?

You will need to give up some of your time to take part. By taking part you will help us gain an understanding of how sound and listening can affect our ability to maintain posture and balance. At the end of the session we will give you a £20 gift voucher to thank you for your time.

Can I withdraw from the study at any time?

Yes. Your participation is voluntary and so you can change your mind about taking part and withdraw your participation at any time (prior to and during the test session) without giving a reason. If you withdraw part way through the test session any data already collected will be deleted. However, it will not be possible to withdraw your data after the test session as all data will be saved and stored anonymously.

Will the data obtained in the study be confidential?

Yes, all information provided and data collected will be kept confidential. Your name and contact details, and the signed consent forms, will be kept confidential, stored in a secure place and only accessed by the researchers involved in the study. Personal information will not be linked to the test results in any way. Gender and age (not birth dates) will be recorded however; this will not be linked to any names. Test data will be anonymised and kept for 3 years.

What will happen to the results of the study?

The results of the experiments will be recorded, analysed and kept for 3 years prior to deletion. The results of the study will be written up in the form of a PhD thesis and may be written up and submitted for publication in an academic journal. All data used will be anonymised.

<u>What if there is a problem?</u>

If you have any complaints or concerns about the study you can speak to the clinic that referred you to the research team and they will do their best to answer your questions. Should you have a complaint about the way the study is being conducted, please contact my supervisor Dr Nicholas Thyer (Tel: 0113 343 1238; email: N.J.Thyer@leeds.ac.uk) or the Faculty Head of Research and Innovation Support Ms Clare Skinner (Tel: 0113 343 4897; email: c.e.skinner@leeds.ac.uk).

<u>Who has reviewed this study?</u>

Ethical approval has been granted by the School of Healthcare Research Ethics Committee (SHREC/RP/296).

If you agree to take part, would like more information or have any questions about the study please contact:

Mohammad Alshamrani

3.35 PhD suite

Baines Wing

School of Healthcare

Faculty of Medicine and Health

University of Leeds

LS2 9UT

Tel: 0113 3431374

Email: hcmaa@leeds.ac.uk

Thank you for taking the time to read this information sheet.

Appendix 11: Stabilogram diffusion analysis code

11-1 cop.m

```
%COP Data Analysis Program by Andrea Stamp summer 1997
%Commercial use of this software must be approved by the author:
imandie@bu.edu
%This program analyzes center of pressure data and requires the
following:
%       signals toolbox from MATLAB
%       cpdf.m          plotgr.m
%       datcp.m         plotmult.m
%       findtime.m      plotraw.m
%       sparam.m
%
%This specific m file just sets up the graphical user interface for
my program

% Parameter setups for the control window
h = figure;
set(h,'name','Andrea''s COP Analysis Program');
set(h,'menubar','none');
set(h,'color',[0 0 0]);


% Analyze pulldown menu
analyze = uimenu(h,'Label','Analyze');
adattocp = uimenu(analyze,'Label','dat -> cp','CallBack','datcp,');
acptodf = uimenu(analyze,'Separator','on','Label','cp ->
df','CallBack','cpdf,');
adattodf = uimenu(analyze,'Separator','on','Label','dat ->
df','CallBack',['datcp,','cpdf,']);

% Random Options pulldown menu
options = uimenu(h,'Label','Random Options');
oparam = uimenu(options,'Label','Show
Parameters','Checked','on','CallBack',['if
strcmp(get(oparam,''Checked''),''off''),',
'set(oparam,''Checked'',''on''),','else,','set(oparam,''Checked'','
'off''),','end,']);
opoints = uimenu(options,'Separator','on','Label','# of data
points');
o3000 =
uimenu(opoints,'Label','3000','Checked','On','CallBack',['if
strcmp(get(o3000,''Checked''),''off''),',
'set(o3000,''Checked'',''on''),','set(o6000,''Checked'',''off''),',
'end,']);
o6000 = uimenu(opoints,'Label','6000','CallBack',['if
strcmp(get(o6000,''Checked''),''off''),',
'set(o6000,''Checked'',''on''),','set(o3000,''Checked'',''off''),',
'end,']);

% Graphics pulldown menu
graphics = uimenu(h,'Label','Graphics');
```

```
greg = uimenu(graphics,'Label','Show Regression
Lines','Checked','on','CallBack',['if
strcmp(get(greg,''Checked''),''off''),',
'set(greg,''Checked'',''on''),','else,','set(greg,''Checked'',''off
''),','end,']);
gx = uimenu(graphics,'Separator','on','Label','X');
gxlin = uimenu(gx,'Label','Linear','Checked','on','CallBack',['if
strcmp(get(gxlin,''Checked''),''off''),',
'set(gxlin,''Checked'',''on''),','else,','set(gxlin,''Checked'',''o
ff''),','end,']);
gxlog = uimenu(gx,'Label','Log','Checked','on','CallBack',['if
strcmp(get(gxlog,''Checked''),''off''),',
'set(gxlog,''Checked'',''on''),','else,','set(gxlog,''Checked'',''o
ff''),','end,']);
gy = uimenu(graphics,'Label','Y');
gylin = uimenu(gy,'Label','Linear','Checked','on','CallBack',['if
strcmp(get(gylin,''Checked''),''off''),',
'set(gylin,''Checked'',''on''),','else,','set(gylin,''Checked'',''o
ff''),','end,']);
gylog = uimenu(gy,'Label','Log','Checked','on','CallBack',['if
strcmp(get(gylog,''Checked''),''off''),',
'set(gylog,''Checked'',''on''),','else,','set(gylog,''Checked'',''o
ff''),','end,']);
gr = uimenu(graphics,'Label','R');
grlin = uimenu(gr,'Label','Linear','Checked','on','CallBack',['if
strcmp(get(grlin,''Checked''),''off''),',
'set(grlin,''Checked'',''on''),','else,','set(grlin,''Checked'',''o
ff''),','end,']);
grlog = uimenu(gr,'Label','Log','Checked','on','CallBack',['if
strcmp(get(grlog,''Checked''),''off''),',
'set(grlog,''Checked'',''on''),','else,','set(grlog,''Checked'',''o
ff''),','end,']);
graw = uimenu(graphics,'Separator','on','Label','Plot Raw CP data',
'CallBack','plotraw,');
gmult = uimenu(graphics,'Separator','on','Label','Plot Multiple df
files','CallBack','plotmult,');

% Editable box in lefthand corner of the screen
fileframe = uicontrol(h,'Style','frame','Position',[0 310 315
110]);
filetext = uicontrol(h,'Style','text','String','Input
Options','Position',[35 365 200 50]);
filenum = uicontrol(h,'Style','text','String','# of
files:','Position',[5 375 65 20]);
filenum2box = uicontrol(h,'Style','frame','Position',[100 375 50
20]);
filenum2 =
uicontrol(h,'Style','edit','String','10','foregroundcolor',[0 0
0],'Position',[100 375 50 20]);
filepatho = uicontrol(h,'Style','text','String','Path to open
from:','Position',[5 355 120 20]);
filepatho2 =
uicontrol(h,'Style','edit','String','/home/users/','foregroundColor
',[0 0 0],'Position',[125 355 185 20]);
filepathc = uicontrol(h,'Style','text','String','Path to save
to:','Position',[7 335 100 20]);
```

```
filepathc2 =
uicontrol(h,'Style','edit','String','/home/users/','foregroundColor
',[0 0 0],'Position',[125 335 185 20]);
filedata =
uicontrol(h,'Style','text','String','Files:','Position',[7 315 40
20]);
filedata2 = uicontrol(h,'Style','edit','foregroundColor',[0 0
0],'Position',[45 315 265 20]);
warning = uicontrol(h,'Style','frame','Position',[110 200 350 40]);
warning2 = uicontrol(h,'Style','text','String','Warning: YOU MUST
PRESS RETURN AFTER YOU','Position',[110 200 350
40],'foregroundcolor',[0 0 0]);
warning3 = uicontrol(h,'Style','text','String','CHANGE THE TEXT IN
THE OPTIONS BOX ABOVE','Position',[125 200 330
20],'foregroundcolor',[0 0 0]);
```

11-2 cpdf.m

```
% This is the main function of my graphical user interface program
%
%
%
% Get some parameters from the user's screen
set(h,'Pointer','watch');
datafilesnum = str2num(get(filenum2,'String'));       %get
parameters from screen
datapatho = get(filepatho2,'String');
datapathc = get(filepathc2,'String');
datafiles = get(filedata2,'String');

if strcmp(get(o3000,'Checked'),'on'),
        NPTS = 3000;
else
        NPTS = 6000;
end

LAGMAX = 1000;          %This is the number of data points in the
resulting df files (ie 10sec worth)
        X = zeros(1,LAGMAX +2);
        Y = zeros(1,LAGMAX +2);
        data1 = strtok(datafiles);
for i=1:datafilesnum,  %Loop through all datafiles: load, calculate
correlations, average all trials
                data = strtok(datafiles);
                load([datapatho,data,'.cp'])
                datafiles = strrep(datafiles,strtok(datafiles),' ');


        temp = eval(data);

        [a b] = size(temp);
        if a ~= NPTS,
                error('Your files do not contain the data points
specified');
        end

        tempX = temp(1:NPTS,1);
        tempY = temp(1:NPTS,2);
        tempX = tempX';
```

```
        tempY = tempY';
        tempX = tempX - mean(tempX);
        tempY = tempY - mean(tempY);


%       corrX = acf(tempX,NPTS,LAGMAX);               %This acf
function is a c program written with the help of Michael Lauk
%       corrY = acf(tempY,NPTS,LAGMAX);               %This c
function does not run as fast anymore and has been commented from
this program

        corrX = xcorr(tempX,tempX,'biased');
        fix = corrX;   %calculate correlations
        corrX = corrX(NPTS:NPTS+LAGMAX +1);   %take last half of
symmetric corrX
        corrX = 2*var(tempX) - 2.*corrX;

        corrY = xcorr(tempY,tempY,'biased');  %Repeat for Y
        corrY = corrY(NPTS:NPTS+LAGMAX+1);
        corrY = 2*var(tempY) - 2.*corrY;

%       corrX = corrX(LAGMAX + 1:2*LAGMAX +2);%Only needed if acf c
program used
%       corrY = corrY(LAGMAX + 1:2*LAGMAX +2);
%       corrX = 2*var(tempX) - 2.*corrX;
%       corrY = 2*var(tempY) - 2.*corrY;


        X = X + corrX;                                %Add all correlated
files
        Y = Y + corrY;
        if i == 5,
                fprintf('\n\nAre we having fun yet????\n');
        end
end
        X = X./datafilesnum;                          %average all files
        Y = Y./datafilesnum;
        R = X + Y;

%save .df data
        fprintf('\nWriting data to file: %s\n',[data1,'.df']);
        fid = fopen([datapathc,data1,'.df'],'wt');
        Z = [X; Y];
        fprintf(fid,'%f %f\n',Z);
        fclose(fid);

%find fit times
        filt = firls(40,[0 1],[1 1],'differentiator');       %find
coefficients for filter and filter data twice for second derivativ
        derdata = filter(filt,1, X);
        der2data = filter(filt,1,derdata);
        [fitTimeX,critpointX] = findtime(der2data);     %Use my
findtime.m function to find the critical points and times used to
fit the regression lines

        derdata = filter(filt,1, Y);                  %Repeat for Y
        der2data = filter(filt,1,derdata);
        [fitTimeY, critpointY] = findtime(der2data);
```

```
        derdata = filter(filt,1, R);                    %Repeat for R
        der2data = filter(filt,1,derdata);
        [fitTimeR, critpointR] = findtime(der2data);

%fit linear lines
        tau  = 0.01:.01:(LAGMAX +2)*.01;

        tau1Xplot = fitTimeX(1):.01:fitTimeX(2)+1.2;          %These
extra variables are to plot the lines longer than the times used to
fit the curves
        tau2Xplot = fitTimeX(3)-1.2:.01:fitTimeX(4);
        fitcoeffx1 =
polyfit(fitTimeX(1):.01:fitTimeX(2),X(fitTimeX(1)*100:fitTimeX(2)*1
00),1);
        linex1 = polyval(fitcoeffx1,tau1Xplot);

        fitcoeffx2 =
polyfit(fitTimeX(3):.01:fitTimeX(4),X(fitTimeX(3)*100:fitTimeX(4)*1
00),1);
        linex2 = polyval(fitcoeffx2,tau2Xplot);

        tau1Yplot = fitTimeY(1):.01:fitTimeY(2)+1.2;
        tau2Yplot = fitTimeY(3)-1.2:.01:fitTimeY(4);
        fitcoeffy1 =
polyfit(fitTimeY(1):.01:fitTimeY(2),Y(fitTimeY(1)*100:fitTimeY(2)*1
00),1);
        liney1 = polyval(fitcoeffy1,tau1Yplot);

        fitcoeffy2 =
polyfit(fitTimeY(3):.01:fitTimeY(4),Y(fitTimeY(3)*100:fitTimeY(4)*1
00),1);
        liney2 = polyval(fitcoeffy2,tau2Yplot);

        tau1Rplot = fitTimeR(1):.01:fitTimeR(2)+1.2;
        tau2Rplot = fitTimeR(3)-1.2:.01:fitTimeR(4);
        fitcoeffr1 =
polyfit(fitTimeR(1):.01:fitTimeR(2),R(fitTimeR(1)*100:fitTimeR(2)*1
00),1);
        liner1 = polyval(fitcoeffr1,tau1Rplot);

        fitcoeffr2 =
polyfit(fitTimeR(3):.01:fitTimeR(4),R(fitTimeR(3)*100:fitTimeR(4)*1
00),1);
        liner2 = polyval(fitcoeffr2,tau2Rplot);

%fit log-log lines
        X(1) = X(1) + .00001;
        Y(1) = Y(1) + .00001;
        R(1) = R(1) + .00001;
        logX = log10(X);
        logY = log10(Y);
        logR = log10(R);


        fitcoeffx1log =
polyfit(log10(fitTimeX(1):.01:fitTimeX(2)),logX(fitTimeX(1)*100:fit
TimeX(2)*100),1);
```

```
        linex1log =
(tau1Xplot.^fitcoeffx1log(1))*10^fitcoeffx1log(2);

        fitcoeffx2log =
polyfit(log10(fitTimeX(3):.01:fitTimeX(4)),logX(fitTimeX(3)*100:fit
TimeX(4)*100),1);
        linex2log =
(tau2Xplot.^fitcoeffx2log(1))*10^fitcoeffx2log(2);

        fitcoeffy1log =
polyfit(log10(fitTimeY(1):.01:fitTimeY(2)),logY(fitTimeY(1)*100:fit
TimeY(2)*100),1);
        liney1log =
(tau1Yplot.^fitcoeffy1log(1))*10^fitcoeffy1log(2);

        fitcoeffy2log =
polyfit(log10(fitTimeY(3):.01:fitTimeY(4)),logY(fitTimeY(3)*100:fit
TimeY(4)*100),1);
        liney2log =
(tau2Yplot.^fitcoeffy2log(1))*10^fitcoeffy2log(2);

        fitcoeffr1log =
polyfit(log10(fitTimeR(1):.01:fitTimeR(2)),logR(fitTimeR(1)*100:fit
TimeR(2)*100),1);
        liner1log =
(tau1Rplot.^fitcoeffr1log(1))*10^fitcoeffr1log(2);

        fitcoeffr2log =
polyfit(log10(fitTimeR(3):.01:fitTimeR(4)),logR(fitTimeR(3)*100:fit
TimeR(4)*100),1);
        liner2log =
(tau2Rplot.^fitcoeffr2log(1))*10^fitcoeffr2log(2);

figure;
plotgr;                         %This plotgr.m script file graphs
what is specified by the user
axis([0 10 0 max(R)+10]);
sparam;                         %This sparam.m script file prints the
numerical data if requested


%Save parameters in file with .dfp extension (for df parameters)

        fprintf('\nWriting parameters to file:
%s\n',[data1,'.dfp']);
        fid = fopen([datapathc,data1,'.dfp'],'wt');
        param = [critpointX*.01; X(critpointX); fitcoeffx1(1)/2;
fitcoeffx2(1)/2; fitcoeffx1log(1)/2;
        fitcoeffx2log(1)/2; critpointY*.01; Y(critpointY);
fitcoeffy1(1)/2; fitcoeffy2(1)/2;
        fitcoeffy1log(1)/2; fitcoeffy2log(1)/2; critpointR*.01;
R(critpointR); fitcoeffr1(1)/2;
        fitcoeffr2(1)/2; fitcoeffr1log(1)/2; fitcoeffr2log(1)/2];
        fprintf(fid,'%f\n',param);
        fclose(fid);
set(h,'Pointer','arrow');

%This .dfp file is saved as follows: X critical point lag time
```

```
%                                   The value at this lag time
%                                   The slope/2 for the short term
linear X
%                                   The slope/2 for the long term
linear X
%                                   The slope/2 for the short term
log X
%                                   The slope/2 for the long term
log X
% This pattern repeats for the Y values and finally for the R
values where each
% parameter is on a separate line. (total of 18 parameters)!
```

11-3 sparam.m

```
%This program is used when the user requests to see the numerical
parameter results.  It is
%shown in the command window.


%Check to see if user wants to see results
if strcmp(get(oparam,'Checked'),'on'),

        fprintf('\nTimes Used to Fit Lines:\n\n');

        fprintf('For X:\nLINE 1: initial lag = %7.3f,  final lag =
%7.3f\n', fitTimeX(1)-.01, fitTimeX(2));
        fprintf('LINE 2: initial lag = %7.3f,  final lag = %7.3f\n',
fitTimeX(3), fitTimeX(4));
        fprintf('\nFor Y:\nLINE 1: initial lag = %7.3f,  final lag =
%7.3f\n', fitTimeY(1)-.01, fitTimeY(2));
        fprintf('LINE 2: initial lag = %7.3f,  final lag = %7.3f\n',
fitTimeY(3), fitTimeY(4));
        fprintf('\nFor R:\nLINE 1: initial lag = %7.3f,  final lag =
%7.3f\n', fitTimeR(1)-.01, fitTimeR(2));
        fprintf('LINE 2: initial lag = %7.3f,  final lag = %7.3f\n',
fitTimeR(3), fitTimeR(4));

        fprintf('\nDiffusion Coefficients:\n\n');
        fprintf('Dxs = slope/2= \t%7.3f\n', fitcoeffx1(1)/2);
        fprintf('Dxl = slope/2= \t%7.3f\n', fitcoeffx2(1)/2);
        fprintf('Dys = slope/2= \t%7.3f\n', fitcoeffy1(1)/2);
        fprintf('Dyl = slope/2= \t%7.3f\n', fitcoeffy2(1)/2);
        fprintf('Drs = slope/2= \t%7.3f\n', fitcoeffr1(1)/2);
        fprintf('Drl = slope/2= \t%7.3f\n', fitcoeffr2(1)/2);

%Show log-log data
        fprintf('\nScaling Exponents:\n\n');

        fprintf('Hxs = slope/2= \t%7.3f\n', fitcoeffx1log(1)/2);
        fprintf('Hxl = slope/2= \t%7.3f\n', fitcoeffx2log(1)/2);
        fprintf('Hys = slope/2= \t%7.3f\n', fitcoeffy1log(1)/2);
        fprintf('Hyl = slope/2= \t%7.3f\n', fitcoeffy2log(1)/2);
        fprintf('Hrs = slope/2= \t%7.3f\n', fitcoeffr1log(1)/2);
        fprintf('Hrl = slope/2= \t%7.3f\n', fitcoeffr2log(1)/2);

%Show critical times
        fprintf('\nCritical Point Coordinates:\n\n');
```

```
        fprintf('delta txc= \t%7.3f\n',critpointX*.01);
        fprintf('<X^2>= \t\t%7.3f\n',X(critpointX));
        fprintf('delta tyc= \t%7.3f\n',critpointY*.01);
        fprintf('<Y^2>= \t\t%7.3f\n',Y(critpointY));
        fprintf('delta trc= \t%7.3f\n',critpointR*.01);
        fprintf('<R^2>= \t\t%7.3f\n',R(critpointR));
end
```

11-4 datcp,m

```
%This script file is called by the program before cpdf.m if the
user has .dat files that need to
%be converted to .cp files before analysis is done.

%Get some parameters from the screen

if strcmp(get(o3000,'Checked'),'on'),
        NPTS = 3000;
else
        NPTS = 6000;
end

LAGMAX = 1000;
datafilesnum = str2num(get(filenum2,'String'));
datafiles = get(filedata2,'String');
set(h,'Pointer','watch');
datapatho = get(filepatho2,'String');
datapathc = get(filepathc2, 'String');

X = zeros(1,LAGMAX+2);
Y = zeros(1,LAGMAX+2);
for i=1:datafilesnum,                   %Loop through the data files:
load, convert from .dat to .cp
                data = strtok(datafiles);
                fid = fopen([datapatho,data,'.dat'],'rt');

                gain = fscanf(fid,'%d',1);
                [temp1,mysize] = fscanf(fid,'%d',[3000,2]);
                fclose(fid);
                temp2 = fixit(temp1);
                if mysize ~= 6000,
                        error('30 seconds worth of data was not
scanned');
                end

%convert .dat files to .cp files and save (the next few lines are
%taken from cpkist4.c to do the conversion from .dat files to .cp
files)
                if gain == 3,
                        softgain = 8;
                elseif gain == 2,
                        softgain = 4;
                elseif gain == 1,
                        softgain = 2;
                else softgain = 1;
                end
```

```
                k = 50*20/4096.0/softgain;
                temp = temp2*k;

                fid = fopen([datapathc,data,'.cp'],'wt');
                fprintf(fid,'%f %f\n',[temp(:,1)'; temp(:,2)']);
                fclose(fid);


                datafiles = strrep(datafiles,strtok(datafiles),' ');

        [a b] = size(temp);
        if a ~= 3000,
                error('You do not have 30 sec. worth of data');
        end
end
set(h,'Pointer','arrow');
```

11-5 findtime.m

```
function [fitTime, critpoint] = findtime(der2data)

%This function traverses the second derivative of the data and
finds the times
%that the linear regression curves should be fitted with.

fitTime(1) = 1;                                      %First time is
0.0
        [trash fitTime(2)] = max(der2data(1:100));  %Second time is
first max from 0 to 1 sec.
        [trash critpoint] = min(der2data(1:250));   %critical point
is first min of filter from 0 to

%This is my alogrithm to find the third and fourth fitTimes.  I
traveled the data
%backwards from 9 sec. with a window of 25 data points in order to
account for noisy
%derivatives and slight fluctuations.  When I encountered a window
that had an absolute
%max or a min enclosed, I used the max or min function to calculate
the absolute max or
%min of that window.  The fourth time is the first max to the left
of the first minimum
%traveling backwards from 9 sec.  If this time is below 7 sec. then
the fourth point is
%automatically 9 sec.  The third time is the second max from the
start of the data at 0
%sec.

%Find first minimum from righthand side of derivative

        min1 = 0;
        i = 900;
        while min1 == 0,
                if der2data(i-25) < der2data(i),
                        j = i;
                        while der2data(j-25) < der2data(j),
                                j = j - 1;
```

```
                        end
                        [trash min1] = min(der2data(j-25:j));
                end
        i = i - 1;
        end

%Find first maximum to the left of the minimum just found
        min1 = min1 + j - 25;
        i = min1;
        if der2data(i-25) > der2data(i),
                j = i;
                while der2data(j-25) > der2data(j),
                        j = j - 1;
                end
                [trash max1] = max(der2data(j-25:j));
                max1 = max1 + j - 25;
        else
                [trash max1] = max(der2data(i-25:i));
                max1 = max1 + i - 25;
        end

%Check to make sure the maximum is greater than 7 seconds
        if max1 < 700,
                max1 = 900;
        end

        fitTime(4) = max1;

%Find the second maximum from the start of the derivative and set
it equal to the 3rd fit time
        maxlow = critpoint;
        if der2data(maxlow + 25) <= der2data(maxlow),
                [trash maxlownew] = max(der2data(maxlow:maxlow +
25));
        else
                while der2data(maxlow + 25) > der2data(maxlow),
                        maxlow = maxlow + 1;
                end
                [trash maxlownew] = max(der2data(maxlow:maxlow +
25));
        end

        maxlow = maxlownew + maxlow;
        fitTime(3) = maxlow;
        fitTime = fitTime *.01;
```

11-6 plotgr.m

```
%This script file checks to see what graphs the user wants and then
subplots the graphs
%accordingly so that the space on the screen is maximized for
efficiency

if (strcmp(get(gxlin,'Checked'),'on') |
strcmp(get(gylin,'Checked'),'on') |
strcmp(get(grlin,'Checked'),'on')) &
(strcmp(get(gxlog,'Checked'),'on') |
```

```
strcmp(get(gylog,'Checked'),'on') |
strcmp(get(grlog,'Checked'),'on'))


        subplot(2,1,1);
        if strcmp(get(gxlin,'Checked'),'on'),
                plot(tau,X,'b'), hold on;
                if strcmp(get(greg,'Checked'),'on'),

        plot(tau1Xplot,linex1,'c',tau2Xplot,linex2,'c'),hold on;

        plot(fitTimeX,X(fitTimeX/.01),'b*',critpointX*.01,X(critpoin
tX),'cx'), hold on;
                end
        end
        if strcmp(get(gylin,'Checked'),'on'),
                plot(tau, Y, 'y'), hold on;
                if strcmp(get(greg,'Checked'),'on'),

        plot(tau1Yplot,liney1,'g',tau2Yplot,liney2,'g'), hold on;

        plot(fitTimeY,Y(fitTimeY/.01),'yo',critpointY*.01,Y(critpoin
tY),'gx'), hold on;
                end
        end
        if strcmp(get(grlin,'Checked'),'on'),
                plot(tau, R, 'r'), hold on;
                if strcmp(get(greg,'Checked'),'on'),

        plot(tau1Rplot,liner1,'m',tau2Rplot,liner2,'m'), hold on;

        plot(fitTimeR,R(fitTimeR/.01),'r*',critpointR*.01,R(critpoin
tR),'mx'), hold on;
                end
        end
        title('Linear Stabilogram-Diffusion plot')
        text(8.5,10,'X','color','b');
        text(9,10,'Y','color','y');
        text(9.5,10,'R','color','r');
        axis([0 10 0 max(R)+10]);

%Plot loglog graphs
        subplot(2,1,2);
        if strcmp(get(gxlog,'Checked'),'on'),
                loglog(tau,X,'b'),hold on;
                if strcmp(get(greg,'Checked'),'on'),

        loglog(tau1Xplot,linex1log,'c',tau2Xplot,linex2log,'c'),hold
on;

        loglog(fitTimeX,X(fitTimeX/.01),'b*',critpointX*.01,X(critpo
intX),'cx'), hold on;
                end
        end
        if strcmp(get(gylog,'Checked'),'on'),
                loglog(tau, Y, 'y'), hold on;
                if strcmp(get(greg,'Checked'),'on'),
```

```
        loglog(tau1Yplot,liney1log,'g',tau2Yplot,liney2log,'g'),hold
on;

        loglog(fitTimeY,Y(fitTimeY/.01),'yo',critpointY*.01,Y(critpo
intY),'gx'), hold on;
                end
        end
        if strcmp(get(grlog,'Checked'),'on'),
                loglog(tau, R, 'r'), hold on;
                if strcmp(get(greg,'Checked'),'on'),

        loglog(tau1Rplot,liner1log,'m',tau2Rplot,liner2log,'m'),
hold on;

        loglog(fitTimeR,R(fitTimeR/.01),'r*',critpointR*.01,R(critpo
intR),'mx'), hold on;
                end
        end
        axis([.01 10 .1 150]);
        xlabel('time interval (s)');
        zoom on
        title('Log-log Stabilogram-Diffusion plot')
        text(3,.2,'X','color','b');
        text(5,.2,'Y','color','y');
        text(7,.2,'R','color','r');

elseif (strcmp(get(gxlin,'Checked'),'on') |
strcmp(get(gylin,'Checked'),'on') |
strcmp(get(grlin,'Checked'),'on')) |
(strcmp(get(gxlog,'Checked'),'on') |
strcmp(get(gylog,'Checked'),'on') |
strcmp(get(grlog,'Checked'),'on'))

        if strcmp(get(gxlin,'Checked'),'on'),
                title('Linear Stabilogram-Diffusion plot')
                text(8.5,5,'X','color','b');
                text(9,5,'Y','color','y');
                text(9.5,5,'R','color','r');
                plot(tau,X,'b'), hold on;
                if strcmp(get(greg,'Checked'),'on'),

        plot(tau1Xplot,linex1,'c',tau2Xplot,linex2,'c'),hold on;

        plot(fitTimeX,X(fitTimeX/.01),'b*',critpointX*.01,X(critpoin
tX),'cx'), hold on;
                end
        end
        if strcmp(get(gylin,'Checked'),'on'),
                title('Linear Stabilogram-Diffusion plot')
                text(8.5,5,'X','color','b');
                text(9,5,'Y','color','y');
                text(9.5,5,'R','color','r');
                plot(tau, Y, 'y'), hold on;
                if strcmp(get(greg,'Checked'),'on'),

        plot(tau1Yplot,liney1,'g',tau2Yplot,liney2,'g'), hold on;
```

```
        plot(fitTimeY,Y(fitTimeY/.01),'yo',critpointY*.01,Y(critpoin
tY),'gx'),hold on;
                end
        end
        if strcmp(get(grlin,'Checked'),'on'),
                title('Linear Stabilogram-Diffusion plot')
                text(8.5,5,'X','color','b');
                text(9,5,'Y','color','y');
                text(9.5,5,'R','color','r');
                plot(tau, R, 'r'), hold on;
                if strcmp(get(greg,'Checked'),'on'),

        plot(tau1Rplot,liner1,'m',tau2Rplot,liner2,'m'), hold on;

        plot(fitTimeR,R(fitTimeR/.01),'r*',critpointR*.01,R(critpoin
tR),'mx'), hold on;
                end
        end

%Plot loglog graphs
        if strcmp(get(gxlog,'Checked'),'on'),
                title('Log-log Stabilogram-Diffusion plot')
                axis([.01 10 .1 150]);
                text(3,.2,'X','color','b');
                text(5,.2,'Y','color','y');
                text(7,.2,'R','color','r');
                loglog(tau,X,'b'),hold on;
                if strcmp(get(greg,'Checked'),'on'),

        loglog(tau1Xplot,linex1log,'c',tau2Xplot,linex2log,'c'),hold
on;

        loglog(fitTimeX,X(fitTimeX/.01),'b*',critpointX*.01,X(critpo
intX),'cx'), hold on;
                end
        end
        if strcmp(get(gylog,'Checked'),'on'),
                title('Log-log Stabilogram-Diffusion plot')
                axis([.01 10 .1 150]);
                text(3,.2,'X','color','b');
                text(5,.2,'Y','color','y');
                text(7,.2,'R','color','r');
                loglog(tau, Y, 'y'), hold on;
                if strcmp(get(greg,'Checked'),'on'),

        loglog(tau1Yplot,liney1log,'g',tau2Yplot,liney2log,'g'),hold
on;

        loglog(fitTimeY,Y(fitTimeY/.01),'yo',critpointY*.01,Y(critpo
intY),'gx'), hold on;
                end
        end
        if strcmp(get(grlog,'Checked'),'on'),
                title('Log-log Stabilogram-Diffusion plot')
                axis([.01 10 .1 150]);
                text(3,.2,'X','color','b');
                text(5,.2,'Y','color','y');
```

```
                text(7,.2,'R','color','r');
                loglog(tau, R, 'r'), hold on;
                if strcmp(get(greg,'Checked'),'on'),

        loglog(tau1Rplot,liner1log,'m',tau2Rplot,liner2log,'m'),
hold on;

        loglog(fitTimeR,R(fitTimeR/.01),'r*',critpointR*.01,R(critpo
intR),'mx'), hold on;
                end
        end

        zoom on
        xlabel('time interval (s)');
end
```

11-7 plotmult.m

```
% This script file is used only when the user requests that
multiple df files are to be plotted
%simultaneously.  It loads the .df files and calls to the plotgr.m
script file so that the
%requested plots are graphed accordingly

%Get some parameters from the user's screen
set(greg,'Checked','off');
datafilesnum = str2num(get(filenum2,'String'));
datapatho = get(filepatho2,'String');
datapathc = get(filepathc2,'String');
datafiles = get(filedata2,'String');

for i=1:datafilesnum,          %Loop through datafiles: load .df
files, call to plotgr.m script file and plot in separate windows
what is requested for each file listed on the user's screen
                figure;
                data = strtok(datafiles);
                load([datapatho,data,'.df'])
                datafiles = strrep(datafiles,strtok(datafiles),' ');
        temp = eval(data);
        X = temp(:,1);
        Y = temp(:,2);
        [a, b] = size(X);
        tau = 0.01:.01:a*.01;
        X = X';
        Y = Y';
        R = X + Y;

        plotgr;
        axis([0 10 0 max(R)+10]);
        title(data);
end
```

11-8 plotraw.m

```
%This script file is used only when the user wants to look at the
raw .cp data files to make
```

```
%sure that data was actually collected


%Get parameter's from user's interactive window
set(greg,'Checked','off');
set(gxlog,'Checked','off');
set(gylog,'Checked','off');
set(grlog,'Checked','off');
datafilesnum = str2num(get(filenum2,'String'));
datapatho = get(filepatho2,'String');
datapathc = get(filepathc2,'String');
datafiles = get(filedata2,'String');
if strcmp(get(o3000,'Checked'),'on'),
        NPTS = 3000;
else
        NPTS = 6000;
end


tau  = 0.01:.01:NPTS*.01;
for i=1:datafilesnum,          %Loop through datafiles: load .cp
files, call to plotgr.m script file to plot what is requested
accordingly
                figure;
                data = strtok(datafiles);
                load([datapatho,data,'.cp'])
                datafiles = strrep(datafiles,strtok(datafiles),' ');
        temp = eval(data);
        X = temp(:,1);
        Y = temp(:,2);
        X = X';
        Y = Y';
        R = X + Y;
        plotgr;
        title(data);
end
```

11-9 readme.txt

```
Hi!
If you have any troubles you can contact me at: imandie@bu.edu.
This program assumes that all data has been sampled at 100Hz.  I am
presently working on a new version that can handle different
sampling frequencies.  This program runs in matlab and has been
used both in windows and in unix.  It was created in version 4.1
but has been running great in the new 5.1 version.  A few warning
messages have been noticed, however, they can just be ignored.
In addition to the basic matlab software, this program also
requires the signals toolbox to run.  The name of the program is
"cop".  I have also included seven other ".m" files that I wrote
that are utilized in the cop program.  Make sure the following
eight files are all in the same directory:
     cop.m     findtime.m   plotmult.m
     cpdf.m    sparam.m     plotraw.m
     datcp.m   plotgr.m
```

To run the program, simply change directories at the matlab prompt to the directory where all of the files are saved. Note, the data need not be saved in the same directory as the programs. Then, type "cop" at the next prompt and a new figure window will pop up. If the colors are obnoxious, just let me know and I can tell you how to fix them. I have noticed that different screens cause the colors to have slightly different shades that may become annoying after a while. You will notice some pull down menus at the top of the figure and a rectangular box in the top left corner of the figure. Click on the spaces in the "input options" box and enter the specified file directories with a slash at the end and the names of the data files without the extensions. Be sure to press return after you are finished filling the data for each category or the program will not register a new change in input. Note: the raw data files must be saved with a ".cp" extension. Next, go through the two pull down menus on the right and check the options that you want the computer to perform. After everything is set, go up to the "Analyze" pulldown menu and click on the "cp->df" button to convert the ".cp" raw data files to a ".df" stabilogram diffusion file. We usually take 10 trials (ie 10 ".cp" files) to produce 1 stabilogram diffusion plot. That is, 10 ".cp" files yield 1 ".df" file. However, this program is flexible and can handle a different number of data files so be sure you specify the number of trials in the rectangular box. This program also produces a ".dfp" file, the p of which stands for parameters.

This ".dfp" file is saved as follows: X critical point lag time
The value at this lag time
The slope/2 for the short term linear X
The slope/2 for the long term linear X
The slope/2 for the short term log X
The slope/2 for the long term log X
This pattern repeats for the Y values and finally for the R values where each parameter is on a separate line. (total of 18 parameters)!

It is assumed that the ".cp" files contain two colums of data, one of the X (mediolateral) coordinates and one of the Y (anterioposterior) coordinates. The ".df" files are saved in the same format. However, ther are only 1002 points in the files corresponding to the first 10 seconds of time lag of the stabilogram diffusion plot. Since the data was sampled at 100Hz, each data point in the file is assumed to be .01 sec after the one before. For 30 seconds of data collection, 3000 data points will be collected, and for 60 seconds of data collection, 6000 data points will be collected. Be sure you specify whether you have 60 seconds or 30 seconds of data unser the "options" pulldown menu. If by chance you have data files that have ".dat" extensions and are filled with numbers from an A/D board in units of mV then you can convert the numbers to units of mm and therefore ".cp" files by using the "dat->cp" ore "dat->df" buttons under the analyze pulldown menu.
If you want to view just the raw ".cp" data files, specify the correct information in the "input options" box and then go up to

the "graphics" pulldown menu and click on the "plot raw cp data" button and each trial will be plotted in a separate window. If you want to view the ".df" files at a later time, you can specify all of the information in the "input options" box and then click on the "plot multiple df files" button under the "graphics" pulldown menu. It is assumed that the files in the "input options" box have the ".df" extensions. These files will not be plotted with the regression lines. If the "show parameters" box is checked under the "random options" pulldown menu, the critical point, scaling exponents, and diffusion coefficients will be printed in the command window of matlab.

If you have any other questions, please do not hesitate to contact me.
Later
Andrea Stamp

Appendix 12: Ethical approval for experiment three

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UNIVERSITY OF LEEDS

08 May 2014

Mohammad Alshamrani
3.35 PhD Suite
Baines Wing
School of Healthcare
University of Leeds, LS2 9UT

Dear Mohammad

Ref no: SHREC/RP/296 – Amendment_3

Title: The effects of listening on postural control

Thank you for submitting the amendment to the above named project.

This has been reviewed and I can confirm on behalf of the School of Healthcare Research Ethics Committee (SHREC) that ethical approval is granted and the amendment may be implemented based on the documentation received at date of this letter.

Document	*Version*	*Date submitted*
Ethical Review Form	1	16.10.12
SHREC amendment Form	1	17.04.14
Participant Information sheet (V1.3)	1	17.04.14
Participant Recruitment poster (V1.1)	1	17.04.14

Ethical approval does not infer you have the right of access to any member of staff or student or documents and the premises of the University of Leeds. Nor does it imply any right of access to the premises of any other organisation, including clinical areas. SHREC takes no responsibility for you gaining access to staff, students and/or premises prior to, during or following your research activities.

Please note: You are expected to keep a record of all your approved documentation, as well as documents such as sample consent forms, and other documents relating to the study. This should be kept in your study file, and may be subject to an audit inspection. If your project is to be audited, you will be given at least 2 weeks notice.

It is our policy to remind everyone that it is your responsibility to comply with Health and Safety, Data Protection and any other legal and/or professional guidelines there may be.

The committee wishes you continued success with your project.

Yours sincerely

Kuldip Kaur Bharj

Dr Kuldip Bharj, OBE
Deputy Chair, School of Healthcare Research Ethics Committee

Appendix 13: Experiment three participant information sheet

Study title: Simulated hearing loss affects posture control

I would like to invite you to volunteer to take part in the above named research study but before you decide, please read the following information. It is important that you understand why this research study is being done and what it will involve before you volunteer to take part.

Please ask if anything is not clear or if you would like more information.

What is the purpose of this study?

Posture and balance control are often thought to be automatically controlled, requiring little active attention. However, recent research suggests that this may not be the case and that postural control may be affected in situations when attention is divided, for example, when multi-tasking. The aim of this research project is to investigate this further by investigating the effect of performing a listening task whilst performing a simultaneous postural control task.

Who is doing the study?

This study is being done by Mohammad Alshamrani, a PhD student based in School of Healthcare, University of Leeds. This work is being supervised by Drs Nick Thyer and Ruth Brooke, lecturers in Audiology within the School.

Who is being asked to participate?

We are asking individuals who are aged 18-60, who speak English as their first language and who have no history of hearing or balance problems to take part.

What will be involved if I take part in this study?

If you take part you will be required to attend a single test session which will last a maximum of 120 minutes. The session will be arranged during normal working hours at a time that is suitable to you and will take place within the Audiology Suite in the School of Healthcare, at the University of Leeds.

At the beginning of the test session you will be asked to complete a short hearing test which will involve listening to a series of beeps through headphones and telling the researcher whether or not you heard them. If any problems with your hearing are identified, or you report any balance problems, you will unfortunately not be able to take part in the study and you will be advised to visit your GP.

Following the hearing screen, you will be asked to complete short experiments (each last about 30 seconds and repeated 5 times) which will be performed in a random order. Experiment 1 will involve listening to different words delivered through headphones to opposing ears at the same time and telling the researcher what you heard. For experiment 2 you will be asked to stand on a low platform, also called a force plate, with eyes open and eyes closed. For experiment 3 you will be asked to complete experiment 1 and 2 simultaneously. In experiment 4 you will be asked to stand on a force plate with one foot in front of the other (heel to toe) with eyes open and eyes closed. For experiment 5 you will be asked to complete experiment 1 and 4 at the same time.

You will be provided with comfort breaks between each experiment.

What are the advantages and disadvantages of taking part?

You will need to give up some of your time to take part. By taking part you will help us gain an understanding of how sound and listening can affect our ability to maintain posture and balance. At the end of the session we will give you a £20 gift voucher to thank you for your time.

Can I withdraw from the study at any time?

Yes. Your participation is voluntary and so you can change your mind about taking part and withdraw your participation at any time (prior to and during the test session) without giving a reason. If you withdraw part way through the test session any data already collected will be deleted. However, it will not be possible to withdraw your data after the test session as all data will be saved and stored anonymously.

Will the data obtained in the study be confidential?

Yes, all information provided and data collected will be kept confidential. Your name and contact details, and the signed consent forms, will be kept confidential, stored in a secure place and only accessed by the researchers involved in the study. Personal information will not be linked to the test results in any way. Gender and age (not birth dates) will be recorded however; this will not be linked to any names. Test data will be anonymised and kept for 3 years.

What will happen to the results of the study?

The results of the experiments will be recorded, analysed and kept for 3 years prior to deletion. The results of the study will be written up in the form of a PhD thesis and may be written up and submitted for publication in an academic journal. All data used will be anonymised.

<u>What if there is a problem?</u>

If you have any complaints or concerns about the study you can speak to the clinic that referred you to the research team and they will do their best to answer your questions. Should you have a complaint about the way the study is being conducted, please contact my supervisor Dr Nicholas Thyer (Tel: 0113 343 1238; email: N.J.Thyer@leeds.ac.uk) or the Faculty Head of Research and Innovation Support Ms Clare Skinner (Tel: 0113 343 4897; email: c.e.skinner@leeds.ac.uk).

<u>Who has reviewed this study?</u>

Ethical approval has been granted by the School of Healthcare Research Ethics Committee (SHREC/RP/296).

If you agree to take part, would like more information or have any questions about the study please contact:

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Thank you for taking the time to read this information sheet.

Appendix 14: Hearing loss simulation code

14-1 control_impaired_ear.m

```
%% Extensive header updated Jan 2011.
%% modified Oct 09 to do multi-channel processing
%% hacked by MAS, Jun09 for 44.1k sampling,  new executables as well: same
old names
%% hacked by MAS, Oct08 to try to get more consistent results, for Eng
Dept/RNID accessibility project, Feb 2007.

%% (1) take wav file from "roots" variable, append calibration tone and
calibration speech-shaped noise at start of each channel signal
%% for each audiogram in array "audiogram_master", process file........
%% (2) send to MATLAB function that applies linear filtering to simulate
passage from free-field to cochlea
%%          (mainly a bass cut and a mid-range boost, does not increase
signal level)
%% (3) write cochlea-referenced signal as 16 bit binary file for processing
by DOS executable called by batch file
%% (4) impairment simulation is done in two stages: spectral smearing and
recruitment simulation
%% (5) re-call MATLAB function of (2) above that applies INVERSE linear
filtering from that in (2)
%%          (to simulate passage from cochlea to free-field, does increase
signal level)
%% (6) write processed signal out to wav file: filename is automatically
generated by adding the index of the audiogram to the input filename

%% REFERENCES:
%% Spectral smearing:
%% (a) T. Baer and B.C.J. Moore, Effects of spectral smearing on the
intelligibility of sentences in the presence of noise, J. Acoust. Soc. Am.
94: 1229-1241 (1993).
%% (b) T. Baer and B.C.J. Moore, Effects of spectral smearing on the
intelligibility of sentences in the presence of interfering speech, J.
Acoust. Soc. Am. 95: 2277-2280 (1994).
%% Recruitment simulation:
%% B.C.J. Moore and B.R. Glasberg, Simulation of the effects of loudness
recruitment and threshold elevation on the intelligibility of speech in
quiet and in a background of speech, J. Acoust. Soc. Am. 94: 2050-2062
(1993).
%% Filtering to equalise signal cor cochlear presentation:
%% M.A. Stone and B.C.J. Moore, Tolerable hearing-aid delays. I. Estimation
of limits imposed by the auditory path alone using simulated hearing
losses, Ear Hear. 20: 182-192 (1999).

%% In the original recruitment simulation, the filter bandwidth was set to
3 times broader than "normal" (ie 3*ERBn).  This sort of width is suitable
for severe losses, where
%% signal bandwidth is usually limited, hence the original software was
written for 16 kHz samplerate, and used 13 overlapping filters.  This
version, with samplerate set to 44.1 kHz
%% was intended for more moderate losses, hence the recruitment filterbank
comprises 28 filters with *2 broadening (see "whole_ear_simulate.m").
however, the degree of spectral
%% smearing, applied uniformly across frequencies, depends on the average
audiogram between 2 and 8 kHz.  The higher the degree of smearing, the more
"fuzziness" it applies to the
%% signal, by bringing up the level of noise between signal components.  To
a first order, this can be regarded as a form of degradation of IHC signal
quality.
```

```
%% Because of the non-linearity of the processing, and the overlap of the filters, this software does not (and cannot) provide an "exact" simulation, more a qualitative feel
%% for the sorts of problems to be experienced.  So if a calibration tone comes out 1-2 dB away from where one expected it to, then do not be surprised.

%% CAVEATS:
%% (1) because the cochlear signal is reduced at low and high frequencies due to middle-ear filtering, then the signal sent to the recruitment simulation as 16 bits can suffer
%% quantising (especially for high degres of simulated losses).  Once the inverse filter (cochlea to free field) is applied, effectively boosting low & high frequencies, then random noise
%% will appear in the processed signal.  It may then be wise to limit the frequency range of the simulation, so as not to excite this effect.  Hence parameter [UPPER_CUTOFF_Hz].
%% (2) the recruitment simulation was written in C and has been compiled as a win-32, DOS-like executable.  Since input and output is in 16 bits binary files, some quantising/dynamic
%% range problems can arise here.  MAS updated TBs smearing code so that it all ran in MATALB, but same has yet to be done for recruitment simulation.
%% A practical implication of the is that the HEADROOM_dB variable, which reduces the number of clippings introduces quantising noise.  So if a wide range of audiograms are
%% to be simulated, it may be better to split the task up and use two different values of HEADROOM_dB.

%% Calibration method:
%% the system assumes that within a digital wav file, an RMS of [REF_RMSdB] translates to an equivalent real world SPL of [CALIB_dBSPL]
%% variable [target_roots_SPL] contains the dB SPL equivalents of what we have scaled the input files to, ie
%% the levels at which we wish to run simulations for each file in [roots]. Not all processing occurs (a) in MATLAB and (b) in floating point,
%% so we may encounter a 16 bit headroom problem when we write the audio data as binary files for the call to 'C' executables (filterbank & recruitment).
%% A variable [HEADROOM_dB] allows us to scale down the signals before sending out to these executables.  Even so, [whole_ear_simulate]
%% checks to see if clipping will occur, and prints a warning if < 20 clips occur, or stops if >= 20.
%% [measure_rms] is a sophisticated method of measuring RMS in a file: it splits the signal up into 10 msec durations, performs a histogram of levels,
%% calculates an apporximate RMS, and then uses that RMS to calculate a threshold level in the histogram and then re-measures the RMS only using
%% those durations whose individual RMS exceed that threshold.  In this way it avoids that the RMS measure is influenced by long perioods of near-silence.

%% The program measures the file digital RMS from channel 1.
%% Each channel being processed has inserted on its start a 500 Hz tone, followed by a short burst of SII (ANSI-1997) shaped noise:
%% SII shape: Flat spectrum level to 500 Hz, then sloping off 9dB/octave above that.  The RMS level of noise and tone are set to [REF_RMSdB]

%% M.A.Stone, T.Baer, Psychoacoutics group, Dept Experimental Psychology, University of Cambridge, Feb 2007

%% user needs to specify:
%% 'audiogram_master' : hearing loss at each of entries in 'audiogram_cfs'
%% 'REF_RMSdB' : RMS calculation of output signal file in dB relative to unity (+/-1 square wave has RMS=0dB)
%% 'roots': names of input WAV files for processing
```

```
%%  (OPTIONAL) file_no     : number (0-99) to uniquely identify file when
it goes through batchfile-called C-executables for impairment simulation

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%
%%%% suggested file calibration levels to give enough headroom for slightly
louder presentations
REF_RMSdB = -12; %% sets peak of output file so that no clipping occurs,
set so that equiv0dBfileSPL > 100dB for LOUD input files
CALIB_dBSPL = 90; %% what RMS of speech file translates to in real world
(unweighted)
equiv0dBfileSPL = CALIB_dBSPL-REF_RMSdB; %% what 0dB file signal would
translate to in dB SPL: constant for whole_ear_simulate function

%% iff some gain is added to the signal to simulate something like a
hearing aid, then you may need to allow extra headroom
%% through the impairment simulation.  This can be done by setting the
variable 'HEADROOM' to be > 0dB.  Normally, impairment results in
%% reduced signal level so don't need to allow extra HEADROOM.
Master_Fs = 44100; %% IMPERATIVE: only works at 44.1 kHz, checked later
against read .wav files
winlen=2*floor(.0015*Master_Fs);  %%% small window to design low-pass FIR
UPPER_CUTOFF_Hz = 15525; %% User-selectable cut off frequency of low-pass
filter at end of simulations: prevents excessive processing noise at high
frequencies.
%% now flat to Cutoff freq, tails below -80 dB
lpf44d1  = fir1(winlen, UPPER_CUTOFF_Hz/(Master_Fs/2), kaiser(winlen+1,8));
%% suitable lpf for signals later converted to MP3, flat to 15 kHz.
NSigChans = 1; %% number of channels in signal file.
audiogram_cfs = [.125, .25, .5, .75, 1, 1.5, 2, 3, 4, 6, 8, 10, 12, 14,
16]*1000;%% Hz, standard audiogram frequencies
%%audiogram_master has one entry for each of the corresponding frequencies
in audiogram_cfs
%% need extra values for 10:2:16 kHz, NOT SENSIBLE TO EXCEED 90 dB HL
%% however if these were not available, simulation extends highest
available to subsequent freqs
audiogram_master = [ %% need 15 values to span .125 to 16 kHz: use several
rows if more that one simulation required
    %% .125, .25, .5, .75,  1, 1.5, 2,  3,  4,  6,  8, 10, 12, 14, 16
    90,   90,   90,   90,  90,  90,  90,  90,  90,  90,  90,  90,  90,  90,
90; %% Reference processing
   %100,  100,  100,  100, 100, 100, 100, 100, 100, 100, 100, 100, 70, 70,
70; %% NEW acute: night at the clubber : see MAS workbook LSHTM (rear of
High Freq Audibil) 30.10.09
    %80,   80,   80,   80,  80,  80,  110, 80, 80, 80, 80, 80, 80, 80, 80;
%% New chronic NIHL :18-24 yrs olds: see MAS workbook LSHTM (rear of High
Freq Audibil) 30.10.09
    ];

if max(max(audiogram_master)) > 90, error('Suggest you limit audiogram max
to 90 dB HL, otherwise things go wrong'); end
nmaster =min (size(audiogram_master));  %% number of audiograms to run

roots = [ %% root names of files, all made standard length
    'WN';
    %'list2';
    %'ElectroSonicBeat2';
    %'ElectroSoncCymbls';
    ];
NRoots = min(size(roots)); %% number of files to process
%%% translate sensible levels from "roots" file RMS
target_roots_SPL = [90]; %% parallel indexing to "roots", intended SPL of
signal in each file
HEADROOM_dB = 6;  %% (>0) dB Headroom to allow through whole_ear_simulate
to avoid clipping during 16 bit file reads & writes
```

```
file_no = 0;
DIAGNOSTIC = 0; %% to suppress loads of unnecessary screen output from
gen_sii_nse,
for ix_roots = 1:NRoots  % index to file names
    fprintf(1,'\nStarting new file.......... \n');
    for ix_degree = 1:nmaster %% index through audiogram_master
        audiogram = audiogram_master(ix_degree,:);
        rootsfile = roots(ix_roots,:);
        ipfilename = rootsfile;
        opfilename0 = strcat(rootsfile,'0'); %% input file, but only level
adjusted, as processed
        opfilename = strcat(rootsfile,num2str(ix_degree)); %% input file,
level adjusted, and processed
        file_no= file_no + 1; %% number (0-99) to uniquely identify file
when it goes through batchfile-called C-executables for impairment
simulation

        HEADROOM = 10.^(HEADROOM_dB/20); %% possibility to send reduced
signal level into [whole_ear_simulate] eg after hearing aid gain: internal
reference levels track this change
        [tgt_lr, Fs, nbits] = wavread(ipfilename); %% re-read every time
round since we adjust the level to be correct for output, even unproc
        tgt = tgt_lr(:,1); %% reference RMS off one channel only
        %%%%%  need to know file RMS, and then call that a certain level in
SPL: needs some form of pre-measuring.
        [calced_rms, indexkeys ] = measure_rms(tgt, Master_Fs, -12);  %%
3rd arg is dB_rel_rms (how far below)
        %% rescale input data for (a) rewriting as master file for website
simulation and (b) input to subsequent stages of processing
        change_dB = target_roots_SPL(ix_roots) - (equiv0dBfileSPL +
20*log10(calced_rms));
        tgt_lr = tgt_lr * 10.^(.05*change_dB);  %% both channels identical
        %% and a quick cross-check, using same time samples to calculate
RMS as before.
        tgt = tgt * 10.^(.05*change_dB);  %% both channels identical
        new_rms = sqrt(sum(tgt(indexkeys).^2/length(indexkeys)));
        clear indexkeys
        fprintf(1,'\nimpaired_ear: file [%s] measure RMS as %5.1fdB (wanted
%3.0f SPL)', ipfilename, equiv0dBfileSPL+20*log10(new_rms),
target_roots_SPL(ix_roots) );
        if (Fs ~= Master_Fs), error(sprintf('impaired_ear: HELP: Just read
.wav file where Fs was not what I have assumed (%d != %d)', Fs,
Master_Fs)); end

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
        %% calibration noise to check result of all processing, inserted at
start of signal, with silence around it.
        nse_burst = gen_sii_shape(2, Fs, DIAGNOSTIC);
        nse_burst = nse_burst * (10^(.05*REF_RMSdB)) /
(sqrt(sum(nse_burst.^2)./length(nse_burst)));

        %% nearest channel centre frequency of recruitment simulation to
500 Hz: generate tone burst with same RMS as original speech signal file
        fburst = 520; %% near centre of broadened auditory filter, used as
calibration tone
        ton_burst =
1.4142*(10^(.05*REF_RMSdB))*sin(2*pi*fburst*(1:2*Fs)/Fs);%% 0.5sec
        silence = zeros(1,(50*Fs/1000));
        for ix_lr = 1:NSigChans
            tgt = tgt_lr(:,ix_lr); %% process one channel at a time,
possibly put in processing limit here
```

```
          %% so also pad around noise&tone bursts with 50 msec of silence,
& perform implicit change of vector shape of target signal
            ff_sig = [silence ton_burst silence nse_burst silence tgt'
silence]; %% ton_burst is 0.5sec, nse_burst is 2 secs, silence 50msec
Original Code
            clear  tgt

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
            %% linear filtering and cochlear recruitment performed in
single function

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
            %% function ff_sig_out = whole_ear_simulate(ff_sig_in, Fs,
headroom, file_no, audiogram, equiv0dBfileSPL);
            ff_sigrc = whole_ear_simulate(ff_sig, Fs, HEADROOM, file_no,
audiogram_cfs, audiogram, equiv0dBfileSPL);
            clear ff_sig
            if ix_lr ==1
                ff_sigrc_lr = zeros(length(ff_sigrc),2);
            end
            %% 30-Oct-09 low-pass to something sensible, prevents
exaggeration of > 15 kHz
            ff_sigrc = filter(lpf44d1, 1, ff_sigrc);
            fprintf(1,'\ncontrol_impaired_ear: there are potentially %d
overflows (needs %3.1fdB more headroom)\n',length(find(abs(ff_sigrc)> (1-
1e-5))), 20*log10(max(abs([ff_sigrc; eps])))); %% eps fix for zero log
problem
            ff_sigrc_lr(:,ix_lr) = ff_sigrc; %% reassemble stero file
        end %%ix_lr
        clear tgt_lr ff_sigrc
        wavwrite(ff_sigrc_lr, Fs, 16, opfilename);
        clear ff_sigrc_lr
    end %% for ix_degree = m:n %% index through audiogram_master
end %% for ix_roots = 1:N
fprintf(1,'\nFinished \t control_impaired_ear\n\n');
```

14-2 whole_ear_simulate.m

```
%% Nov09, introduce clear statements to reduce memory load for VEY long
file processing
%% to replace large chunk of code impaired_ear.m
%% ff_sig_out : output signal post processing, assumed N*1, range -1:1-
epsilon
%% ff_sig_in  : input signal post processing, assumed N*1,  range -1:1-
epsilon
%% Fs         : sampling rate
%% headroom   : modulus of signal level that will translate to full scale
digital in 16 bit binary file to C-executable code
%% file_no    : unique number for each time impairment simulation is
called: range 0-99  due to old DOS 8.3 filename length limits
%% audiogram_cfs: frequency grid on which audiogram is specified
%% audiogram  : impairment in dB HL at each frequency in audiogram_cfs
%% equiv0dBfileSPL: equivalent level in dB SPL for file RMS of 0dB, ie full
aplitude square wave (RMS per sample)

function ff_sig_out = whole_ear_simulate(ff_sig_in, Fs, headroom, file_no,
audiogram_cfs, audiogram, equiv0dBfileSPL);
FORWARD = 1; %% direction signal for ff_cochlea_filt
BACKWARD= -FORWARD; %% inverse direction signal for ff_cochlea_filt
CATCH_UP = 105; %% dBHL where impaired catches up with normal
```

```
%% recruitment simulation comes with 2 degrees of broadening of auditory
filters: different set of centre freqs between simulations.
%% do not use these for 44.1k
%% recruit_cfs_severe =
[100,190,306,452,640,879,1184,1572,2067,2698,3503,4529,5837]; %%  13
channel cfs for recruitment simulation, x3 broadening, fs=16kHz
%% June 09: updated for 44.1 k sampling, 28 channels.
recruit_cfs_moderate = [  65.8,    95.6, 134.9, 185.6,  249.4, 328.2,
423.9, 538.6,  674.3,  833.8, 1020.0,...
                1238.7, 1495.7,  1797.8,  2152.7,  2569.9,  3060.1, 3636.1,
4313.1, 5108.6, 6043.4, 7142.0,...
                8433.0, 9950.1, 11733.0, 13828.1, 16290.2, 19183.5]; %% 28
channels, 44.1 kHz, *2 broadening,%% NEW Jun09: channel cfs for recruitment
simulation, x2 broadening

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%% check and categorise audiogram: currently ALWAYS recruit with x2
broadening: it's the smearing that changes
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
impairment_freqs = find ( (audiogram_cfs >= 2000) & (audiogram_cfs <=
8000));
impairment_degree= mean(audiogram(impairment_freqs));  %% calculate mean
hearing loss between 1 & 4 kHz

%% impairment degree just affects smearing simulation, not recruitment
%% (assuming we do not have too much SEVERE losses present)
if impairment_degree > 56  %% set flag for degree of processing: should
never get here
    degree_txt = 'SEVERE';
    severe_not_moderate = 1; %% flag
elseif impairment_degree > 35  %% set flag for degree of processing
    degree_txt = 'MODERATE';
    severe_not_moderate = 0; %% flag
elseif impairment_degree > 15 %% set flag for degree of processing
    degree_txt = 'MILD';
    severe_not_moderate = -1; %% flag
else %% set flag for degree of processing
    degree_txt = 'NOTHING';
    severe_not_moderate = -2; %% flag
end
fprintf(1,'\nwhole_ear_simulate: ********* simulate %s smearing *********',
degree_txt);
recruit_cfs = recruit_cfs_moderate;  %% STILL assumes use of wfilt_x2 &
recruit_x2 in recruitment simulation

%%if audiogram_cfs(end) < recruit_cfs(end)
%%    error(sprintf('**** whole_ear_simulate: need to specify audiogram up
to at least %f Hz\n ****',recruit_cfs(end)));
%%end
%%  function op_sig = ff_cochlea_filt(ip_sig, FORWARD, Fs); %% or BACKWARD
coch_sig = ff_cochlea_filt(ff_sig_in, FORWARD, Fs);
clear  ff_sig_in
exceed = length(find(abs(coch_sig) > headroom)); %% check for possible
exceedances in later file write
maxheadroom = 20*log10(headroom./max(abs(coch_sig)));
fprintf(1,'\nwhole_ear_simulate: ********* File number %d, HEADROOM safe
for another %5.1fdB *********\n', file_no, maxheadroom);
if exceed > 20,
    fprintf(1,'\nwhole_ear_simulate: ********* File number %d, potentially
%d clips *********\n', file_no, exceed);
    error('CHANGE [headroom] VARIABLE');
end
```

```
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%
%% do Tom Baer's work here, hacked by MAS Oct 08 to
%% (a) vary degree of smearing according to degree of loss and
%% (b) use MATLAB arrays as input and output medium for call to smearing
routine
%% NHMC mod to call Tom's code
do_smear = 1;
if severe_not_moderate > 0
%%    fsmear = makesmearmat3(3, 3); %% Tom's arguments for symmetric severe
smearing
    fsmear = makesmearmat3(4, 2); %% MAS: asymmetric severe smearing
elseif ~severe_not_moderate  %% use lower degree of smearing
%%    fsmear = makesmearmat3(2, 2); %% Tom's arguments for symmetric
moderate smearing
    fsmear = makesmearmat3(2.4, 1.6); %% MAS asymmetric moderate smearing
elseif severe_not_moderate == -1 %% use very low degree of smearing
%%    fsmear = makesmearmat3(1.35, 1.35); %% Tom's arguments for symmetric
mild smearing
    fsmear = makesmearmat3(1.6, 1.1); %% MAS asymmetric mild smearing
elseif severe_not_moderate == -2 %% use NO smearing
    fsmear = makesmearmat3(1.001, 1.001); %% Tom's arguments for symmetric
mild smearing
    do_smear = 0;
end
%% just a check here to see if there is any change in RMS across smearing
routine
[pre_rms, key] = measure_rms(coch_sig(1+2.75*Fs:end), Fs, -12); %% exclude
first 2.75 secs from RMS measure, where calibration signals and silence
reside
if do_smear %% check on what happens here
    coch_sig_smr = smear3(coch_sig', fsmear); %% call to perform smearing:
note change of shape of array
    coch_sig_smr = coch_sig_smr'; %% and re-change shape of array
else
    coch_sig_smr = coch_sig; %% NO smearing: NO change of shape of array
end
clear  coch_sig
%%%
post_rmsdB = 10*log10(sum(coch_sig_smr((2.75*Fs+key)).^2)/length(key)); %%
measure exactly same samples at output as at input (assumes no time shift)
fprintf(1,'\nwhole_ear_simulate: RMS around Smear: Pre %5.1f, Post %5.1f
dB', 20*log10(pre_rms), post_rmsdB);
%% end of NHMC/MAS mod
%%%%%%%%%%%%%%%%%%%%%%%
temp_filename = sprintf('coch%d',file_no); %% NB old DOS 8.3 filename
length limitations
%%%%%%%%%%% Move to relevant directory for recruitment simulation
eval('cd RecruitSmear'); %% subdirectory in which the executables sit
fh = fopen(temp_filename, 'w');
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%
%% during file operation, reduce by headroom (ie calib level (-d option in
recruit_x2) MUST track this): also scale for binary file writing rather
than  wavlevels (-1:1)
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%
%% NHMC now calls coch_sig_smr rather than coch_sig
if fh > 0, fwrite(fh, coch_sig_smr *(32768/headroom), 'short'); fclose(fh);
clear temp_filename; %% write binary file, appropriate for recruitment
simulation prog
else
```

```
    error('whole_ear_simulate: Invalid file handle for file open: pre-
recruitment');
end
clear  coch_sig_smr
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%% Now generate relevant audiogram-derived control file for
recruitment program
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
fh = fopen (sprintf('%d.rec',file_no),'wt'); %% GENERATE control file for
recruitment simulation prog
%% each line contains control for each channel of recruitment sim.,
contains expansion coefficient followed by channel gain (here 0 dB since
aid function done elswhere)
if fh > 0,  %% with recruitment simulation, control file requires triplets
of [chan_exp_ratio eqloud_catchupdB channel_gaindB]
    for ix_rec = 1:length(recruit_cfs) %% start so as not to get
extrapolation from audiogram_cfs: produces error
        %% each line consists of expansion ratio, recruitment catch-up
level (dB SPL) and channel gain (dB)
        if recruit_cfs(ix_rec) < audiogram_cfs(1)
            fprintf(fh,'%4.2f %s 0\n',CATCH_UP/(CATCH_UP-
audiogram(1)),num2str(CATCH_UP)); %% extend audiogram
            %% (continued comment) flat below lowest freq measured
        elseif recruit_cfs(ix_rec) > audiogram_cfs(end)
            fprintf(fh,'%4.2f %s 0\n',CATCH_UP/(CATCH_UP-
audiogram(end)),num2str(CATCH_UP)); %% extend audiogram
            %% (continued comment) flat above highest freq measured
        else %% in the sensible region
            audiog_cf = interp1(audiogram_cfs, audiogram,
recruit_cfs(ix_rec),'linear');
            fprintf(fh,'%4.2f %s 0\n',CATCH_UP/(CATCH_UP-
audiog_cf),num2str(CATCH_UP)); %% assumes catch-up at CATCH_UP dB (typ 100-
105)
        end
    end
    fclose(fh);
else
    error('whole_ear_simulate: Invalid file handle for recruitment control
file');
end
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%% Now call batch file to run C executables externally
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
eval( sprintf('!do_sim_x2 %d %4.1f', file_no, (equiv0dBfileSPL +
20*log10(headroom)) ) );
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%% Now transform from impaired cochlea to free-field referenced
signal:
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
temp_filename = sprintf('coch%drc',file_no); %% file produced by
recruitment simulation NB old DOS 8.3 filename length limitations

fh = fopen(temp_filename, 'rb'); %% read file from simulation
%%fh = fopen(temp_filename, 'r'); %% FLOAT file : read file from simulation
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%% during file operation, increase by headroom : also re-scale for binary
file reading and going back to wavlevels (-1:1)
```

```
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%keyboard
if fh > 0, coch_sigr = fread(fh, inf, 'short') * headroom/32768 ;
fclose(fh);  %% read & then delete binary file
    eval(sprintf('!del %s',temp_filename)); clear temp_filename;
else
    error('whole_ear_simulate: Invalid file handle for file open: post-
recruitment');
end
%%%%%%%%%%% Return from relevant directory for recruitment simulation
eval('cd ..');
ff_sig_out = ff_cochlea_filt(coch_sigr, BACKWARD, Fs); %% and reverse
transform to free-field sig
```

14-3 audfilt.m

```
function filter=audfilt(rl, ru)

% filter=audfilt(rl,ru) - calculate an auditory filter array
%  rl = broadening factor on the lower side
%  ru = broadening factor on the upper side
%
% array size has been set to 128
% sampling frequency has been set to 16000

size=128;
sampfreq=16000;

filter=zeros(size,size);
filter(1,1)=1.;
% uncomment or comment the following if/if-not dividing by the erb to
remove spectral tilt from the excitation pattern
filter(1,1)=filter(1,1)/((rl+ru)/2);

for i=2:1:size
     fhz=(i-1)*sampfreq/(2*size);
     erbhz=24.7*((fhz*.00437)+1.0);
     pl=4.0*fhz/(erbhz*rl);
     pu=4.0*fhz/(erbhz*ru);
     j=[1:i-1];                            % for lower side of the filter
          g(j)=abs((i-j)/(i-1));
          filter(i,j)=(1+(pl*g(j))).*exp(-1.0*pl*g(j));
     j=[i:size];                           % for upper side of the filter
and center
          g(j)=abs((i-j)/(i-1));
          filter(i,j)=(1+(pu*g(j))).*exp(-1.0*pu*g(j));
          % uncomment the following if dividing by the erb to remove the
spectral tilt
          filter(i,:)=filter(i,:)/(erbhz*(rl+ru)/(2*24.7));
end
```

14-4 ff_cochlea_filt.m

```
%% free-field to cochlea filter forwards or backward direction, depends on
'direction' switch.
%% ff_cochlea.m: need to reference signals to cochlea and then back to
free-field
%% NO LONGER via 2 steps.  ff to eardrum and then via middle ear: use same
length FIR  5-12-97.
function op_sig = ff_cochlea_filt(ip_sig, direction, fs) ;
% simulate middle and outer ear transfer functions
%%%%%%%%%%%%%%% original %%%%%%%%%%%%%%%%%%%%%%%%%
% Table data file, from BRG May 2001
%-----------------------------------------------------------------------
```

```
%-  This is the current file 2008
%----------------------------------------------------------------------

hz = [ 0.,...
      20.,    25.,   31.5,   40.,    50.,    63.,   80.,  100.,  125.,
160.,  200.,   250.,   315.,   400.,  500.,  630.,...
     750.,   800., 1000.,  1250.,  1500.,  1600., 2000., 2500., 3000.,
3150., 4000.,  5000.,  6000.,  6300., 8000., 9000.,...
     10000., 11200.,12500., 14000., 15000., 16000.,20000.,48000];

%---------------------------------------------------------------------------
----

%- This is the AES paper midear correction with slight increases in the
corrections of 80 Hz & below
%- USED TO CALC FILES ff.32k, df.32k, hd580.32k
% Killion JASA 63 1501-1509  (1978)

midear=[ 50.0,...
      39.6,   32.0,  25.85,  21.4,   18.5,  15.9,  14.1,  12.4,  11.0,
9.6,   8.3,    7.4,    6.2,   4.8,   3.8,   3.3,...
       2.9,    2.6,   2.6,    4.5,    5.4,   6.1,   8.5,  10.4,   7.3,
7.0,   6.6,    7.0,    9.2,  10.2,  12.2,  10.8,...
      10.1,   12.7,  15.0,   18.2,   23.8,  32.3,  50.0,  50.0];

%---------------------------------------------------------------------------
----

% Free field (frontal)FF_ED correction for threshold  (was ISO std Table 1
- 4.2 dB)
% i.e. relative to  0.0 dB at 1000 Hz, Shaw 1974

ff_ed= [0.0,...
       0.0,    0.0,   0.0,    0.0,    0.0,    0.0,   0.0,   0.0,   0.1,
0.3,   0.5,    0.9,    1.4,    1.6,   1.7,   2.5,...
       2.7,    2.6,   2.6,    3.2,    5.2,    6.6,  12.0,  16.8,  15.3,
15.2,  14.2,   10.7,    7.1,    6.4,   1.8,  -0.9,...
      -1.6,    1.9,   4.9,    2.0,   -2.0,    2.5,   2.5,   2.5];

fprintf(1,'\nff_cochlea_filt:\tperforming outer/middle ear corrections');
siglen = length(ip_sig);
%%% make sure that response goes only up to fs/2
nyquist = fs/2;
ixf_useful = find(hz < nyquist);
hz_used = hz(ixf_useful);
hz_used = [hz_used nyquist];
% sig from free field to cochlea: 0 dB at 1kHz
corrn = ff_ed - midear ;
last_corrn = interp1(hz, corrn, nyquist);  %% generate synthetic response
at Nyquist
corrn_used = [corrn(ixf_useful) last_corrn];
corrn_used = exp(direction*log(10)*corrn_used/20); %% NB direction switches
from ff->coch to coch->ff

%% 23 msec window to do reasonable job down to about 100 Hz, scales with
fs, falls over with longer windows in fir2
n_wdw = 2*floor((fs/16e3)*368/2);
b = fir2(n_wdw, hz_used./nyquist, corrn_used, kaiser(n_wdw+1,4)); % f= 1 is
Nyquist
%%freqz(b,1,2048,fs); grid on; set(gca,'ylim',[-40 20],'xlim',[50
nyquist],'xscale','log');pause;
op_sig = filter(b, 1, ip_sig);
```

14-5 gen_sii_shape.m

```
%% to take white noise and re-shape to ideal SII, ie flat to 500 Hz, and
sloping -9db/oct
%% beyond that. Durn in secs, and fs is sampling rate.

%% to perform pre-emphasis, rising 10 dB from 500 to 4k Hz at 3.3 dB/oct
%% removed from NchanNmod_AGC in Feb 2003, should be backward compatible
%% FIR filter width is always 20 msec
function sii_nse = gen_sii_noise(durn, fs, diagnostic);
if diagnostic, fprintf(1,'\ngen_sii_shape:\tgenerate speech-weighted
calibration noise'); end
%%%%% ideal pre-emphasis (Speech Intelligibility Index, ANSI 1997)
hz =       [0,100,200,450, 550,707,1000,1414, 2000,
2828,4000,5656,8000,16000,32000];
emphasis = [0.,0., 0., 0, -0.5,-4.5, -9.,-13.5, -18,-22.5,-27,-31.5,-36.,-
45,-54];

%%% make sure that response goes only up to fs/2
nyquist = fs/2;
ixf_useful = find(hz < nyquist);
hz_used = hz(ixf_useful);  %% subselect
hz_used = [hz_used nyquist];
last_emph = interp1(hz, emphasis, nyquist);  %% generate synthetic response
at Nyquist
emph_used = [emphasis(ixf_useful) last_emph];

m_used = exp(log(10)*emph_used/20);
%%% whatever fs, filter has 10 msec window, but make sure that it is even
b = fir2(2*ceil(10*(fs/2000)), hz_used./nyquist, m_used); % f= 1 is equiv
to Nyquist
%%freqz(b,1,2048,fs); grid on; set(gca,'ylim',[-40 2],'xlim',[50
nyquist],'xscale','log');
%%%% white noise, 0 DC
nburst = (rand(1,durn*fs)-0.5);
sii_nse = filter(b,1,nburst);
if diagnostic, fprintf(1,'\ngen_sii_shape:\ttime_advance compensates for
FIR filter'); end
%% this introduces a delay so remove it, ie time-ADVANCE audio
dly_shift = floor(length(b)/2); %%% compensating shift to time-align all
filter outputs
if diagnostic, fprintf(1,'\ngen_sii_shape:\tcompensating shift to time-
align all filter outputs'); end
valid_len = length(sii_nse)-dly_shift;%% _advance_ filter outputs
if diagnostic, fprintf(1,'\ngen_sii_shape:\t _advance_ filter outputs');
end
sii_nse(1:valid_len) =  sii_nse(1+dly_shift:end); %%% time advance
if diagnostic, fprintf(1,'\ngen_sii_shape:\t time advance'); end
sii_nse(1+valid_len:end) = 0.; %%% kill the rest
if diagnostic, fprintf(1,'\ngen_sii_shape:\tkill the rest'); end
```

14-6 generate_key_percent.m

```
%%% to track a certain percentage of frames in order to get measure of rms

%%% derivative of generate_key

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%

%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%%%%%%%%%%%%%%%%%%%%%%%%

%%%%%%%%%% wavlevel.m to measure the rms of a 16 kHz monaural wav file and
return mean and stddev
```

```
%%%%%%%%%% adaptively sets threshold after looking at histogram of whole
recording

%%%%%%%%%% MAS 25-9-00

function [key, used_thr_dB] = generate_key_percent(sig,thr_dB,winlen); %%
NB threshold is in dB



if winlen ~= floor(winlen) %%% whoops on fractional indexing: 7-March 2002

    winlen = floor(winlen);

    fprintf(1,'\nGenerate_key_percent:\tWindow length must be integer: now
%d',winlen);

end



siglen = length(sig);

%% new Dec 2003. Possibly track percentage of frames rather than fixed
threshold

if length(thr_dB) > 1,  track_percent = 1; percent2track = thr_dB(2);
fprintf(1,'\nGenerate_key_percent:\ttracking %.1f percentage of frames ',
percent2track);

else                    track_percent = 0;
fprintf(1,'\nGenerate_key_percent:\ttracking fixed threshold'); end

expected = thr_dB(1); %%%% expected threshold

non_zero = 10.^((expected-30)/10);  %% put floor into histogram
distribution



nframes = 0;

totframes = floor(siglen/winlen);

every_dB = zeros(1,totframes);



for ix = 1:winlen:(winlen*totframes)

    nframes = nframes + 1;

    this_sum = sum(sig(ix:(ix+winlen-1)).^2); %%%%% sum of squares

    every_dB(nframes) = 10*log10(non_zero + this_sum/winlen);

end %%%% of ix loop

every_dB = every_dB(1:nframes);  %%%% from now on save only those analysed

[nbins, lvls] = hist(every_dB(1:nframes),140);  %%%%% Bec 2003, was 100 to
give about a 0.5 dB quantising of levels
```

```
if track_percent %%% new 1-Dec-2003

    inactive_bins = (100-percent2track)*nframes/100; %% min number of bins
to use

    nlvls = length(lvls);

    inactive_ix = 0; ixcnt = 0;

    for ix = 1:nlvls

        inactive_ix = inactive_ix + nbins(ix);

        if inactive_ix > inactive_bins, break; else ixcnt = ixcnt+1; end;

    end

    if ix == nlvls, error('\nGenerate_key_percent:\tErrrrr, no levels to
count');

    elseif ix == 1, fprintf(1,'\nGenerate_key_percent:\tCounted every
bin.........'); end

    expected = lvls(max(1,ixcnt)); %% set new threshold conservatively to
include more bins than desired (rather than fewer)

end

%%plot(lvls,nbins,'b',[thr_dB(1) thr_dB(1)], [0 max(nbins)],'r', [expected
expected], [1 max(nbins)],'g'); grid on; set(gca,'ylim',[3
300],'yscale','log','xlim',[expected-25 expected+45]);drawnow;

used_thr_dB = expected; %% for feeding back to calling program



%%%%% histogram should produce a two-peaked curve: thresh should  be set in
valley

%%%%% between the two peaks, and set threshold a bit above that, as it
heads for main peak

frame_index = find(every_dB >= expected);

valid_frames = length(frame_index);

key = zeros(1,valid_frames*winlen);

%%%%% convert frame numbers into indices for sig

for ix = 1:valid_frames

    meas_span = 1+((frame_index(ix)-1)*winlen):(frame_index(ix))*winlen;

    key_span  = 1+((ix-1)*winlen):ix*winlen;

    if min(key_span) < 1

        fprintf(1,'\n\t\tkey_span: Trapped erroneous indexing %d:%d:
PAUSED',1+((ix-1)*winlen),ix*winlen);

        pause

    end
```

```
    key(key_span) = meas_span;

end
```

14-7 makesmearmat3.m

```
function fsmear=makesmearmat3(rl,ru)
% fsmear = makesmear(rl,ru);  make the smearing filter matrix
%   rl, ru: filter broadening factor on the lower and upper side,
respectively
%   filter matrices are size FFTSIZE/2 X FFTSIZE/2
FFTSIZE=512;   %FFTSIZE is assumed to contain a factor of 4
nyquist=FFTSIZE/2;
fnor=audfilt(1,1,nyquist);
fwid=audfilt(rl,ru,nyquist);
fnext=[fnor zeros(nyquist,nyquist/2)];      % extend the normal matrix so
that it the left-divide works better
for i=nyquist/2+1:nyquist
    fnext(i,nyquist+1:min(2*i-1,3*nyquist/2))=fnor(i,2*i-nyquist-1:-
1:max(1,2*i-3*nyquist/2));  % this extends the upper side of the higher-cf
filters
end

fsmear=fnext\fwid;      % this is equalvalent to multiplying (convolving)
the inverse of the normal filters with the wide filters.
fsmear=fsmear(1:nyquist,:);  % pruning to remove the extra bit
function filter=audfilt(rl, ru, size)
% audfilt - calculate an auditory filter array
%  rl = broadening factor on the lower side
%  ru = broadening factor on the upper side
% array size has been set to 128
% sampling frequency has been set to 16000
sampfreq=16000;
filter=zeros(size,size);
filter(1,1)=1.;
% uncomment or comment the following if/if-not dividing by the erb to
remove spectral tilt from the excitation pattern
filter(1,1)=filter(1,1)/((rl+ru)/2);
for i=2:1:size
     fhz=(i-1)*sampfreq/(2*size);
     erbhz=24.7*((fhz*.00437)+1.0);
     pl=4.0*fhz/(erbhz*rl);
     pu=4.0*fhz/(erbhz*ru);
     j=[1:i-1];
% for lower side of the filter
          g(j)=abs((i-j)/(i-1));
          filter(i,j)=(1+(pl*g(j))).*exp(-1.0*pl*g(j));
     j=[i:size];
% for upper side of the filter and center
          g(j)=abs((i-j)/(i-1));
          filter(i,j)=(1+(pu*g(j))).*exp(-1.0*pu*g(j));
          % uncomment the following if dividing by the erb to remove the
spectral tilt
          filter(i,:)=filter(i,:)/(erbhz*(rl+ru)/(2*24.7));
end
```

14-8 measure_rms.m

```
%% Sept 2005, standardise on measuring RMS, not tracking percentage of
frames
%% measures total power of all 10 msec frames that are above a user-
specified threshold.
%%  dB_rel_rms is threshold relative to first-stage rms, if it is made ot a
2x1 array, second value over rules.
%% This is the percentage of frames that are required to be tracked for
measuring RMS (useful when DR compression changes histogram shape).
```

```
%% returns calculated rms (linear), "key" array of indices of samples used
in RMS calculation, and actual threshold
%% used relative to RMS, for calculation
function [rms, key, rel_dB_thresh] = measure_rms(signal, fs, dB_rel_rms);
%% measure RMS of audio signal in a standard way
WIN_SECS = .01; %% 10 msecs
%% first RMS is of all signal.
first_stage_rms = sqrt(sum(signal.^2)/length(signal));
%% use this RMS to generate key threshold to get more accurate RMS
key_thr_dB = max(20*log10(first_stage_rms) +  dB_rel_rms(1), -80);%%
Nov2003, put max in here for when signal is close to 0

if length(dB_rel_rms) > 1 %% possibility of tracking percentage of frames,
rather than threshold (%-age comes in as second parameter in dB_rel_rms)
    [key used_thr_dB]= generate_key_percent(signal, [key_thr_dB
dB_rel_rms(2)], round(WIN_SECS*fs)); %%%% move key_thr_dB to account for
noise less peakier than signal
else
    [key used_thr_dB]= generate_key_percent(signal, key_thr_dB,
round(WIN_SECS*fs)); %%%% move key_thr_dB to account for noise less peakier
than signal
end
active = 100*length(key)/length(signal); %%% statistic to be reported
later, BUT save for later (for independent==1 loop where it sets a target
for rms measure)
rms = sqrt(sum(signal(key).^2)/length(key));
rel_dB_thresh = used_thr_dB-20*log10(rms); %% for returning to main program
```

14-9 smear3.m

```
%% post MAS, since called from MATLAB, keep MATLAB precision between input
and output, so do not use 16-bit file as transfer medium.
function outbuffer = smear3_postMAS(inbuffer, fsmear)

%smear2(input,output,fsmear)
%      Calculate smeared waveform to simulate broadened auditory filters.
%      'input' and 'output' should contain filenames.  Data are assumed to
be in WAV format sampled at 16 kHz.
%      Filter must have been stored in array 'fsmear'. It must be a square
matrix of size FFTSIZE/2 (128?).

FFTSIZE=512;
INFRAMESIZE=256;
SHIFT=64;

inlength=length(inbuffer);
inpointer=0;
outbuffer=zeros(ceil(inlength/SHIFT)*SHIFT,1);
outpointer=0;

overlaps=INFRAMESIZE/SHIFT;
outwave=zeros(INFRAMESIZE,overlaps); % ring buffer
buffer=[1:overlaps];       % pointers to ring buffer. buffer(1) is current.
nyquist=FFTSIZE/2;

window=0.5-0.5*cos(2*pi*([1:INFRAMESIZE]-.5)/INFRAMESIZE)'; % modified
hanning window
window=window/sqrt(1.5);
%outwin(offset+1:offset+INFRAMESIZE)=ones(INFRAMESIZE,1);
%outwin(1:offset)=0.5-0.5*cos(pi*([1:offset]-.5)/offset)';
%outwin(FFTSIZE:-1:FFTSIZE-offset+1)=outwin(1:offset);
%outwin=outwin/sqrt(1.5);

samplecount=min(INFRAMESIZE,inlength-inpointer);
inwave=inbuffer(inpointer+[1:samplecount]);
while samplecount>0
```

```
  winwave=zeros(FFTSIZE,1);

  winwave(1:INFRAMESIZE)=window.*inwave;

  spectrum=fft(winwave,FFTSIZE);

  power=spectrum(1:nyquist).*conj(spectrum(1:nyquist));

  mag=sqrt(power);

  phasor=spectrum(1:nyquist)./(mag+(mag==0)); % to prevent divide by zero

  smeared=fsmear*power;

  spectrum(1:nyquist)=sqrt(smeared).*phasor;

  spectrum(nyquist+1)=0;

  spectrum((nyquist+2):FFTSIZE)=conj(spectrum(nyquist:-1:2));

  winwave=real(ifft(spectrum,FFTSIZE));

  outwave(:,buffer(1))=winwave(1:INFRAMESIZE).*window;

  outframe=zeros(SHIFT,1);

  j=0;

  for i=1:overlaps,

    outframe=outframe+outwave(j+[1:SHIFT],buffer(i));

    j=j+SHIFT;

  end

  outbuffer(outpointer+[1:SHIFT])=outframe;

  outpointer=outpointer+SHIFT;


  buffer([2:overlaps,1])=buffer;   %rotate ring buffer

  inwave=[inwave(SHIFT+1:INFRAMESIZE);zeros(SHIFT,1)];

  inpointer=inpointer+SHIFT;

  samplecount=min(SHIFT,inlength-inpointer);

  inframe=inbuffer(inpointer+[1:samplecount]);

  inwave(INFRAMESIZE-SHIFT+1:INFRAMESIZE-SHIFT+samplecount)=inframe;

end
for k=overlaps-1:-1:1,     % flush the overlap/add buffer
  outframe=zeros(SHIFT,1);
  j=(overlaps-1-k)*SHIFT;
  for i=1:k,
    outframe=outframe+outwave(j+[1:SHIFT],buffer(i));
    j=j+SHIFT;
  end
  outbuffer(outpointer+[1:SHIFT])=outframe;
  outpointer=outpointer+SHIFT;
end
```

Appendix 15: Figures and tables for Chapter Three

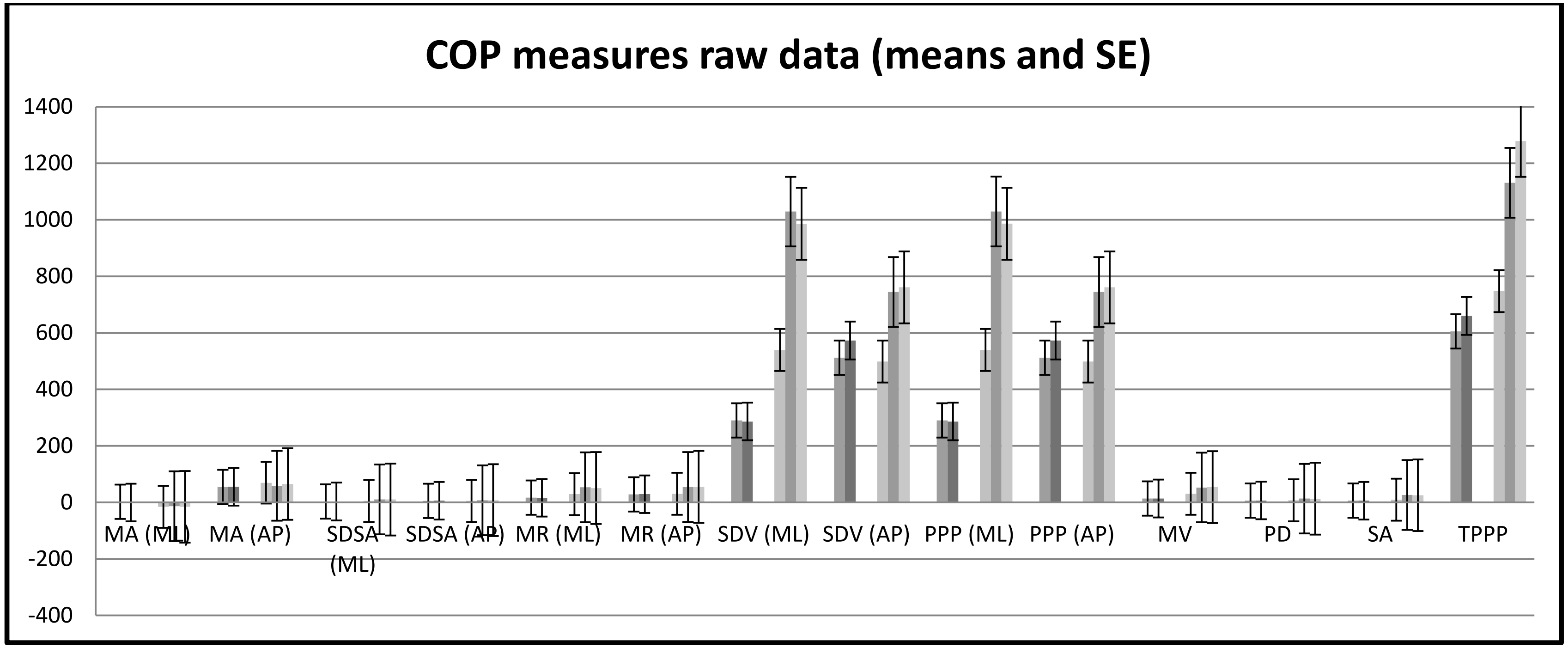

Figure Appenix 15 -1 COP measures raw data means in different test conditions. Unites of COP measures are as follows: mm (amplitude/SD amplitude/Range/planar deviation); mm/s (SD velocity/mean velocity); mm²/s (Area); phase plane and total phase plane is in an arbitrary unit.