The contribution of hearing and hearing loss to balance control

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TITLE PAGE

Title: The Contribution of Hearing and Hearing Loss to Balance Control.

Short Title: Hearing and Balance Control.

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Key Words: Vestibular, hearing loss, posturography, hearing aid, spatial orientation, hearing, stabilometry, postural sway, sensory reweighting.

Acronyms and Abbreviations: SNHL = Sensorineural hearing loss, CHL = Conductive hearing loss, B = Bilateral, U = unilateral, Asym = Asymmetrical, HF = high frequency, LF = low frequency, UVD = unilateral vestibular dysfunction, SCD = superior canal dehiscence, BPPV = benign paroxysmal positional vertigo, N/A = Not Applicable, BTE = Behind The Ear, RITE = Receiver In The Ear, ITE = In The Ear, ITC = In The Canal, CIC = Completely In The Canal, dBHL = decibels hearing level, sec = seconds, 4FA = four frequency average, cVEMP = cervical Vestibular Evoked Myogenic Potential, oVEMP = ocular Vestibular Evoked Myogenic Potential, WBB = Wii Balance Board, COP = Centre of Pressure, ANOVA = Analysis of Variance.

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ABSTRACT

This study investigated the hypothesis that a hearing 'map' of our surroundings is used to maintain balance control. We investigated the effects of sound on postural sway using centre of pressure analysis in 50 normally hearing subjects, 28 subjects with hearing loss and 19 subjects with vestibular dysfunction. The acoustic environments utilised sound cues that were either present or absent. It was found that auditory cues are utilized by normally hearing subjects to improve postural sway. The ability to utilize sound for postural control is diminished when there is a hearing loss, but this appears to be overcome by the use of a hearing aid. Patients with additional vestibular deficits exploit auditory cues to a greater degree, suggesting a sensory weighting to enhance the use of auditory cues may be applied when there is diminished sensory redundancy.

INTRODUCTION

Balance is a complex process involving sensory and motor integration. The classic sensory systems that are known to contribute to balance control are the visual, somatosensory and vestibular systems. While the proportional input contributions will vary depending on the environment as well as any deficits in the systems, growing evidence suggests that there is greater contribution from the somatosensory inputs, and lesser on the vestibular and visual inputs [Mergner et al., 2002].

The auditory system is also thought to contribute spatial sound cues which supplement balance control. However, the nature of the auditory systems involvement is unclear with conflicting results between studies which suggest that a lack of sound [Kanegaonkar et al., 2012] or the presence of sound may increase postural sway in normally hearing subjects [Park et al., 2011; Raper and Soames, 1991; Tanaka et al., 2001].

If hearing/sound does contribute to balance control then it stands to reason that a loss of hearing ability will negatively affect balance. Hearing loss has been associated with self-reported falls [Lin and Ferrucci, 2012], and reduced mobility [Viljanen et al., 2009a], which has been linked, in part, to poorer postural control [Viljanen et al., 2009b]. Studies which have directly compared postural sway in hearing impaired subjects to those with normal hearing have shown both poorer [Huang et al., 2011; Kluenter et al., 2009] and better [Walicka-Cuprys et al., 2014] postural stability in the hearing impaired. When controlling for underlying vestibular deficits, which are often concomitant with hearing loss, no significant differences were found between normally hearing and hearing impaired subjects [Suarez et al., 2007].

The relationship between hearing/sound and postural sway has also been investigated by examining the effect of wearing a hearing device. The assumption is that a person with hearing loss may benefit from a hearing device which improves the audibility of auditory spatial cues used in postural control. However, this premise has primarily been investigated in the cochlear implant population, where the results have been contradictory [Buchman et al., 2004; Cushing et al., 2012; Huang et al., 2011; Kluenter et al., 2009; Suarez et al., 2007]. Investigating this cohort also has its limitations in that subjects are largely fitted unilaterally, eliminating any spatial benefits of binaural hearing which could contribute to postural stability. Moreover, the underlying etiology of the cochlea loss may be associated with additional vestibular deficits and there is additional ambiguity regarding inadvertent electrical stimulation of the vestibular nerve via the cochlear implant.

Given the lack of clarity around the contribution of sound to balance control, this study first aimed to clarify the effect of sound environment on postural sway in normally hearing subjects. The secondary aim was to ascertain the consequences of hearing loss on sway and investigate the role that hearing aids may play in balance control. Our final aim was to establish whether the weighting of auditory cues changes when there is an additional vestibular deficit.

METHOD

PARTICIPANTS

All subjects gave their informed consent to participate in this ethically approved study (Royal Victorian Eye and Ear Hospital Human Research Ethics Committee).

Normal hearing participants were recruited by means of an institution wide advertisement. They were required to have no hearing loss (>20dBHL in both ears across the frequency range) or history of dizziness/balance issues. Subjects with hearing and/or vestibular loss were identified and recruited from the University of Melbourne Audiology Clinic via a letter of invitation. Participants who expressed interest in the study were screened for eligibility before informed consent was obtained. The patients' audiometric function tests were used to categorize subjects into normal, hearing impaired only or vestibular impaired groups.

A hearing impaired participant was included if the average left and right ear four frequency average (4FA) hearing level was more than 20dBHL. Participants in the hearing loss group were excluded if they reported any vertiginous episodes or imbalance problems. A proportion of these participants were also required to have hearing aids in order for the effects of these devices to be measured.
Vestibular impaired subjects were required to have a documented vestibular dysfunction of any degree or configuration. This was verified using the participant's most recent vestibular function test results. Clinical vestibular testing included assessment of canal and otolith function via video head impulse, caloric, air-conducted cVEMP and bone-conducted oVEMP tests. Threshold levels of at least a gain of < 0.8 with large amplitude catch up saccades, > 25% caloric unilateral weakness or > 30% VEMP asymmetries respectively, classified a subject as having a significant vestibular dysfunction.

All subjects were free from any orthopaedic or neurological conditions that may affect balance and did not require any balance or walking aids to maintain quiet stance.

The demographics of the each of the groups examined are found in Table 1.

Insert Table 1 here

POSTURAL SWAY MEASURES

Static postural sway measures were obtained via a Nintendo Wii Balance Board (WBB). The WBB data were recorded on a laptop computer using Bluetooth and custom software created in LabVIEW 2009 (National Instruments, Austin, TX, U.S.A.). Calibration was achieved by placing a variety of known loads at different positions on the WBB, as discussed in [Clark et al., 2010], and data were collected and processed in accordance with [Holmes et al., 2013] The WBB has been validated for balance assessment in a number of prior studies [Huurnink et al., 2013; Scaglioni-Solano and Aragón-Vargas, 2014]. The outcome measure used in this study was total Centre of Pressure (COP) path length, which quantifies the total amount of sway during a trial, and which is known to be a reliable and valid measure of standing balance [Salavati et al., 2009].

Each participant was required to stand with feet parallel and 10 cm apart with their hands by their side. Participants were instructed to stand as still as possible while sway was measured over a single 60 second period. Participants were allowed to rest between trials.

Participants completed a total of 16 trials in four standing conditions across four different acoustic environments. The four standing conditions selectively eliminated visual and/or proprioceptive cues by instructing the subject to close their eyes and/or stand on a piece of foam respectively. The foam piece was made of high density ethylene-vinyl acetate measuring 50cm x 30cm x 5cm allowing it to completely cover the platform of the WBB. The four standing conditions were firm surface eyes open, firm surface eyes closed, foam surface eyes open and foam surface eyes closed. The four sound environments consisted of an ambient room of similar dimensions to the sound treated room, sound treated room with the subject wearing EAR earplugs, sound treated room with a continuous white noise presented from a front speaker 106cm away, and a sound treated room with a moving noise. The moving noise condition was achieved by the sequential presentation of white noise (0.5ms ramp time and a 1 sec plateau) through 8 speakers moving from left to right and back. The 8 speakers were arranged in a 180 degree semicircular array, spaced equidistantly. The WBB was placed in the centre of the array with a radius of 106cm from each speaker. This simulated a moving sound source which took approximately 8 seconds to pan from left to right. As the degree and configurations of the subjects hearing losses were varied, we chose a white noise stimulus which was presented at a comfortable listening level. The levels used in this study ranged between 60dBA-70dBA and were kept consistent within each subject. In order to eliminate the potential effects of fatigue/practice, the order of testing was randomized.

To assess the effect of hearing loss with and without hearing aids, participants with a hearing loss were assessed in both aided (wearing their hearing device/s turned on) and unaided conditions with the exception of the 'aided' EAR plug condition. If subjects had different programs in their hearing aid/s, they were instructed to use their primary or most used hearing aid program. All trials were randomized across both aided/unaided and sound and standing orders to avoid the effects of fatigue on repeated trials.

Statistical Analysis

Missing data (<0.5%) were imputed using averages for the corresponding subject cohort in that particular standing condition and sound environment. As the COP path length data was positively skewed, the data was LOG (base 10) transformed. A general linear model 2 factor repeated measures ANOVA was used to determine the effects of the four sound environments and the four standing conditions in normally hearing subjects.

In order to investigate whether hearing loss affects balance control we intended to compare the COP path length in normally hearing subjects with the hearing loss subjects. This analysis was, however, confounded by a significant age discrepancy between the groups. As such, we took the approach of analyzing the subjects within the hearing loss group only, to determine if there was any effect of the degree of hearing loss (4FA), age and sound environment on sway. A general linear model repeated measures ANOVA was used to determine the effects of the sound environment, standing condition, degree of hearing loss, age and their interactions on LOG path length. We examined the effect of wearing a hearing aid on sway using a repeated measures general linear model ANOVA with sound environment, standing condition, and aided/unaided condition as factors.

To investigate whether the weighting of auditory cues changes when there is an additional vestibular deficit, we compared the vestibular impaired group to the hearing loss group since they were of a more comparable age and range of hearing loss. A general linear model repeated measures ANOVA was applied to the vestibular and hearing loss cohort to ascertain whether vestibular status, sound environment and standing condition had an effect on path length. Age and degree of hearing loss was also entered into the model as covariates in order to control for differences in age and degree of hearing loss between the groups.

RESULTS

In the fifty normally hearing subjects there was a significant effect of sound environment ($p=0.018$) and standing position ($p<0.001$), but no significant interactions ($p>0.05$). The effects of standing position showed higher path lengths as visual (eyes closed) and proprioceptive (foam) information was removed, as shown in Figure 1. Post hoc analysis of individual differences showed that each of the four standing conditions were significantly different from each other (Tukey simultaneous comparison $p<0.001$). This pattern of standing condition result is well established in the literature [Baloh et al., 1998]. As this result was significant and consistent across all further analyses in this study, it will not be further commented on.

Insert Figure 1 here

The contribution of sound environment to sway measures is substantially smaller relative to the effects of vision and proprioception, as shown in Figure 1. The mean path length of all four standing conditions was highest in the ambient environment and consecutively lower in the EAR plug, stationary sound and moving sound environments respectively. Post hoc analysis, however, only revealed a significant difference between the ambient and moving sound conditions (Tukey simultaneous comparison, $p=0.015$). Alternatively, if we look at the sound environment trends across the different standing conditions (Figure 1) there is some individual variation. For example, the order of highest to lowest mean sway for the 'standing eyes closed' condition was EAR Plugs, ambient, stationary and moving. Alternatively, the order for foam eyes closed was ambient, EAR Plugs, stationary, moving. This indicates that while there were subtle differences between orders of mean sway between the individual sound environments, there was a trend for the two conditions devoid of sound to consistently have the highest sway measures. Therefore the overall meaning of the post hoc analysis showing a difference only between ambient and moving noise is difficult to decipher. The more meaningful trend is perhaps the overall presence or absence of sound, especially as this trend was also apparent in the analysis for the other subject cohorts. As such, for all further analyses, the ambient room and EAR Plug data are collapsed (No Sound), and the stationary and moving sound data were collapsed (With Sound).
Table 2 shows the outcomes for each of the analyses across the study cohorts using the collapsed sound conditions.

Insert Table 2 here

Figure 2 shows the significant effect of sound environment on LOG path length in normally hearing subjects where sway was higher in the absence of sound. Alternatively, in subjects with hearing loss (Figure 3), there was no significant effect of sound environment which suggests that subjects with hearing loss don't/can't use sound to improve balance. Age was also associated with higher sway measures, consistent with previous literature [Baloh et al., 1994].

Insert Figure 2 here

Insert Figure 3 here

The hearing impaired cohort was not able to utilize auditory sound environment cues, presumably due to their hearing loss. Interestingly, wearing a hearing aid while not increasing stability overall, did interact with the sound environment. As Figure 4 shows, wearing a hearing aid affects the use of sound environment cues differently. Unaided, sound appears to slightly increase sway, but when the subject is wearing their hearing aid in the presence of sound, they have less sway. It appears that on average, the best scenario for someone with hearing loss is to be wearing their hearing aids in the presence of sound.

Insert Figure 4 here

The final analysis investigated whether having a pre-existing vestibular deficit makes sound a more important cue for balance control, whilst controlling for age and degree of hearing loss. As expected, vestibular subjects had higher path lengths compared to the subjects who had normal balance (Figure 5). Figure 5 also shows the interaction effects where there is increased sway in the absence of sound in subjects with a balance problem. Thus vestibular subjects appear to utilize their remaining hearing when sound cues are present while the hearing impaired/normal balance subjects do not.

Insert Figure 5 here

DISCUSSION

Sway in normally hearing subjects was increased in conditions where there were fewer sound cues available. This suggests that a spatial hearing map contributes to balance control, although the effect size was considerably smaller than the contribution from the somatosensory and visual systems. The premise that sound cues can only contribute to balance control if they are audible was also demonstrated in the hearing impaired subjects; specifically, the presence of sound did not affect sway in the unaided condition, but was associated with reduced sway when users wore their hearing aids. The amplification provided by the hearing aids appears to increase access to spatial cues in order for them to be utilized for balance control.

It is encouraging to see that hearing aid processing of sound did not adversely affect sway. Nonetheless, sound processing through the hearing aid could have impacted the potential effectiveness of providing amplified spatial cues. Hearing aids use a variety of sound processing algorithms with the primary emphasis on enhancing speech perception and reducing background noise. As such, auditory spatial cues most important for balance control could have been distorted by the processing. This notion may explain the finding that mean sway was slightly worse in the aided - no sound condition than it was for the unaided - no sound condition. It's possible that there are effects of hearing aid processing for soft sounds that interfere with the ability of the listener to use very soft 'ambient' sounds for balance. It is not uncommon for hearing aids to reduce gain for low level or unmodulated inputs as a method of reducing the annoyance of low level background noise. As such, unaided users may have access to at least some ambient sound cues until they wear their hearing aids. This would particularly affect users with minimal hearing loss or hearing loss only affecting certain frequency ranges. This was the case for many hearing impaired subjects in this study. Given the wide array of hearing configurations, hearing aid styles and features, it is a limitation of this study that we were not able to control or investigate these aspects. This could be an avenue for further research.

Until recently, there was limited data on the effects of hearing interventions on balance function outside of the cochlear implant population. Rumalla et al., (2015) showed the benefits of bilateral hearing aids in extending the duration of stable quiet stance in older adults. These results are congruent with the improved COP sway measures observed in this study. We have shown that hearing aids have additional benefits outside of the conventional audibility and speech perception enhancements even in subjects with mild to moderate hearing losses utilizing a broad range of hearing aids.

The last part of this study investigated whether subjects with pre-existing vestibular deficits relied more on sound cues given that they have reduced sensory redundancy. The results suggest they do. The vestibular impaired group reduced their sway in the presence of sound, while those without vestibular impairment did not. This is despite both groups having some degree of hearing loss and controlling for the degree of hearing loss in the statistical model. Thus, vestibular patients are utilizing their remaining hearing, more so than the hearing loss subjects without vestibular deficits and may reflect an increase in auditory sensory weighting involved in the vestibular compensation process. Auditory biofeedback systems could play a role in facilitating this reweighting and have already been explored as an adjunct to vestibular rehabilitation programs [Dozza et al., 2007; Ernst et al., 2007; Hegeman et al., 2005]. However, application of auditory biofeedback systems is low and may be related to the relatively small effect that auditory cues play amongst the other sensory systems. None of the subjects in this study undertook a formal auditory biofeedback program which suggests that sensory reweighting may occur partly through natural exposure to everyday sounds.

Sensory reweighting in vestibular patients also highlights another important aspect; the effect of experience and adaptation. Kanegaonkar et. al., (2012) proposed that each individual has a pre-existing hearing spatial map which helps with postural control. Therefore the effects of familiarity of auditory cues may contribute to postural sway. However, when the sensory inputs conflict with this pre-existing map, a disturbance to balance is perceived/measured. A conflict with the template could arise from a change in hearing cues via the removal of sound cues through a loss of hearing or by immersing oneself in an unfamiliar sound environment, or an environment devoid of sound. This concept may help to explain the anecdotal reports from patients that they feel less oriented when in a sound proof room, the perception of mild 'disorientation' that initial hearing aid users may remark on after their first hearing aid fitting or a sudden change in hearing levels. As this phenomenon appears to lessen with more experience, it suggests that a new internal spatial template can be (re)mapped over time. Indeed, exposure to sensory stimuli has been shown to be crucial in order for vestibular compensation and sensory (re)integration to occur [Fetter and Zee, 1988; Lacour et al., 1976; Newlands and Perachio, 1991]. The effect is also context specific [Baker et al., 1987; Godaux et al., 1983; Lisberger et al., 1983]. Accordingly, exposure and optimized audibility of sounds, via a hearing aid in a range of environments, should be encouraged to make the most of auditory cues in balance control.

Conclusion

Sound cues facilitate postural control in normally hearing and aided hearing impaired subjects. The role of sound/hearing plays a relatively minor role in the context of the contribution from the other sensory systems, but appears to be heightened in patients with vestibular impairment. There are significant interrelationships between hearing loss, aging and falls risk [Koh et al., 2015; Lin and Ferrucci, 2012; Lopez et al., 2011]. The benefit of hearing aids in stabilizing the hearing impaired is appealing when considering the substantial disability, mortality and socioeconomic burden of falls in this population [Sartini et al., 2010; Sterling et al., 2001].

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Table 1. Subject demographics for each of the study groups.

Group	N	Male:Female	Mean Age in years (range)	Mean Height in cm (SD)	Mean 4FA Right (SD)	Mean 4FA Left (SD)	Mean Time with hearing loss (SD)	Characteristics of hearing loss	Characteristics of vestibular loss
Normal Hearing	50	10:40	28.84 (21-56)	164.6 (8.1)	3.8 (3.5)	3.3 (3.5)	N/A	Normal	Normal
Hearing Loss	28	13:15	65.3 (26-80)	168.7 (9.2)	42.8 (16.7)	43.4 (14.6)	15.9 (13.4)		
Unaided	9	4:5	66.7 (45-80)	164.4 (7.4)	29.9 (6.3)*	34.2 (13.2)*	8.8 (17.0)*	6 x BSNHL, 2 x Asym SNHL, 1 x Asym SNHL and mixed on other ear	Normal
Aided	19	9:10	64.7 (26-79)	169.9 (9.7)	49.0 (16.7)*	47.8 (13.4)*	18.1 (12.2)*	12 x BSNHL, 5 x Asym SNHL HF, 1 x Flat SNHL and mixed on other ear, 1 x SNHL HF and mixed in other ear. 17 were fitted with hearing aids bilaterally (4 BTE, 6 RITE, 1 ITE, 2 ITC, 3 CIC) and 2 were fitted with 1 BTE.	Normal
Vestibular loss	19	10:9	63.6 (34-83)	169.6 (10.9)	27.9 (13.4)*	28.0 (21.7)*	7.7 (8.4)*	8 x BSNHL, 3 x USNHL, 3 x Asym SNHL, 3 x Asym noise notch, 2 x CHL LF with BSNHL HF	17 x UVD, 2 x SCD and BPPV

4FA is the four frequency (0.5,1,2 & 4kHz) average hearing level in dBHL, SNHL = Sensorineural hearing loss, CHL = Conductive hearing loss, B = Bilateral, U = unilateral, Asym = Asymmetrical, HF = high frequency, LF = low frequency, UVD = unilateral vestibular dysfunction, SCD = superior canal dehiscence, BPPV = benign paroxysmal positional vertigo, N/A = Not Applicable, BTE = Behind The Ear, RITE = Receiver In The Ear, ITE = In The Ear, ITC = In The Canal, CIC = Completely In The Canal. * indicates significant difference ($p<0.05$) in 4FA between the aided hearing loss group and both the unaided and vestibular impaired groups.

Table 2 shows the results of the repeated measures general linear model statistical analyses.

Primary Research Question	Cohort	Factors	Result
Does sound environment affect the maintenance of quiet stance in normally hearing subjects?	Normal hearing (N=50)	Standing condition (SEO, SEC, FEO, FEC) Sound environments (No/With sound) Interaction	**p<0.001** **p=0.025** p>0.05
Does the presence of a hearing loss affect how sound is used to maintain balance?	Hearing loss (N=28)	Standing condition (SEO, SEC, FEO, FEC) Sound environments (No/With sound) Interaction Covariates Age 4FA	**p<0.001** p>0.05 p>0.05 **p<0.001** p>0.05
In subjects with hearing loss, does wearing a hearing aid change how sound is used to maintain balance?	Aided hearing loss (N=19)	Standing condition (SEO, SEC, FEO, FEC) Sound environments (No/With sound) Aided condition (Aided/Unaided) Interaction Aided condition*Sound environment Standing condition*Sound environment	**p<0.001** p>0.05 p>0.05 **p=0.048** p>0.05
Does having a pre-existing balance problem make hearing/sound a more important cue for balance control?	Hearing loss only (N=28) Vestibular loss (N=19)	Standing condition (SEO, SEC, FEO, FEC) Sound environments (No/With sound) Vestibular status (Normal/Abnormal) Interaction Vestibular status*Sound environment Standing condition*Sound environment Covariates Age	**p<0.001** p>0.05 **p=0.019** **p=0.048** p>0.05 **p<0.001**

		4FA	p>0.05

SEO = Standing eyes open, SEC = Standing eyes closed, FEO = Foam eyes open, FEC = Foam eyes closed.

Figure Legend

Figure 1. The mean LOG Path Length for normally hearing subjects in 4 different standing conditions with sound cues either present of absent. Error bars indicate -/+ 2 standard residual errors.

Figure 2. The effect of sound on the mean LOG Path Length. Data points represent the mean LOG path length for all four standing conditions in that sound environment. Error bars are -/+ 2 mean residuals. *Path length was significantly higher in the absence of sound ($p=0.025$).

Figure 3. The effect of sound environment on the mean LOG Path Length in 28 hearing loss subjects. Data points represent the mean LOG path length for all four standing conditions. Error bars are -/+ 2 mean residual/errors. There was no significant effect of sound environment ($p>0.05$).

Figure 4. The effect of sound environment in aided and unaided conditions on mean LOG path length in 19 hearing aid users. Data points represent the mean LOG path length for all four standing conditions in that sound environment and aided condition. Error bars are displayed as -/+ 2 standard residual errors. There was no significant effect of sound environment or aided condition ($p>0.05$) but there was a significant interaction ($p=0.048$).

Figure 5. The effect on sound environment on mean LOG path length in subjects with normal (N=28) and impaired vestibular function (N=19). Data points represent the mean LOG path length for all four standing conditions. Error bars are displayed as -/+ 2 standard residual errors. There was no significant main effect of sound environment but there was a significant increase in sway in subjects with vestibular impairment ($p=0.019$). There was also an interaction effect between sound environment and vestibular deficit ($p=0.048$).