The Contribution of Cochlear Implants to Postural Stability

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Objectives: To determine whether spatial auditory cues provided by cochlear implants can improve postural balance in adults with severe deafness.

Methods: In the presence of spatial white noise, 13 adult cochlear implant patients wore head and lumbar-mounted inertial sensors while standing in the dark for 30 seconds in two auditory conditions: hearing assistive devices on and off.

Results: Stability was improved with implants on (aided condition) compared to off (unaided condition) with respect to differences in mean head velocity (Cohen’s d = 0.912, P = 0.006) as well as root mean square (RMS) acceleration (Cohen’s d = 0.456, P = 0.048). This was particularly evident in measures of anteroposterior accelerations (mean difference = 0.034 m/s²; Cohen’s d = 0.612; P = 0.011).

Conclusion: The decrease in RMS head acceleration and velocity while wearing cochlear implants suggests that they could be recognized as balance implants in addition to auditory implants. The clinical importance of this finding in various patient populations remains to be determined.

Key Words: Fall, cochlear implant, audition, inertial sensor, sensory weighting, posture, stability, balance.

Level of Evidence: 4.

INTRODUCTION

Postural control is a complex motor function derived from the integration of several neural components, including sensory and movement strategies, orientation in space, biomechanical constraints, and cognitive processing.1 Postural instability arises when one or more of these components are compromised. Postural instability is one of the most common causes of reduced quality of life, falls, and subsequent immobilization in older adults. A current model of postural control includes visual, vestibular, and proprioceptive sensory modalities.2 Measures of postural sway during quiet standing are often used to characterize postural control3 and have been shown to be sensitive to postural instability caused by neurological disease4–6 and aging/falls.7,8 Nowadays, postural sway can be quantified with portable, lightweight, inertial sensors in a similar way to what has been reported with force plate technology.9

Recent evidence has emerged showing that external auditory sources such as speaker arrays can contribute to maintaining balance, possibly by serving as fixed environmental reference points.10–13 Notably, this effect has been seen in hearing aid users, whose postural control becomes worse when they are not wearing their hearing aids, reducing their access to environmental sounds.14,15

The effect of auditory inputs provided by cochlear implants on postural stability has not yet been examined to a similar extent. A recent case report found that a bilateral cochlear implant user was able to use sound provided by cochlear implant speech processors to improve balance during gait, as measured clinically by the Mini-Balance Evaluation System Test, a multimodal balance test including static (standing) and dynamic (moving) balance measures.16,17 A follow-up finding with 12 individuals using bilateral cochlear implants suggested that this effect may be limited to those who are particularly imbalanced.18 Although this recent finding suggests potential clinical balance benefit to bilaterally aided cochlear implant and hearing aid users, the study did not objectively measure postural stability.

These preliminary reports focusing on gait suggest that cochlear implants may indeed provide a meaningful contribution toward maintaining static postural control. The public health significance of this effect could be particularly meaningful given that older people who often benefit from implantation are also at the greatest risk of injuries from falling. Here, we explore the extent to which sound inputs provided by cochlear implants improve postural stability.

MATERIALS AND METHODS

This study was carried out with the approval of the Oregon Health & Science University (OHSU) Institutional Review Board. Individuals with upcoming audiologic evaluations at the
OHSU Department of Otolaryngology formed a sample of convenience. Inclusion criteria were as follows: adults greater than 18 years old with at least 3 months of experience with bilateral cochlear implants or bimodal (one cochlear implant and one hearing aid) instrumented hearing, having unaided thresholds worse than 45 dB hearing level (HL) in their better ear, intact cognitive function (passed the Short Blessed Test\textsuperscript{15}), understanding of English-spoken directions, and the ability to ambulate unassisted. Participants were paid for their participation and were informed that the purpose of the study was to investigate the effects of hearing on balance. All participants completed the Activities-specific Balance Confidence Scale (ABC)\textsuperscript{20} and were asked to respond before and after testing to the three-alternative forced choice question, "Do you feel that your balance is better with your hearing devices on, off, or is there no difference?"

Postural stability was measured without vision for 30 seconds by performing a Romberg test. Participants were instructed to keep their feet together (no space between feet) and cross their arms (hands on shoulders) for each trial. All participants were asked to remove their shoes. Participants were tested on the most challenging substrate they could tolerate (firm ground or foam pad; Balance-pad, Airex AG, Sins, Switzerland). Participants completed three trials of each condition (implant processors on and off) in a random order, and trials of the same condition were averaged for analysis. Participants were instrumented with lumbar (L5) and head-mounted Opal inertial sensors (APDM, Inc., Portland, OR) that recorded three-dimensional accelerations; these were further used for data analysis (Fig. 1). Briefly, the accelerometers provide an acceleration vector derived from the direction and magnitude of the gravity vector with respect to the sensor. With the subject upright, the sensor reads zero, but with increased tilt the sensor’s reading increases correspondingly in the direction of tilt. Thus, the acceleration vector generally corresponds to the position of the center of pressure. By convention, measurements in the anteroposterior (AP) direction are positive with anterior acceleration (corresponding to tilting forward) and in the mediolateral (ML) direction are positive with rightward acceleration (tilting right). Hence, the sensor output registers the length of the acceleration vector as sway (approximating tilt, or distance of the center of pressure (CoP) from the origin) in units of m/s\textsuperscript{2}; the sway path length over which the tip of the acceleration vector moved in units of m/s\textsuperscript{2} (approximating path length of CoP); and the mean velocity of the tip of the acceleration vector in units of acceleration over time, or m (approximating mean velocity of CoP). The output of these inertial sensors has been validated against standard posturography and can output measures more sensitive than CoP measures taken from a pressure plate.\textsuperscript{9} Mobility Lab (APDM) and MatLab (MathWorks, Natick, MA) software were used to analyze the postural sway data tracings.

Broadband white noise (0–4 kHz, 65 ± 5 dB sound pressure level (SPL)) was presented for the duration of the experiment. This frequency band was chosen because some hearing devices apply frequency compression above 4 kHz, which could otherwise cause a confounding effect when comparing performance in the amplified versus unamplified condition. The white noise was played by a speaker (frequency response of 0.1 to 22.0 kHz Model R1; YC Cable, Ontario, CA) placed 1 meter directly in front of the participants and adjusted to be at ear level.

The outcome measures included root mean square (RMS) of the acceleration of the head (sway dispersion) and lumbar-mounted inertial sensor, decomposed into both AP and ML directional subcomponents and mean velocity, as previously described in clinical populations.\textsuperscript{9} Additional variables included time domain measures such as head and lumbar sway path length. A two-tailed, nonparametric, Wilcoxon sign-ranked test was performed on the paired datasets in Prism (GraphPad, La Jolla, CA) in order to determine significance. Effect size was calculated in terms of Cohen’s d.\textsuperscript{21}

RESULTS

Participant demographics are reported in Table I. Ages ranged from 23 to 84 (mean: 62.8; standard deviation 19.1). Both before and after testing, all participants reported that they perceived no difference in their balance as a result of having their hearing devices on or off, with the exception of S12, who consistently noted an improvement in the device-on condition. On the ABC, 12 participants scored in the range, indicating a high level of confidence in their physical functioning. One participant, S12, scored in the range, indicating a medium level of confidence in their physical functioning.

Outcome measures for the 13 participants with cochlear implants can be found in Table II. Two trials out of 39 were discarded prior to analysis due to abnormally large accelerometer spikes correlated with observed volitional movement.

The overall mean RMS acceleration of the head sway was 0.234 m/s\textsuperscript{2} with hearing assistive devices and 0.263 m/s\textsuperscript{2} without. This difference was statistically significant ($P = 0.048$), with 95% confidence intervals of 0.20 to .27 in the aided condition and 0.22 to 0.31 in the unaided condition. Differences in individual participants are shown via slope in Figure 2.

Limiting analysis to RMS acceleration of the head in the AP direction only, the effect of augmenting sound cues was clearer ($P = 0.011$, 95% CIs 0.16–0.21 aided, and 0.18–0.26 unaided) (Table II) (Fig. 2). In the ML direction, the reduction of sway with hearing assistive devices was not significant ($P = 0.735$, confidence intervals 0.11–0.16 aided and 0.11–0.17 unaided).
The mean velocity of head motion of individual participants is also shown in Figure 2. A large effect of audition was seen with this outcome measure (Cohen’s $d = 0.912$). Similar effects of direction were observed, with the effect being largely in the AP direction. Analogous measurements for the lumbar sensor are shown in Table II. No statistically significant effects were observed with the lumbar sensor outcome measures, including RMS acceleration, mean velocity, and path length.

**DISCUSSION**

Our results show that bimodal or bilateral cochlear implant conditions significantly stabilize the head while standing, and at the same time reduce lumbar accelerations but to a smaller degree. This indicates that the availability of external auditory information may improve stability in at least some patients with bilateral cochlear implants and bimodal hearing.

The findings of two previous works examining gait in bilateral cochlear implant users support the results presented here. A small case series, including one bilateral cochlear implant user, found that spatial sound could improve gait velocity and step length as well as balance on a Romberg test. A larger report, focusing only on gait parameters, found that some cochlear implant recipients were able to use a single, point-source spatial sound to improve gait parameters (including velocity, double support phase, stride length variability, and swing time variability), although benefit was not available to everyone. The present dataset, with 85% (11 of 13) of individuals improving in the aided condition, suggests that most, but not necessarily all, implant users may benefit from the effect of audition. The two participants, S01 and S06 (Table I) (Fig. 2), who did not improve.

**TABLE I.**

<table>
<thead>
<tr>
<th>Participant Code</th>
<th>Sex</th>
<th>Age (years)</th>
<th>Type</th>
<th>Hearing Device Type (manufacturer*)</th>
<th>Device Experience (years)</th>
</tr>
</thead>
<tbody>
<tr>
<td>S01</td>
<td>M</td>
<td>71</td>
<td>Bilateral</td>
<td>Advanced Bionics (CI-both)</td>
<td>L: 2 R: 10</td>
</tr>
<tr>
<td>S02</td>
<td>M</td>
<td>81</td>
<td>Bilateral</td>
<td>Advanced Bionics (R-CI), Widex (L-HA)</td>
<td>L: 20 R: 9</td>
</tr>
<tr>
<td>S03</td>
<td>M</td>
<td>23</td>
<td>Bilateral</td>
<td>Cochlear (CI-both)</td>
<td>L: 8 R: 14</td>
</tr>
<tr>
<td>S04</td>
<td>M</td>
<td>63</td>
<td>Bilateral</td>
<td>Med-El (L-CI), Phonak Naida (R-HA)</td>
<td>L: 2 R: 20</td>
</tr>
<tr>
<td>S05</td>
<td>M</td>
<td>57</td>
<td>Bilateral</td>
<td>Advanced Bionics (R-CI), Phonak Naida (L-HA)</td>
<td>L: 0.5 R: 1</td>
</tr>
<tr>
<td>S06</td>
<td>F</td>
<td>65</td>
<td>Bilateral</td>
<td>Med-El (CI-Both)</td>
<td>L: 3 R: 2</td>
</tr>
<tr>
<td>S07</td>
<td>M</td>
<td>81</td>
<td>Bilateral</td>
<td>Cochlear (CI-Both)</td>
<td>L: 7 R: 0.5</td>
</tr>
<tr>
<td>S08</td>
<td>M</td>
<td>64</td>
<td>Bilateral</td>
<td>Cochlear (CI-Both)</td>
<td>L: 11 R: 1</td>
</tr>
<tr>
<td>S09</td>
<td>M</td>
<td>65</td>
<td>Bilateral</td>
<td>Cochlear (CI-Both)</td>
<td>L: 0.5 R: 12</td>
</tr>
<tr>
<td>S10</td>
<td>M</td>
<td>29</td>
<td>Bilateral</td>
<td>Cochlear (CI-Both)</td>
<td>L: 13 R: 6</td>
</tr>
<tr>
<td>S11</td>
<td>F</td>
<td>84</td>
<td>Bilateral</td>
<td>Cochlear (CI-Both)</td>
<td>L: 7 R: 5</td>
</tr>
<tr>
<td>S12</td>
<td>F</td>
<td>81</td>
<td>Bilateral</td>
<td>Med-El (CI-Both)</td>
<td>L: 1 R: 12</td>
</tr>
<tr>
<td>S13</td>
<td>M</td>
<td>52</td>
<td>Bilateral</td>
<td>Advanced Bionics (CI-both)</td>
<td>L: 4 R: 14</td>
</tr>
</tbody>
</table>

*Advanced Bionics, LLC (Valencia, California, U.S.A.); Cochlear (Sydney, Australia); Med-El (Innsbruck, Austria); Phonak Naida (Warrenville, Illinois, U.S.A.); Widex (Lynge, Denmark). CI = cochlear implant; F = female; HA = hearing aid; L = left-sided; M = male; R = right-sided.

**TABLE II.**

<table>
<thead>
<tr>
<th>Outcome Measures</th>
<th>Units</th>
<th>Mean Unaided</th>
<th>SD Unaided</th>
<th>Mean Aided</th>
<th>SD Aided</th>
<th>$P$ Value</th>
<th>Cohen’s $d$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Head Total RMS Sway</td>
<td>m/s²</td>
<td>0.263</td>
<td>0.0730</td>
<td>0.234</td>
<td>0.055</td>
<td>0.048</td>
<td>0.456</td>
</tr>
<tr>
<td>AP RMS Subcomponent</td>
<td>m/s²</td>
<td>0.220</td>
<td>0.0652</td>
<td>0.186</td>
<td>0.045</td>
<td>0.011</td>
<td>0.612</td>
</tr>
<tr>
<td>ML RMS subcomponent</td>
<td>m/s²</td>
<td>0.1404</td>
<td>0.0452</td>
<td>0.1385</td>
<td>0.042</td>
<td>0.735</td>
<td>0.043</td>
</tr>
<tr>
<td>Mean velocity</td>
<td>m/s</td>
<td>1.412</td>
<td>0.595</td>
<td>0.957</td>
<td>0.379</td>
<td>0.006</td>
<td>0.912</td>
</tr>
<tr>
<td>Path length</td>
<td>m/s²</td>
<td>34.19</td>
<td>13.29</td>
<td>34.96</td>
<td>12.30</td>
<td>0.094</td>
<td>–0.060</td>
</tr>
<tr>
<td>Lumbar Total RMS sway</td>
<td>m/s²</td>
<td>0.181</td>
<td>0.0676</td>
<td>0.1753</td>
<td>0.0628</td>
<td>0.542</td>
<td>0.088</td>
</tr>
<tr>
<td>Mean velocity</td>
<td>m/s</td>
<td>0.695</td>
<td>0.376</td>
<td>0.600</td>
<td>0.320</td>
<td>0.273</td>
<td>0.270</td>
</tr>
<tr>
<td>Path length</td>
<td>m/s²</td>
<td>16.49</td>
<td>8.071</td>
<td>17.05</td>
<td>8.188</td>
<td>0.080</td>
<td>–0.068</td>
</tr>
</tbody>
</table>

**Notes:** Measures in bold type were significantly different ($P < 0.05$) between the aided and unaided auditory conditions. Explanations of outcome measures may be found within the methods section. AP = anteroposterior; ML = mediolateral; RMS = root mean square; SD = standard deviation.
demonstrate that improvement had the lowest amount of baseline sway.

Previously, other studies have examined balance performance in unilateral implantees in the presence of sound. An early report showed a small improvement in performance on computerized dynamic posturography testing following cochlear implant activation, although statistical significance was not achieved. Cushing et al. measured performance on the Bruininks-Oseretsky Test of Motor Proficiency, Second Edition balance subtest. The results suggest a very small quantitative improvement in static and dynamic balance with implant processors on versus off, and their parents reported an anecdotal difference in their children’s balance when their children were wearing their implant processors. Jacob and Stelzig studied a cohort of patients receiving cochlear implants for their single-sided deafness and reported a 10% to 30% improvement in CoP sway when patients had their implant processors on relative to their performance with them off.

The data presented here add to our growing understanding that spatial audition can give a measurable balance improvement in people with normal and impaired hearing. A group of 14 older, experienced bilateral hearing aid users were shown to remain stable for a longer period of time standing on a solid surface or on foam while wearing their hearing aids compared to without. A more recent study, again investigating a population wearing hearing aids, used variability of the center of pressure as a main outcome measure and reported similar results. Here, we include three individuals with bimodal hearing (one cochlear implant and one hearing aid). Although our statistical power to draw comparisons for this subgroup is insufficient, principles of spatial hearing would suggest that two auditory inputs, regardless of the mode of hearing (bilateral vs. bimodal), are necessary for making use of spatial auditory landmarks. Given the importance of a spatial auditory map, unilateral implant recipients with bilateral hearing loss would likely receive less balance-related benefit from audition.

Despite many positive reports, the literature has not been entirely consistent regarding the magnitude of benefit with audition. Vitkovic et al. reported that greater availability of environmental sound generally improved balance, but they did not find that wearing hearing aids by itself was sufficient to bring out this effect. Stevens et al. found a statistically significant improvement in balance in the presence of enhanced auditory inputs, but the overall effect was dependent on three individuals who improved greatly and had baseline imbalance. Another group examined normal hearing patients, finding that the effect of sound was only strong when the participants were consciously attending to the stable sound source as a spatial reference point. These studies used CoP outcomes and showed variable results. Similarly, our lumbar measurements did not demonstrate a significant effect. However, in line with Zhong and Yost, we did find that head movements were reduced. This motivates future work to determine if it is primarily head stabilization, rather than body stabilization, that sound may improve.

An important contributor to variability in results among subjects may depend on the availability and reliability of other balance-related sensory inputs and the accuracy of their central integration. Here, we limited vision and proprioception, a clinical scenario commonly seen in other clinical populations, including the elderly and those with severe diabetes. In other experiments on integration of balance-related sensory cues, such as haptic and visual inputs, participants have been shown to differentially weight inputs based on their relative perceived accuracy to maintain balance. In line with these findings, we expect that audition would play a larger role in maintaining balance in proportion to the availability and reliability of auditory and other sensory cues.

Our analysis demonstrated that sway reduction as a result of audition via cochlear implants occurred primarily
In the AP direction, suggesting that implant-related auditory balance contributions may have a direction-dependent effect. Past work in a normal hearing population indicates that sound through headphones may reduce both AP and ML sway. It is not entirely surprising that cochlear implants may have a directional effect. The mechanism here relies on earth-fixed rather than head-fixed cues, and interaural level differences providing directional cues may be better in the frontal direction. Bilateral cochlear implant users struggle to make use of interaural timing differences, but further technological improvements in the field, such as synchronizing bilateral implant processors to retain useful temporal information for spatial hearing, could improve performance even more. A simpler explanation for particularly strong improvements in the AP direction is that there is less sway in the ML direction at baseline.

Few of our participants noted that they noticed a change in stability with their devices on, which is similar to previous findings. The subclinical benefit of audition could be because it is relatively down-weighted during typical daily activities that can depend on visual and proprioceptive inputs to balance. It may also be that auditory sensory input is so closely associated with “hearing” that people are unlikely to identify that it has other impacts on their lives.

This study only examined postural sway, a balance measure that is less likely than dynamic imbalance to be associated with falling and cannot reliably distinguish fallers from nonfallers. The balance improvements be associated with falling and cannot reliably distinguish that measure is less likely than dynamic imbalance to other impacts on their lives.

**CONCLUSION**

The findings presented here suggest that cochlear implants may function to improve head postural stability while standing and potentially reduce the risk of falling. These results join with past literature to suggest that auditory input may function as a fourth balance input (including vestibular, vision, and proprioception), and provide several suggestions as to the potential mechanism of auditory balance contributions.

**BIBLIOGRAPHY**


