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We investigated the contribution of auditory inputs to balance control in healthy young adults and older adults with normal hearing by simulating hearing loss, as well as in older adult with hearing loss by testing with and without hearing aids. Twenty healthy young adults with normal hearing, twenty older adults with normal hearing, and twenty older adults with hearing aids completed single- and dual- tasks consisting of a standardized audiology test (BKB-SIN) and maintaining standing balance in response to surface translations. Participants performed an auditory task of repeating back sentences from a standardized audiological test, the Bamford-Kowal-Bench Speech-In-Noise (BKB-SIN), played through wireless noise-cancelling headphones under randomized normal hearing and simulated hearing loss conditions or through surrounding speakers under hearing aid or no hearing aid condition. Simulated hearing loss was achieved using Adobe Audition software and a FFT logarithmic curve to manipulate sound volume and frequencies of standardized sentences according to age-related moderate hearing loss documented in literature. Backward surface translation perturbations inducing a forward loss of balance were synchronized with the auditory task and presented randomly at three levels ($0 m/s^2$, $2 m/s^2$, and $5 m/s^2$). Primary outcome measures included: maximum Center of Pressure – Center of Mass (COP-COM) distance in response to perturbation during the first compensatory step, reaction time for initiating the first compensatory step, number of steps after loss of balance, and performance on the BKB-SIN. Repeated measures ANOVA were conducted for each dependent variable with respect to perturbation level and auditory condition. Results show reaction time decreases, maximum COP-COM distance increases, and number of steps increases as perturbation level increases across all groups. BKB-SIN scores and reaction time were significantly worse under the simulated hearing loss condition. Hearing aids significantly improved BKB-SIN scores, but not balance scores. Hearing loss affects reactive balance control, particularly while simultaneously attending to auditory tasks. Older adults maintain the ability to initiate compensatory steps, but they require an increase number of steps to regain balance. Individuals with hearing loss may be at greater risk of falling compared to individuals with normal hearing due to age-related cognitive and neurodegenerative changes associated with hearing loss.

THE EFFECT OF HEARING LOSS ON BALANCE CONTROL

DISSERTATION

Presented to the Graduate Council of the

Graduate School of Biomedical Sciences

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For the Degree of

DOCTOR OF PHILOSOPHY

By

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CHAPTER 1

INTRODUCTION

I. INTRODUCTION

Falls are a prevalent problem in the United States, particularly in adults 65 years and older. In 2015, Medicare used $31 billion dollars to cover the cost of fall-related injuries (Burns, Stevens, & Lee, 2016). One-third of older adults fall annually on average, costing approximately $34 billion dollars for direct medical expenses related to medical procedures and hospitalization (Oliver, Daly, Martin, & McMurdo, 2004; Stevens, Corso, Finkelstein, & Miller, 2006). Falls are not only financially costly; falls also burden families taking care of the older adult, stress the constantly shrinking budget for Medicare, decrease the quality of life for the older adult, and may even lead to death of the older adult (Sherrington, Tiedemann, Fairhall, Close, & Lord, 2011; Shumway-Cook et al., 2009).

Another common problem prevalent in older adults is hearing loss. Age-related hearing loss affects approximately 51% of people aged 70-79 and 78% of those 80 and older (NIDCD, 2016). In the USA, hearing loss has expanded at a rate of 160% of the total population growth and continues to grow due to aging of the population. However, only 20-25% of hearing-impaired individuals seek to improve hearing with hearing aids (Kochkin, 2009; F. R. Lin et al., 2013). Evidence now suggests that older adults should address hearing loss, because untreated hearing loss may have indirect psychosocial and physical consequences, such as social isolation, depression, increased risk for falls, increased risk for hospitalization, and even mortality (Dalton et al., 2003; Feeny et al., 2012; Girard et al., 2014; Hidalgo et al., 2009; Lin & Ferrucci, 2012; Pryce & Gooberman-Hill, 2012).

Noise-Induced Hearing Loss (NIHL) is another common problem with older adults, a topic that will be discussed only briefly because it is not the focus of this dissertation. NIHL may occur due to an excessively loud noise, such as an explosion, but NIHL more commonly occurs from exposure to loud noise over an extended period of time (NIDCD, 2018). According to data from the 2009-2010 National Health and Nutrition Examination Survey, approximately 2% of adults 45-54 years old have disabling hearing loss, this rate increases to 8.5 % for adults aged 55-64 years old. Nearly 25% of adults 65-74 years old and 50% of adults 75 years and older have disabling hearing loss (Agrawal, Platz, & Niparko, 2008). Contrary to popular belief, hearing loss does not favor older adults. Within the United States of America, an estimated 15% of all individuals between the ages of 20 to 69 have NIHL. This percentage trend continues to rise, particularly with the younger population, due to use of personal listening devices (Widén, Båsjö, Möller, & Kähäri, 2017). Given excess noise often contributes to or accelerates age-related hearing loss; the close relationship between noise-induced and age-related hearing loss should be taken into account with older adults (Kujawa & Liberman, 2006; Yamasoba et al., 2013).

While it is now clear that both the incidence and negative consequences of hearing impairments and falls increase during aging, epidemiologic recent evidence suggests that rather than occurring in parallel, there may be a possible link between hearing loss and balance deficits (Viljanen, Kaprio, Pyykko, et al., 2009).

A. Literature Review: Hearing Loss and Postural Control Mechanisms

Traditionally, either an age-related or pathologically-related decline in sensory systems – specifically, the visual, vestibular, and somatosensory system – is considered to contribute to poorer control of balance and increased risk for falling in older adults (Manchester, Woollacott, Zederbauer-Hylton, & Marin, 1989). Recent epidemiological evidence suggests that changes in another sensory system, the auditory system, may contribute to or be associated with the control of balance and an increased risk of falling in older adults (Lin & Ferrucci, 2012; Viljanen, Kaprio, Pyykko, et al., 2009).

A literature search was performed to retrieve recent studies investigating the link between hearing loss, balance impairments and increased risk for falls in older adults, as well as the various mechanisms by which auditory impairments may affect balance. PubMed, Scopus, CINAHL, Cochrane, ScienceDirect, and Medline databases were queried for articles published between January 2000 and January 2018. The key terms used were: hearing loss, auditory impairment, older adults, elderly, balance, falls, hearing aids, hearing devices, gait, locomotion, cognition, and postural control. Article inclusion parameters required: (1) an available abstract, (2) human subjects only, (3) English language only, and (4) publication in an academic journal. A total of 541 articles were found. Two reviewers screened the articles based on the relevance to the topic and narrowed the inclusion number to 115 articles. Literature reviews and articles about pediatrics were excluded, which further narrowed the inclusion number to 80 articles.

Epidemiological evidence

As mentioned previously, falls due to poor postural control are a common problem within the older adult population, leading to many negative outcomes such as fractures, hospitalization, and even death (Kingma & Duis, 2000; Sihvonen, Era, & Helenius, 2004; Stel, Smit, Pluijm, & Lips, 2004; Wojszel & Bień, 2004). Recent epidemiological research supports the notion that hearing loss may affect postural control and leads to an increased risk for falls in older adults. Hearing loss has been more specifically linked to slow walking speed, poor static balance scores, poor quality of life, frailty, increased risk for injuries and hospitalizations due to falls, and increased risk for mortality (Table 1) (Anstey, Luszcz, Giles, & Andrews, 2001; Çakmur, 2015; Criter & Honaker, 2013; Feeny et al., 2012; Grue, et al., 2009; Kamil et al., 2016; Kulmala et al., 2009; Lacerda, E Silva, De Tavares Canto, & Cheik, 2012; Li, Simonsick, Ferrucci, & Lin, 2013; Lopez et al., 2011; Rumalla, Karim, & Hullar, 2015; Sihvonen et al., 2004; Skalska et al., 2013; Stephens & Ken, 2003; Viljanen, Kaprio, Pyykko, et al., 2009; Viljanen, Kaprio, Pyykkö, et al., 2009; Weaver, Shayman, & Hullar, 2017; Wojszel & Bień, 2004). Although this recent literature alerts clinicians and clinical researchers regarding the association between hearing loss and balance deficits, it fails to explain how and why hearing loss affects balance. Indeed, there is no clearly defined underlying mechanism establishing a cause and effect relationship and explaining *how* and *why* individuals with hearing loss fall more often than individuals with normal hearing. Several theories explaining the relationship between hearing loss and increased risk for falls have been proposed, including physiological, social, perceptual and cognitive mechanisms (Figure 1).

1) Physiological Mechanism:

The physiological mechanism has various hypotheses and sub-mechanisms explaining how hearing loss may cause balance deficits, ranging from a common blood supply between the vestibular and cochlear system, cross-talk between the vestibulocochlear nerve, solvent exposure, low bone mineral density of the inner ear, and a common gene that causes both hearing loss and muscular weakness (Table 2) (Agrawal et al., 2008; Kakarlapudi, Sawyer, & Staecker, 2003; J. Y. Kim, Lee, Lee, & Kim, 2016; Mendy et al., 2014; Oh et al., 2014; Purchase-Helzner et al., 2004; Zamyslowska-Szmytke, Politanski, & Sliwinska-Kowalska, 2011).

According to this mechanism, hearing loss could cause balance deficits due to the close connection of the vestibular nerve and the cochlear nerve, allowing for "cross-talk" between one another (Viljanen, Kaprio, Pyykkö, et al., 2009). Both nerves share the same blood supply and eventually run together and to form the afferent vestibulocochlear nerve; damage to the cochlear nerve could inherently affect the vestibular nerve (Rosenhall, 1973; Schuknecht & Gacek, 1993; Viljanen, Kaprio, Pyykkö, et al., 2009). Minimal evidence exists to refute or deny this sub-theory.

Another sub-mechanism discusses the notion of bone-mineral density being a coincidental link between hearing loss and balance deficits. Individuals with low Bone Mineral Density (BMD) may have poor bone mineral density globally, including the temporal bone that houses the inner ear bones (Mendy et al., 2014; Purchase-Helzner et al., 2004). Low BMD of the inner ear bones may coincidentally negatively affect hearing ability; poor bone mineral density of the lower extremity bones, such as the femur, may increase the risk for fractures related to falls (Mendy et al., 2014; Purchase-Helzner et al., 2004). Although little evidence exists regarding this sub-mechanism, the evidence is mixed.

Hearing loss and balance difficulties have also been linked to physiological changes in the inner ear, which includes but is not limited to: microvascular changes and ototoxic changes (Cunningham & Tucci, 2017). Recent evidence has linked hearing loss to diabetes mellitus, hypertension, and smoking due to microvascular changes in small arteriole blood vessels (Agrawal et al., 2008; Cruickshanks, Klein, et al., 1998; Kakarlapudi et al., 2003; Oh et al., 2014). The central and peripheral nerves of inner ear may also be affected by chronic, low-level ototoxic exposure of various industrial materials, chemicals, and solvents; however, the extent of neuronal damage is currently unknown (Estill, Rice, Morata, & Bhattacharya, 2017). Current evidence suggests a significant link exists between solvent exposure and vestibular hypofunction, increased postural sway, and workplace accidents that include slipping, tripping, or falling (Herpin et al., 2008; Hunting, Matanoski, Larson, & Wolford, 1991; Zamyslowska-Szmytke et al., 2011). Several therapeutic drugs have been well-documented to cause neurodegeneration of both the cochlear and vestibular nerve (Cunningham & Tucci, 2017). These drugs include, but are not limited to: antibiotics (i.e. aminoglycosides, vancomycin, and erythromycin), loop diuretics, non-steroidal anti-inflammatory drugs, malaria drugs, and cancer drugs (Rybak, 1986). Given the prevalence of these drugs taken by adults as they age, the role of ototoxic drugs in hearing loss and balance deficits may be currently overlooked or may confound results of future mechanistic research.

Lastly, gene inheritance or mutations are known to affect hearing loss (Cunningham & Tucci, 2017). Both hearing loss and muscular weakness has also been linked to the insulin-like growth factor 1 (IGF-1) gene (Barbieri et al., 2003; Cappola, Bandeen-Roche, Wand, Volpato, & Fried, 2001; Varela-Nieto et al., 2004; Viljanen, Kaprio, Pyykkö, et al., 2009). The potential effect of genes on both hearing loss and postural control cannot be ignored.

The number of physiological sub-mechanisms muddles the evidence and current literature does not focus on one particular topic. It is currently unknown whether hearing aids may improve hearing ability or postural control for individuals with hearing loss due to physiological mechanisms.

2) *Social Mechanism:*

An abundance of evidence links hearing loss to decreased balance control and risk for falls due to lack of socialization (Arlinger, 2003). Consequences of lack of socialization due to hearing loss has been associated with depression, decreased ability to perform Activities of Daily Living (ADLs) and Instrumental Activities of Daily Living (IADLs), and reported difficulty with functional mobility. (Table 3) (Bazargan, Baker, & Bazargan, 2001; Brink & Stones, 2007; Chen et al., 2014; Chia et al., 2007; Dawes et al., 2015; Grue, et al., 2009; Heyl & Wahl, 2012; Hidalgo et al., 2009; Hogan, O'Loughlin, Miller, & Kendig, 2009; Hung, Ross, Boockvar, & Siu, 2012; Jagger, Spiers, & Arthur, 2005; Kiely, Anstey, & Luszcz, 2013; Lacerda et al., 2012; M. Y. Lin et al., 2004; Lopez et al., 2011; Loprinzi, Smit, Lin, Gilham, & Ramulu, 2013; Lupsakko, Kautiainen, & Sulkava, 2005; Mikkola et al., 2015; Pryce & Gooberman-Hill, 2012; Tomioka et al., 2015). Older adults with hearing loss experience the vicious cycle of social isolation due to difficulty hearing and communicating (Crews & Campbell, 2004; Dawes et al., 2015). Social isolation is known to lead to general deconditioning and weakness; thus leading to balance difficulty and an increased risk for falls, secondary to the general deconditioning and weakness, with older adults (Brink & Stones, 2007).

Not only has hearing loss been associated with social isolation and decreased social engagement due to difficulty with communication, but also a higher incidence of poorer mood and depression have been reported in older adults with hearing loss compared to older adults with normal hearing (Brink & Stones, 2007; Dawes et al., 2015; Kiely et al., 2013). Regarding physical functioning and overall health, individuals with hearing loss have also reported decreased quality of life, difficulty with Activities of Daily Living (ADLs) (i.e. bathing, dressing, eating, toileting), and Instrumental Activities of Daily Living (IADLs) (i.e. driving, shopping, laundry) (Grue, Ranhoff, et al., 2009; Heyl & Wahl, 2012; Hidalgo et al., 2009; Jagger et al., 2005; Kiely et al., 2013).

3) Perceptual Mechanism:

Hearing loss may be associated with poor balance control and increased risk for falls due to inability to localize potentially hazardous sounds (Lau, Pichora-Fuller, Li, Singh, & Campos, 2016). Hearing loss may create an incomplete or inaccurate representation of environmental sounds (ie. the proximity a fire truck's siren), putting the individual at risk of unexpected events that could lead to a fall (Arlinger, 2003; Cox & Alexander, 1995; Girard et al., 2014; Lundälv, 2004). An individual often uses sound to discern the external environment and changes within the environment (Palmer, D'angelo, Harris, Linaker, & Coggon, 2015). The perceptual mechanism suggests an individual with hearing loss has difficulty perceiving the environment around him or her, preventing an accurate representation of the environment (Lundälv, 2004). An individual with hearing loss may perceive a noise that is in close proximity as being distant, so a signal, such as a person coming behind, may startle the individual with hearing loss, potentially causing a loss of balance and a fall (Arlinger, 2003; Girard et al., 2014). Studies assessing the perceptual mechanism have ranged from finding hearing loss is associated with increased risk for workplace injuries to studying standing balance tasks with varying types of noise (Table 4) (Girard et al., 2014; Kanegaonkar, Amin, & Clarke, 2012; Lau et al., 2016; Negahban & Nassadj, 2017; Palmer et al., 2015; Vitkovic, Le, Lee, & Clark, 2016).

Hearing loss has not only been associated with an increased number of falls, but also varying types of injurious accidents (M. Picard et al., 2008). Noise-induced hearing loss, specifically, has been significantly correlated with a greater risk for industrial work-related accidents, which increases as the number and amount of environmental hazards increase, due to inability to perceive dangerous sounds (M. Picard et al., 2008; Zwerling et al., 2000). Individuals with noise-induced hearing loss have also been associated with an increased risk for traffic accidents and non-speeding traffic citations due to inability or inaccurate perception of dangerous noises (ie. car horn), potentially in combination with divided attention from various environmental sounds (Michel Picard et al., 2008). Pedestrians and cyclists with hearing loss are also at an increased risk for traffic-related accidents. Furthermore, those who use hearing aids would often turn off hearing aids when environmental sounds became too distracting, further putting these individuals at risk for an accident due to increased difficulty hearing dangerous sounds (Lundälv, 2004). Currently, not enough evidence exists to determine whether hearing aids would reduce the number of accidents and falls for individuals with hearing loss.

New literature is emerging that attempts to link auditory input to balance control through sound localization under various balance conditions (Jayakody, Friedland, Eielboom, Martins, & Sohrabi, 2017). Young and older adults appear to have increased sway when sound is distorted; however, older adults with hearing loss may not be as affected by localization of auditory input due to inability to perceive it (Jayakody et al., 2017; Lau et al., 2016; Vitkovic et al., 2016). More research is needed to determine whether poor sound localization in individuals with hearing loss causes impaired balance control and increased risk for falls.

4) Cognitive Mechanism:

The cognitive mechanism suggests that individuals with hearing loss have difficulty maintaining postural control while attending to speech and sounds. This process becomes a dual-task in which the person reallocates or divides attention otherwise used for balance to processing sounds and hear more accurately, leading to balance issues and an increased risk for falls with older adults (Kerr, Condon, & McDonald, 1985; Lajoie, Teasdale, Bard, & Fleury, 1993; Lundin-Olsson, Nyberg, & Gustafson, 1997; Shumway-Cook & Woollacott, 2000;

Viljanen, Kaprio, Pyykko, et al., 2009; Viljanen, Kaprio, Pyykkö, et al., 2009; Woollacott & Shumway-Cook, 2002). Studies assessing the cognitive mechanism have used variable assessments to come to the following conclusions: hearing loss is associated with poorer Mini-Mental State Exam (MMSE) scores, poorer self-report of cognitive abilities, poorer cognitive performance on various cognitive tests and dual-tasks, poor verbal working memory and executive function abilities, and dementia (Table 5) (AKYİĞİT et al., 2014; Bazargan et al., 2001; Brink & Stones, 2007; Bruce et al., 2017; Bush, Lister, Lin, Betz, & Edwards, 2015; Chia et al., 2007; Choi, Shim, Lee, Yoon, & Joo, 2011; Da, Lee, & Lee, 2015; Davis, 2003; Dawes et al., 2015; Dupuis et al., 2015; Edwards et al., 2016; Gatehouse, Naylor, & Elberling, 2003; Glyde, Cameron, Dillon, Hickson, & Seeto, 2013; Grue, et al., 2009; Gurgel et al., 2014; Gussekloo, de Craen, Oduber, van Boxtel, & Westendorp, 2005; Hallgren, Larsby, Lyxell, & Arlinger, 2005; Heyl & Wahl, 2012; Hidalgo et al., 2009; Hogan et al., 2009; L. E. Humes, 2002; Larsby, Hallgren, & Lyxell, 2008; Larsby, Hallgren, Lyxell, & Arlinger, 2005; Lau et al., 2016; F. R. Lin et al., 2011; F. R. Lin et al., 2013; M. Y. Lin et al., 2004; Lopez et al., 2011; Lunner, 2003; Lupsakko et al., 2005; McCoy et al., 2005; Mikkola et al., 2015; Moore et al., 2014; Ng, Rudner, Lunner, Pedersen, & Ronnberg, 2013; Pearman, Friedman, Brooks, & Yesavage, 2000; Ronnberg et al., 2011; Rudner, Lunner, Behrens, Thorén, & Rönnberg, 2012; Shahidipour, Geshani, Jafari, Jalaie, & Khosravifard, 2013; Tomioka et al., 2015; Tun, Benichov, & Wingfield, 2010; van Hooren et al., 2005; Wollesen et al., 2018; Wu, Stangl, Zhang, Perkins, & Eilers, 2016; Zekveld, Deijen, Goverts, & Kramer, 2007; Zekveld, Kramer, & Festen, 2011).

Although few studies directly investigated hearing loss and postural dual-tasks, strong evidence links hearing loss to cognitive decline, dementia, and potentially even psychiatric disorders in older adults. The reason behind how and why hearing loss is associated with cognitive decline is unknown (Blazer, 2018; F.R. Lin et al., 2011). Four major explanations of how hearing loss is associated with cognitive decline currently exists, the: 1) *common cause*, 2) *information-degradation*, 3) *sensory-deprivation*, and 4) the *cognitive-load-on-perception* (L.E. Humes, Busey, Craig, & Kewley-Port, 2013; Mudar & Husain, 2016; Wayne & Johnsrude, 2015). According to the "common cause" explanation, a global decline of hearing and cognitive function occurs simultaneously and the correlation is coincidental (L.E. Humes et al., 2013). The information-degradation explanation suggests increased auditory processing overloads the auditory cortex and other cognitive functions cannot perform optimally. The sensory-deprivation explanation suggests increased auditory processing causes neurodegenerative changes of the auditory cortex and the cognitive regions of the brain, leading to cognitive decline (Wayne & Johnsrude, 2015). Neurodegenerative changes in the frontal lobe and the limbic region of the brain in individuals with hearing loss have been reported, as well, which may support either the information-degradation or the sensory-deprivation explanations (Mudar & Husain, 2016). The cognitive-load-on-perception explanation suggests cognitive decline leads to less cognitive resources available for auditory processing, which appears as hearing loss (Wayne & Johnsrude, 2015).

Clinicians and clinical researchers, in particular, have studied the association between hearing loss and cognitive functioning extensively (V. Y. Lin et al., 2017). Hearing loss has been widely associated with increased listening effort, which suggests an increased amount of cognitive resources must be reallocated to listen and process speech (Gosselin & Gagné, 2010). Hearing loss has been widely associated with poorer performance on short-term memory tasks (Ronnberg et al., 2011) and with lower scores on outcome measures, such as the mental component of the Short Form (SF)-36 and the Mini Mental State Exam (MMSE) (Chia et al., 2007; Hogan et al., 2009; F. R. Lin et al., 2011; M. Y. Lin et al., 2004; Lopez et al., 2011).

State of Current Evidence

In summary, the physiological mechanism has a varying number of sub-theories, which makes determining the strength of the evidence complicated. Not enough evidence exists to support or refute the perceptual mechanism. A large amount of evidence suggests hearing loss and balance difficulty may be associated with the social mechanism and research is beginning to test interventions (ie. social engagement) (Bazargan et al., 2001; Mudar & Husain, 2016). Strong evidence suggests hearing loss and balance difficulty may be associated with the cognitive mechanism (Bruce et al., 2017); however, mixed evidence exists regarding the effectiveness of hearing aids on communication abilities of older adults (Brink & Stones, 2007; Bruce et al., 2017; Vitkovic et al., 2016). Research is also attempting to create comprehensive auditory rehabilitation to prevent potential negative effects of hearing loss and cognitive decline (Mudar & Husain, 2016).

Limited evidence exists to determine whether the correlation between hearing loss and increased risk for falls, particularly with older adults, is merely coincidental or is cause-and-effect (Bruce et al., 2017). Epidemiological and more recent balance research have identified specific types of balance deficits (Li et al., 2013). Individuals with hearing loss have a slower gait speed, self-report poor physical mobility and increased number of falls compared to normal hearing individuals, and have increased Center of Pressure (COP) sway measurements during quiet stance with background noise (Sogebi, Oluwole, & Mabifah, 2015; Viljanen, Kaprio, Pyykkö, et al., 2009; Vitkovic et al., 2016). However, there is a paucity of evidence regarding the effect of hearing impairment on reactive balance, namely the ability to regain balance after a stumble, trip, slip, or near loss of balance (Bruce et al., 2017). In a real-world setting, individuals with hearing loss are constantly performing a dual-task of attending to auditory sounds, such as speech, while simultaneously attempting to stand, walk, or cross obstacles. An individual with hearing loss may either be distracted by or use a high level of processing resources to attend to environmental sounds, therefore being more at risk of loss of balance and falling.

B. Balance and Postural Control Assessment

Postural Control Outcome Measures

Both anticipatory and reactive balance controls are regulated by the Central Nervous System (CNS) in order to maintain posture (Kanekar & Aruin, 2014). Anticipatory responses are internal, voluntary initiations performed prior to an anticipated movement, in which the subconscious choice of a particular movement is based on prior experience (Kanekar & Aruin, 2014; Patla, Ishac, & Winter, 2002). Reactive responses respond to an external or unexpected disturbance, such as a loss of balance, and work toward regaining stability and equilibrium within the balance system (Kanekar & Aruin, 2014; S.-I. Lin & Woollacott, 2005). Falls most often occur during situations that require reactive balance control, such as tripping or slipping, particularly with older adults (Niino, Tsuzuku, Ando, & Shimokata, 2000; Papa, Garg, & Dibble, 2015). Clinical research has begun to focus on studying reactive balance control in an attempt to identify older adults at fall risk and reduce the risk for falling in the older adult population (Carty et al., 2014; Dijkstra, Horak, Kamsma, & Peterson, 2015; Paquette, Li, Hoekstra, & Bravo, 2015).

Laboratory experiments investigating reactive balance create a life-like type of loss of balance in a safe and controlled environmental setting. Biomechanical outcome measures commonly utilized in research settings to assess reactive balance abilities include: maximum Center of Pressure (COP) – Center of Mass (COM) distance during compensatory steps, reaction time to initiate the first compensatory step, and number of steps after a loss of balance (Burleigh, Horak, & Malouin, 1994; Horak, Dimitrova, & Nutt, 2005; Kanekar & Aruin, 2014; Mansfield, Peters, Liu, & Maki, 2010; McIlroy & Maki, 1996).

Center of Pressure (COP) is the collected average of the pressure from the bottom of the feet and is a measure of the motor system and is often described in research as ‘sway’ (Ruhe, Fejer, & Walker, 2011). Center of Mass (COM) is a kinematic measure considered the weighted average of all joint segments moving at a particular moment and this measure favors a position of stability for a person, which typically is located around the sacrum (Winter, 1979; F. Yang & Pai, 2014). COP and COM interact closely with one another to maintain postural stability during both anticipatory stepping (ie. gait initiation) and compensatory stepping (ie. unexpected loss of balance) (Shumway-Cook & Woollacott, 2007). Compensatory stepping occurs with an initial COM displacement followed by COP displacement that regains balance equilibrium and maintains postural stability (Henry, Fung, & Horak, 2001; Horak et al., 2005; Santos, Kanekar, & Aruin, 2010; Winter, Patla, & Frank, 1990; J. Yang, Winter, & Wells, 1990). Maximal displacements of COP and COM (also known as peak COP and peak COM) have been individually documented in the literature as stability measures. Another important stability measure documented in the literature is the stability margin, the difference between COP and COM (COP-COM) (Jacobs, Dimitrova, Nutt, & Horak, 2005; Kanekar & Aruin, 2014; Santos et al., 2010; Winter, 1979). The COP-COM maximum distance is considered an indicator of robustness in the balance control system (Corriveau, Hébert, Prince, & Raîche, 2001; Jančová, 2008; Lafond, Duarte, & Prince, 2004; Papa, Foreman, & Dibble, 2015; Winter, 1995; Winter, Prince, Frank, Powell, & Zabjek, 1996). COP and COM are interconnected during compensatory stepping; therefore, COP-COM was chosen as the most appropriate outcome measure of postural stability during unexpected compensatory steps (Kanekar & Aruin, 2014).

Reaction time is an important predictor for falls in older adults, in which a slower reaction time indicates a greater likelihood of falling (Lord & Fitzpatrick, 2001). Furthermore, reaction time indicates the quality neuromuscular and physiological response to a sudden and unexpected loss of balance, particularly a loss of balance requiring compensatory steps to regain balance, with a quicker reaction time indicating better control of balance and decreased risk for falling (Mansfield et al., 2010). Reaction time is typically assessed through use of Electromyography (EMG) to measure muscle activation, or through kinetic or kinematic assessments using force plates or marker analysis (Do & Roby-Brami, 1991; McIlroy & Maki, 1996). Reaction time is considered such an important fall risk outcome measure that it's even been suggested that reaction time is a better clinical indicator of fall risk than walking speed (van den Bogert, Pavol, & Grabiner, 2002). We, therefore, chose reaction time during the first compensatory step as an appropriate outcome measure to utilize to assess fall risk with a slower reaction time indicating worse control of balance.

Number of steps is an observable clinical outcome measure that can be used when administering reactive balance tests, such as the Nudge Test, to identify an older adult faller (Granacher, Muehlbauer, & Gruber, 2012; Stone & Skubic, 2011). An increased number of recovery steps after an unexpected loss of balance are associated with an increased risk for falling (Crenshaw & Kaufman, 2014; Hilliard et al., 2008; Pai, Rogers, Patton, Cain, & Hanke, 1998). Number of steps during loss of balance is a simple and affordable clinical test that can be performed in clinical settings without the use of expensive technology (Colagiorgio et al., 2014). We determined number of steps after loss of balance would be an appropriate outcome measure to use due to its ability to be clinically translatable.

Dual-Task Paradigm

Because balance and mobility are underpinnings for functional performance in real life (walking in a grocery store while talking on the phone and shopping at the same time), dual-task paradigms have been used to determine the impact of impairment in a complex system (McFadyen, Gagné, Cossette, & Ouellet, 2017). In the classic dual-task paradigm, performance on each task is measured in isolation (single task) and while performed concurrently (dual-task) with the aim of determining task interference and inferring attentional/cognitive prioritization and implications for safety (Silsupadol et al., 2009). A large number of studies using dual-task paradigms studies have investigated balance and postural control with walking, driving, reaching and grasping, while simultaneously performing other cognitive tasks (counting backwards, remembering words and colors, simple conversation) in a variety of patient populations from healthy young adults to older adults, patients with stroke, Parkinson's disease, Alzheimer's disease, etc. (Bowen et al., 2001; Creem & Proffitt, 2001; Hollman, Kovash, Kubik, & Linbo, 2007; Muir et al., 2012; Penko et al., 2018; Strayer & Johnston, 2001)

We created a novel auditory and balance dual-task paradigm to investigate the effect of hearing loss on reactive balance control. We intend to test whether a cognitive mechanism explains why hearing aids improve balance for older adults with hearing loss (Figure 2).

C. <u>Audiology assessment</u>

The most common form of audiological assessment performed by audiologists is an audiogram. An audiogram displays the frequency, or pitch, and the hearing threshold level, or volume, the subject hears. The subject is exposed to several volume levels at various frequencies. The average hearing threshold level is calculated by the audiologist to diagnose the extent of hearing loss. An individual with normal hearing has an average hearing threshold level below 25 decibels (dB) Hearing Level (HL). An individual with mild hearing loss has an average hearing threshold level between 26-40 dB HL. An individual with moderate hearing loss has an average hearing threshold level between 41-70 dB HL. An individual with severe hearing loss has an average hearing threshold level between 71-90 dB HL. An individual with profound hearing loss has an average hearing threshold level greater than 91 dB HL (ASHA, 2015a; Clark, 1981).

A device, known as an audiometer, records the data through a process known as pure-tone testing. Pure-tone testing often uses both air-conduction testing and bone-conduction testing to determine the type of hearing loss. A subject undergoes air-conduction testing to determine if the individual has sensorineural hearing loss. During air-conduction testing, the subject wears headphones or insert-phones and pushes a button or raises a hand to indicate a sound has been detected. During bone-conduction testing, the subject wears a bone oscillator, which passes sound directly to the inner ear consisting of the cochlea and vestibular system, bypassing the outer and middle ear. The pitch and volume is recorded on the audiogram during both types of testing and is further analyzed by the audiologist to determine type of hearing loss (ASHA, 2015b; Isaacson, 2010).

An audiometer measures the range of pitch and volume an individual can hear. Audiometers can be hand-held, portable, or can be set-up within an audiology clinic using a sound-attenuated booth. Most hand-held and portable audiometers have a set of headphones, which the participant wears, and are utilized in a quiet setting like an empty room. The handheld audiometer typically has less frequency and decibel options and sometimes has an otoscope placed in the patient's ear instead of using headphones. Hand-held and portable devices are commonly used as a screening tool in primary care clinics. Audiometers in the audiology clinic are often used within a sound-attenuated booth (ASHA, 2015b).

Other common forms of audiological screening assessments include: Auditory Brainstem Response (ABR), Otoacoustic Emissions (OAEs), and Speech Reception Threshold (SRT). These tests are performed specifically in audiology clinics. These tests – in combination with other testing methods – are used for differential diagnosis to determine if an older adult has sensorineural hearing loss, conductive hearing loss, or other ear-related pathologies, such as Ménière's disease or an acoustic neuroma. An audiologist uses OAEs to determine cochlear functioning. An audiologist inserts a device into a patient's inner ear, the device sends vibrations and stimulates the cochlea, and the cochlea responds by creating OAEs. The audiologist then determines cochlear function based on the cochlea's response to the signals. Lack of signaling to the probe indicates damage to the cochlea's hair cells (ASHA, 2015b; Isaacson, 2010; Kemp, 2002; Norton et al., 2000). SRT is often used in conjunction with pure-tone testing to confirm the degree of sensorineural hearing loss. The testing procedure requires the subject to repeat various words or phrases presented at various volumes over a loudspeaker or headphones in either silence or background noises. The Signal-to-Noise Ratio (SNR) is a measure of the volume of speech an individual can hear in a particular level background noise. It is assessed by an audiologist to determine the degree of hearing loss (ASHA, 2015b; Isaacson, 2010).

The auditory task selected for single- and dual-task testing was the standardized audiology test, Bamford-Kowal-Bench Speech-In-Noise-Test (BKB-SIN), consisting of a target voice and multi-talker babble. The test is commonly used to determine whether an individual with hearing loss would benefit from a hearing aid or cochlear implant (Litovsky, 2011; Wilson, McArdle, & Smith, 2007). The test consists of simple sentences such as, "The truck drove up the road." Subjects are scored on how accurately they can repeat back the underlined words, which consists of either three or four words per sentence. Scores are tallied and the total score is subtracted from 23.5, providing a speech-in-noise ratio that the test-taker will correctly repeat 50% of the sentences (Bench, Kowal, & Bamford, 1979). A speech-in-noise ratio is the volume of speech relative to the background noise , with +0 being the speech and background noise are the same level and +10 being the speech is 10 dB higher than the background noise (McShefferty, Whitmer, & Akeroyd, 2016). A higher score on the BKB-SIN indicates a poorer performance (Bench et al., 1979).

Hearing loss was simulated using Adobe Audition. Five second clips of each sentence from the standardized audiology test, BKB-SIN, were uploaded into the program. The BKB-SIN consists of a target voice and multi-talker babble/noise. The voice and the multi-talker babble were separated from 1 track into 2 separate tracks with 1 track constituting the target voice and 1 track constituting the multi-talker babble. The decibels were manipulated at particular frequencies associated with moderate hearing loss. Moderate hearing loss values of decibels per frequency were obtained from The National Institute for Occupational Safety and Health (NIOSH) Hearing Loss Simulator (H. P. Kim et al., 2011). Moderate hearing loss was simulated by applying a Fast Fourier Transform (FFT) filter (Logarithmic scale, FFT size: 2048, Blackman window) in Adobe Audition to the separated track of the voice, according to previous research simulating hearing loss (Cruickshanks, Wiley, et al., 1998; Hornsby, Johnson, & Picou, 2011; Korhonen & Kuk, 2008; McPherson, McMahon, Wilson, & Copland, 2012).

Subsequently, the voice and babble/noise were recombined in one file that maintained the Speech-in-Noise (SIN) ratio associated with each sentence of the BKB-SIN test. A total of 3 lists each containing 8 short sentences were manipulated to simulate hearing loss or left in the original state for subjects with hearing aids to perform; the other 3 lists of the BKB-SIN were used in their original state for a normal hearing or no hearing aid condition. No sentence was heard more than one time by each subject in the study. Subjects used Bose® QuietComfort 35 wireless headphones to listen to sentences and limit any additional environmental noise during testing.

Subjects were required to stand and maintain their balance following unexpected surface translations while simultaneously listening and repeating back sentences from the BKB-SIN at an intensity level of 60 dBA. There were three auditory conditions: 1) no audio sound, no repeat back, resulting in the single task of maintaining balance; 2) normal hearing or with hearing aid; and 3) hearing loss or no hearing aid. The conditions were randomized for the normal hearing subjects and were randomly assigned for the hearing loss subjects.

D. Hearing Aids

Types of Hearing Aids

Hearing aids are devices placed either in or behind the ear that improve listening abilities for an individual with sensorineural hearing loss. Three basic components assemble a hearing aid: a microphone, amplifier, and speaker. The microphone receives the sound waves of speech or noise and transmits the signals into the amplifier. The amplifier converts these waves into electrical signals and transmits these signals to the amplifier. The amplifier magnifies the signals for the remaining hair cells. The hair cells convert the signals into neural inputs sent to the brain for processing (NIDCD, 2017).

Presently, two major types of hearing aid circuitry exist on the market: analog hearing aids or digital hearing aids. Analog hearing aids enhance the volume of all sounds without differentiating between sound of interest vs. noise in various environments, such as a quiet museum or a noisy stadium. The hearing aid user can manually adjust between settings on the hearing aids to improve sound quality based on his or her current environment. Digital hearing aids amplify sound signals, and have the ability to selectively amplify specific sound frequencies based on the hearing aid user's needs. Some digital hearing aids can also amplify sound coming from a particular direction (H. H. Kim & Barrs, 2006).

Frequency Modulated (FM) systems are a common accessory to hearing aids used by individuals with hearing loss who have exceptional difficulty hearing speech in a noisy environment (Chisolm, Noe, McArdle, & Abrams, 2007). FM systems are designed to perform 3 major functions: 1) Limit background noise and distractors, 2) increase the volume of speech, and 3) maintain the same volume of speech, regardless of the location of the talker (A.

Boothroyd, 2004; A. Boothroyd & Ross, 1992). FM systems typically are device with a directional microphone wirelessly linked to a hearing aid, in which the sound from the microphone is sent directly into the hearing aid of the hearing aid user (Thibodeau, 2010). These devices allow individuals with hearing loss to improve their understanding of speech in a noisy setting, such as a restaurant (Schafer & Thibodeau, 2006).

Various styles of hearing aids exist and are chosen by an audiologist and patient based on each patient's personal wants and needs. "Behind The Ear" (BTE) hearing aids consist of a computer device fitted behind the ear with the electrical portion fitted in the outer ear. Recently, a smaller and more esthetically pleasing version known as the "Mini" BTE hearing aid has been developed. "In The Ear" (ITE) hearing aids fit completely within the outer ear. "In the Canal" (ITC) hearing aids fit within the ear canal and "Completely-In-Canal" (CIC) hearing aids are fully inserted into the ear canal. Canal hearing aids are primarily effective for mild or moderate hearing loss (NIDCD, 2017; H. H. Kim & Barrs, 2006).

The FDA renewed a bill through the United States Senate in 2017 that will allow hearing aids to be purchased over-the-counter and will be suitable for individuals with mild to moderate hearing loss. Individuals with hearing loss will still be able to go to an audiologist to have the hearing aid serviced, but do not need to be fitted for a hearing aid (Thomas, 2017; Warren & Grassley, 2017). Currently, the topic of purchasing hearing aids over the counter is highly controversial and the amount of adherence by over the counter hearing aid users is yet to be determined (The Hearing Review, 2017).

Reasons for Non-Use of Hearing Aids

As mentioned previously, many older adults who need hearing aids do not wear hearing aids (Lupsakko et al., 2005). Less than 4% of older adults with sensorineural or conductive hearing loss wear hearing aids, which equals an estimated 22.9 million Americans. According to The National Academy Aging Society Group (1999), inadequate Medicare and insurance coverage of hearing aids for seniors is a major contributory factor as to why the majority of older adults do not wear hearing aids. Furthermore, lack of hearing aid use is linked with lower socioeconomic status and lower level education level; therefore, many individuals who need hearing aids either cannot afford the devices or may not know about the benefits of a hearing aid (Bazargan et al., 2001).

Many older adults who are fortunate enough to receive hearing aids are resistant to hearing aid prescription or chose not to wear prescribed hearing aids (Bazargan et al., 2001; Hidalgo et al., 2009; Li et al., 2013; Lupsakko et al., 2005; Pryce & Gooberman-Hill, 2012). Several reasons exist for lack of hearing aid use. The cochlear nerve degrades slowly throughout a lifespan so many older adults are unaware of the extent of their hearing disability and often attribute their hearing loss due to general aging, which causes resistance to wearing hearing aids (Bazargan et al., 2001; Davis, 2003). Resistance to hearing aid use may also be due to self-consciousness or stigmatism of looking or feeling “old” with a hearing aid (Franks & Beckmann, 1985; McCormack & Fortnum, 2013; Meister, Walger, Brehmer, von Wedel, & von Wedel, 2008). It should be noted that hearing aids are beginning to look more discrete and aesthetically pleasing so stigma is not as big of an issue compared to previous years (McCormack & Fortnum, 2013).

A first-time hearing aid user also does not adjust to the device immediately. Adjustment takes a few months requiring neuroplasticity and often tweaking of the hearing aid device to suit the individual. Hearing aids do not completely resolve of hearing loss and hearing aid users may experience inconveniences, such as amplified distorting speech in a noisy environment or buzzing while using a cell phone (NIDCD, 2017). A hearing aid user may find these inconveniences to outweigh the believed benefits of the hearing aid and lead to disuse (Hallgren et al., 2005; Willott, 1996).

One study performed by Lupsakko et al. (2005) found that 25% of subjects who had hearing aids did not wear their listening devices. Subjective reasons as to why these individuals did not use their hearing aids regularly included: 42% of subjects reported they felt a hearing aid did not benefit them and using the device was unnecessary, 21% of subjects reported the hearing aids were too hard to use, and 17% of subjects reported their hearing aid was defective.

Hearing Aids and Socialization

Unfortunately, hearing aids may or may not improve socialization of older adults with hearing loss (Pryce & Gooberman-Hill, 2012). A barrier preventing older adults to receive a hearing aid is the cost of hearing aids or hearing interventions are expensive and not fully covered by Medicare or Medicaid (Cohen-Mansfield & Infeld, 2006). Lack of coverage for hearing aids forces individuals with hearing loss to pay more out-of-pocket annual expenses for medical care compared to individuals with normal hearing, creating a major barrier for hearing aid adoption among older adults (Cunningham & Tucci, 2017). For instance, one study by Kochkin (2007) suggests 76% of individuals who did not adopt hearing aids report inability to afford hearing aids. Individuals unable to afford hearing aids earned approximately $20,000-

$45,000 less annually compared to age-matched individuals who reported the ability to afford hearing aids, creating a health disparity among individuals of lower socioeconomic status.

Hearing Aids and Cognitive Function

Currently, there is mixed evidence as to whether hearing aids may improve cognitive function given that hearing aids do not fully restore hearing abilities. Further confounding the hearing aid controversy, high working memory may influence listening abilities and reduce the benefit of a hearing aid (Chia et al., 2007; Hallgren et al., 2005; Lupsakko et al., 2005; Meister et al., 2008; Ng et al., 2013). Promising recent evidence, on the contrary, suggests that hearing aids as part of a comprehensive rehabilitation program may improve listening ability and slow or prevent cognitive decline for older adults with hearing loss (Mudar & Husain, 2016). The interventions of the comprehensive rehabilitation program include, improving social engagement (i.e. outside home activities), hearing amplification (i.e. hearing aids), cognitive training focused on both verbal and non-verbal tasks in which individuals with hearing loss have the greatest amount of difficulty, and auditory training targeted at improving speech intelligibility (Mudar & Husain, 2016). Further research is needed to determine if a comprehensive auditory rehabilitation program could improve both listening ability and cognitive function for hearing aid users.

II. SIGNIFICANCE and INOVATION

<u>Significance</u>

Individuals with hearing loss may be at greater risk of falling than individuals without hearing loss and the magnitude of hearing loss may relate to the level of fall risk (Gerson, Jarjoura & McCord, 1989; Grue et al., 2009). Viljanen, Kaprio, Pyykko, et al. (2009) reported older adults with poorer hearing to be at a 3-4x higher risk of falling compared to those with better hearing. It has also been shown that two-thirds of patients who have sustained a hip fracture also had a hearing impairment at the time (Grue, Kirkevold, & Ranhoff, 2009); moreover, of those with hearing loss who reported suffering from additional limitations, the most commonly reported limitations were related to “mobility” (65%) compared to less frequently reported limitations (communication 12%, memory 12%, learning 11%) (Stats Canada, 2006).

Determining how hearing loss affects balance in older adults is significant because it is expected to determine if a cognitive mechanism partially or fully explains the link between hearing loss and balance deficits. As mentioned previously, the cognitive mechanism theorizes hearing loss requires an individual to reallocate resources used for maintaining balance towards listening to speech and essential sounds in the environment. This contribution is also significant because it is expected to determine if hearing aids decrease the cognitive demand from hearing loss, allowing an older adult to reallocate cognitive resources formally required for listening towards maintaining balance; thus preventing falls.

Decreasing the number of falls within the older adult community will prevent consequences from fractures, hospitalization, medical procedures, family burden, and financial stress (Oliver et al., 2004; Sherrington et al., 2011; Shumway-Cook et al., 2009; Stevens et al., 2006). In addition, preventing an older adult from falling may lead to positive outcomes, such as improved quality of life, increased mobility, decreased social isolation, and decreased level of depression (Brink & Stones, 2007). Revealing a mechanism to explain the link between hearing loss and balance deficits will also open the door for interventions, such as better acceptance or new types of auditory interventions, to reduce the number of falls in the older adult population.

III. SPECIFIC AIMS

The purpose of this project is to investigate the contribution of auditory input to the control of balance and to determine how and why hearing loss contributes to loss of balance and falls. We will use sophisticated virtual environments to conduct standardized speech recognition and sound localization tests while moving and maintaining balance in realistic, yet controlled conditions. We hypothesize that individuals with hearing loss have poorer postural control compared to individuals without hearing loss because attempting to understand speech or discern sounds while maintaining balance becomes a dual-task. The dual-task requires individuals to re-allocate resources required to maintain balance towards listening and sound processing.

Sample size was calculated from pilot data and standard clinical normative values, using. G*Power 3.1.9.2 based on an ANOVA: Fixed effects, special, main effects, and interactions. Sample size was calculated for gait speed using pilot data for gait speed of older adults (1.5 m/s) and older adults with hearing loss (1.35m/s), as well as normative values for young, healthy adults (1.6 m/s) (Oberg, Karsznia, & Oberg, 1993) (effect size = 0.58, $\alpha = 0.05$, Power = 0.80, Numerator df = 5, and number of groups = 6), leading to a total sample size of 45 subjects required. Accounting for 33% attrition, the total sample size will be 60 subjects total (20 young healthy adults, 20 older adults with normal hearing, and 20 older adults with age-related hearing loss).

Specific Aim 1: Investigate the contribution of auditory inputs to balance control in healthy young adults.

Recent epidemiological research has brought attention to the notion that other sensory impairments, such as hearing loss, may affect balance deficits and lead to an increased risk for falls in older adults (Li et al., 2013; Viljanen, Kaprio, Pyykkö, et al., 2009). Therefore, we intend to determine if hearing loss alone – without additional sensory impairments – causes balance deficits in individuals. Because extensive literature suggests hearing impairment impacts cognitive processing during challenging tasks (Larsby et al., 2008; McCoy et al., 2005), we will require subjects to perform a cognitive task and observe changes in reactive balance.

Twenty healthy, young adults with normal hearing between the ages of 21-35 will perform the dual-task experimental protocol (Figure 2). This dual-task experimental protocol will be performed with headphones under a normal hearing and simulated hearing loss condition. A pair of wireless Bose® QuietComfort 35 wireless headphones will be provided for the subjects to minimize environmental sounds. Hearing loss will be simulated using the Adobe Audition (Adams, Gordon-Hickey, Morlas, & Moore, 2012). The BKB-SIN will be played through the headphones under simulated hearing loss conditions used in audiology research (Adams & Moore, 2009). Surface translations creating a loss of balance will be provided as well to simulate an unexpected event. The order of the tasks with and without headphones will be randomized to prevent bias.

Specific Aim 2: Determine the effect of age-related hearing loss on balance in older adults.

Evidence suggests older adults with hearing loss are particularly vulnerable to less social engagement, less ability to perform Activities of Daily Living (ADL), poor physical functioning, balance difficulty, and increased risk for falls (Chia et al., 2007; Kiely et al., 2013; Kulmala et al., 2009; Pryce & Gooberman-Hill, 2012; Viljanen, Kaprio, Pyykko, et al., 2009; Wojszel & Bień, 2004). Evidence also suggests that older adults with hearing loss have poor cognitive processing during challenging auditory tasks compared to older adults with normal hearing (McCoy et al., 2005; Tun et al., 2010). We, therefore, will investigate if older adults with normal hearing and simulated hearing loss have poor balance compared to older adults with normal hearing while performing a cognitively challenging dual-task. Twenty older adults (65+) with normal hearing and with simulated age-related hearing loss will perform the dual-task while experiencing unexpected surface translations (Figure 2). The auditory test will be played through the headphones. The results of the 20 older adults with normal hearing will be compared to the 20 healthy young adults.

Specific Aim 3: Determine the effect of hearing aids to improve balance in older adults with age-related hearing loss.

Older adults with hearing loss have been known to experience depression, social isolation, loss of ADLs, poor balance, and increased risk of falling (Chia et al., 2007; Kiely et al., 2013; Pryce & Gooberman-Hill, 2012; Viljanen, Kaprio, Pyykko, et al., 2009; Wojszel & Bień, 2004). Hearing aids do not completely restore speech understanding, particularly in a noisy environment, but literature suggests hearing aids may still improve communication, decrease depressive symptoms, and may allow listening tasks to be less cognitively taxing; thereby preventing mental fatigue, information overload, and cognitive decline (Arlinger, 2003; Lupsakko et al., 2005; Young Choi, Shim, Lee, Yoon, & Joo, 2011). Furthermore, literature suggests hearing aids may even improve physical functioning and increase independence performing ADLs (Hogan et al., 2009; Lupsakko et al., 2005). Although this evidence is insightful, the minimal evidence has tested whether a decrease in cognitive load is the reason why a hearing aid improves physical functioning (F. R. Lin et al., 2013; Rumalla et al., 2015). We intend to test whether a cognitive mechanism explains why hearing aids improve balance for older adults with hearing loss. We theorize the older adult with hearing loss who wears a hearing aid does not need to attend as closely to unheard speech or sounds and can dedicate more resources to balance control.

Twenty older adults (65+) with age-related mild (20-40 dB HL) or moderate hearing loss (41-55 dB HL) and with hearing aids will perform the dual-task and receive perturbations (Figure 2). The auditory test will be played through the speakers instead of the headphones. The main speaker positioned top center on the screen of the V-Gait system played the sentences while the other speakers positioned laterally and backwards relative to the person on the treadmill, delivered the surround noise. The dual-task protocol will be performed both with and without the subject's hearing aids, in addition to the unexpected surface translations. The order of the task will be randomized to prevent bias.

We hypothesize older adults with age-related hearing loss will show improved balance while wearing hearing aids versus not wearing hearing aids.

The long-term goal of this research is to improve balance and reduce falls in older adults. This innovative research will improve preventative care and interventions for older adults with both hearing loss and balance impairments, will save millions of dollars to cover the cost of medically treating injurious falls, and will open the door for future research to improve understanding about the mechanisms behind how and why hearing loss is associated with balance deficits.

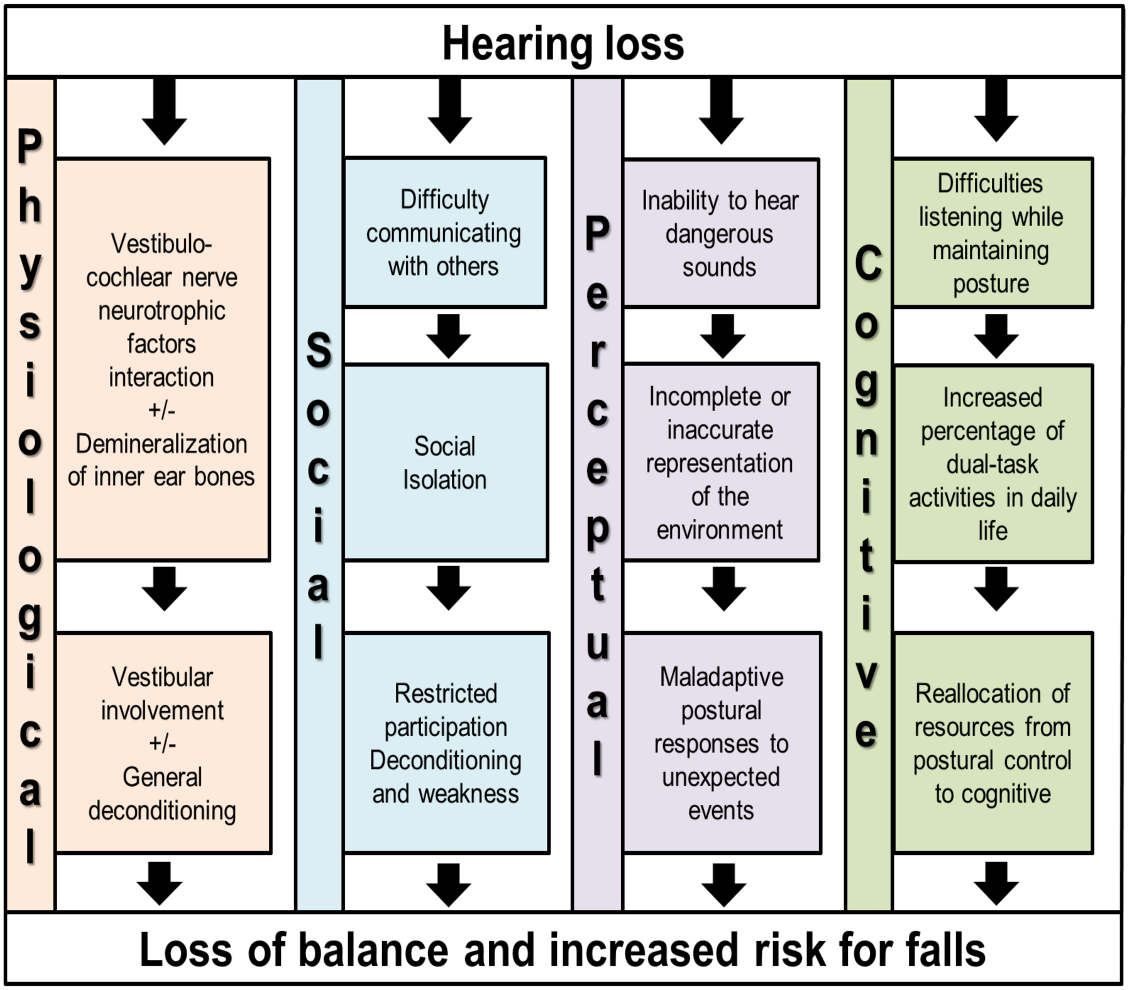

Figure 1. The figure displays the four potential mechanisms hypothesizing why hearing loss negatively affects balance – Physiological, Social, Perceptual, and Cognitive – and provides succinct rational behind the mechanism.

Dual-Task: Auditory + Balance/Mobility

Standing balance perturbations
+ Auditory test

- Balanced test design conditions:
  - Headphones: Yes/No
  - Hearing Aids: Yes/No
- Outcome measures:
  - Max COP-COM distance
  - Reaction Time
  - Number of Steps
  - Performance on auditory test

BKB-SIN

List 9A	Key Words	# Correct	SNR
1. The football player lost a shoe.	4	__________	+21 dB
2. The painter used a brush.	3	__________	+18 dB
3. The lady sat on her chair.	3	__________	+15 dB
4. The milkman brought the cream.	3	__________	+12 dB
5. The dog chased the cat.	3	__________	+9 dB
6. Mother shut the window.	3	__________	+6 dB
7. The apple pie was good.	3	__________	+3 dB
8. Rain falls from the clouds.	3	__________	0 dB
	Total Key Words Correct	__________	
	SNR–50 = (23.5) - (# Correct) =	__________dB	

Figure 2. The figure (top left) illustrates the dual-task performed in the human performance laboratory. Testing involves maintaining standing balance and responding to unexpected surface translation perturbations that require compensatory steps, while simultaneously listening and responding to a standardized audiology test, Bamford-Kowal-Bench Speech-In-Noise (BKB-SIN). The text (top right) describes the balance test conditions of three perturbation levels at 0, 2, and 5m/s^2; Hearing condition was either with headphones playing normal or simulated hearing loss sentences, or with (yes) or without (no) hearing aids; primary outcome measures included: maximum Center of Pressure – Center of Mass (COP-COM) distance, reaction time, number of steps, and performance of the BKB-SIN. The outcome measure (bottom) is an example of one list from the BKB-SIN. These sentences are played in the headphones or surround system and each sentence has a Signal-To-Noise Ratio (SNR), where the voice (signal) is accompanied by varying levels of multi-talker babble (noise). The difficulty increases as the SNR decreases, and subjects are graded on how accurately they are able to correctly repeat back the sentences.

Table 1. Epidemiological Evidence linking hearing loss to balance deficits and increased risk for falls in older adults.

Epidemiological Evidence					
Authors & Years	**Sample**	**Study Design**	**Hearing Assessment**	**Outcome Measures**	**Results**
Anstey, KJ, *et al.* (2001)	n= 1,311 Age Inclusion: 70+ y.o.	Retrospective	Audiometer	• Self-rated health • Death records	• HL not associated w/ increased risk for mortality • ↑ Risk of 2-year decline w/ HL
Çakmur, H. (2015)	n=168 Age Inclusion: 65+ y.o.	Experimental	Whisper test	Fried Frailty Criteria (FFC)	Hearing loss associated w/ ↑ frailty
Criter, RE & Honaker, JA. (2013)	n = 88 Age Inclusion: 60+	Retrospective	Patients in an audiology clinic	• ABC Scale • Generic fall questionnaire	• ABC scale scores: 23.8% at risk of falling • Women were more likely: o Afraid of falling o To fall o To be injured from falling
Feeny, D, *et al.* (2012)	n = 12,375 Age Inclusion: 18+	Retrospective	Self-report or hearing aid use	Health Utilities Index Mark 3 (HUI3)	• Hearing loss associated w/ ↑ HUI3 score = ↑ mortality risk • Hearing loss more likely: males, < high school degree, chronic health conditions, smoker, & physically inactive
Grue, EV, *et al.* (2009)	n = 770 Age Inclusion: 75+	Retrospective	Hearing loss assessment: patient needs quiet room to hear speech	InterRAI questionnaire	• Hearing impairment associated w/: o ↑ risk of falling • Dual-sensory impairment associated w/: o No greater risk of falling
Kamil, RJ, *et al.* (2016)	n=3,075 Age Range: 70-79 y.o.	Prospective	Pure-tone audiometry	• Frailty (gait speed of less than 0.60 m/s) • Self-report # of falls	• Moderate/severe HL ↑ risk of frailty and falls • Per 10 dB HL, ↑ risk of frailty and falls • Hearing aid not associated w/ dec frailty risk or fall risk
Kulmala, J, *et al.* (2009)	n = 428 (women) Age Range: 63-76	Retrospective	Audiometric assessment	• Balance force platform measurement system • Mailed 12-month follow-up for falls • MMSE • Medical examination	• Dual-sensory impairment: ↑ risk of falling • Risk of falling w/ dual-sensory impairment & balance impairment > Risk of falling w/ dual-sensory impairment alone
Lacerda, CF, *et al.* (2012)	n = 56 Age Range: 60-84	Experimental	Audiometric assessment	• BBS • Falls Efficacy International Scale-International (FES-I)	Post-hearing aid fitting: • ↑ BBS scores • No change in FES-I scores

Epidemiological Evidence (continued)					
Li, L, *et al.* (2013)	n = 1,180 (women) Age Range: 50-69	Retrospective	Audiometric assessment	20-ft walk test	• Hearing loss: Gait speed < 1.0 m/s → Increased risk of falling • Hearing aids didn't improve gait speed
Lopez, D, *et al.* (2011)	n = 5,354 Age Range: 76-81	Retrospective	Self-report	• Australian Department of Veterans' Affairs' Preventive Care Trial • SF-36	• Hearing impairment associated w/ ↑ risk of falling • Hearing impairment not associated w/ injury from a fall or mortality
Rumalla, K, *et al.* (2015)	n = 14 mean age: 77 y.o.	Experimental	Audiologic examination	• Activities specific Balance Confidence Scale (ABC) • Romberg test on foam; tandem stance test	• HA: ↑ Romberg & tandem score • No correlation w/ balance tests & ABC test • No correlation w/ HA gain & balance test scores
Sihvonen, S, *et al.* (2004)	n = 40 Age Range: 50-68	Retrospective	Self-report	• Medical records • Generic interview	• Ages 50-58: 22% of fallers = hearing impairment vs. 2% of non-fallers = hearing impairment • Ages 59-68: No difference between fallers w/ hearing impairment & non-fallers w/ hearing impairment
Skalska, A, *et al.* (2013)	n = 4,920 Average Age: 79.4	Retrospective	Hearing loss: only loud speech heard, Deaf:no noise heard	Generic questionnaire about fall-history	Subjects w/ hearing loss fell more often
Stephens, D, & Kerr, P. (2003)	n = 174 Average Age: 66.2	Cross-sectional	Patients in an audiology clinic	• Qualitative questionnaire • Quantitative questionnaire	• 20% = + comment in the qualitative questionnaire • 47.3% = + comment on each quantitative item • Younger adults with long-term hearing impairment = ↑ chance to report positive outcomes
Viljanen, A, *et al.* (2009, a)	n = 828 Age Range: 63-76	Prospective	Audiometric assessment	• Max walking speed for 10m • 6MWT • Self-reported walking difficulties walking 2km	• Hearing impairment associated w/: o ↓ max walking speed and walking endurance o ↑ self-reported major difficulties walking 2 km o 2x ↑ risk or major difficulties walking 2 km • Walking difficulties 12 months later: o 12.5% hearing impaired vs. 6.0% w/o hearing loss
Viljanen, A, *et al.* (2009, b)	n = 429 Age Range: 63-76	Retrospective	Audiometric assessment	• Postural sway based on COP movement • Self-report of falls • Genetic analysis	• Hearing loss associated w/: o ↑ average COP velocity movement & ↑ fall risk & rates • Hearing loss not associated w/ genetics
Weaver, TS, et al. (2017)	n=13 Mean Age: 83 y.o.	Experimental	Pure-tone audiometry	• Gait parameters • TUG	Hearing aid: No improvement w/ gait parameters or TUG
Wojszel, ZB & Bien, B. (2004)	n = 457 Ages: 75+	Cross-sectional	Self-report	Interview	Hearing impairment associated w/ ↑ # of sporadic & frequent falls

Table 2. Literature support for the physiological mechanism explaining the relationship between hearing loss and balance deficits.

Physiological Theory					
Authors & Years	**Sample**	**Study Design**	**Hearing Assessment**	**Outcome Measures**	**Results**
Agrawal, Y, *et al.* (2008)	n = 5,742 Age Inclusion: 20-69 y.o.	Retrospective	Audiometric assessment	Demographics	Hearing loss associated w/ hypertension and diabetes mellitus
Kakarlapudi, V, *et al.* (2003)	n = 74,097 Age Inclusion: N/A	Retrospective	Audiometric assessment	Blood Chemistry	Hearing loss associated w/ increased creatinine levels in diabetic patients
Kim, JY, *et al.* (2016)	n=324 Mean Age=62 y.o.	Retrospective	Pure-tone audiometry	• DXA scan • Bloodwork	• ↑ Hearing loss w/ osteoporosis/osteopenia • Lumbar BMD associated w/ HL
Mendy, A, *et al.* (2014)	n = 8,863 Age Inclusion: 40+	Retrospective	Self-report	• Romberg Test • Dexta Scan	Low BMD associated with: • ↓ Scores on Romberg Test • ↓ Hearing ability
Oh, I-H, *et al.* (2014)	n = 37,773 Mean Age: 44 y.o.	Cross-Sectional	Audiometric assessment	• Blood Pressure • Blood Chemistry	Hearing loss associated w/ diabetes mellitus
Purchase-Helzner, EL, *et al.* (2004)	n = 6,480 (women) Age Inclusion: 65+	Retrospective	Audiometric assessment	• Dexta Scan • Self-report for falls • Patient records for fractures	Hearing loss not associated w/ BMD, falls, or fracture rate
Zamyslowska-Szmytke, E, *et al.* (2011)	n = 170 Mean Age: 38 y.o. (experimental), 37 y.o. (control)	Experimental	Audiometric assessment	• Solvent Exposure • Noise Exposure • Modified Clinical Test of Sensory Integration on Balance (mCTSIB) • Limits of Stability (LOS) • Electronystagmography (ENG)	• No difference mCTSIB scores w/ solvent-exposed vs control group • Solvent exposed group: ↑ HL prevalence, ↑ reaction time on LOS, ↑ nystagmus on ENG

Table 3. Literature support for the social mechanism explaining the relationship between hearing loss and balance deficits.

Social Theory					
Authors & Years	**Sample**	**Study Design**	**Hearing Assessment**	**Outcome Measures**	**Results**
Bazargan, M, *et al.* (2001)	n = 998 Age: 62-99	Cross-sectional	Self-Report Hearing Loss	• The Philadelphia Geriatric Center (PGC) Morale Scale • Inventory of Social Supportive Behaviors • Revised Holmes and Rahe scale • Self-Report	• Hearing impairment associated w/ well-being, education, social support, self-rated health status, & daily activities • Only 4% reporting hearing impairment wore HA
Brink, MA & Stones, M. (2007)	n = 12,254 Average Age = 80	Retrospective	Self-report	• Activities of Daily Living—Short Form • Activity Pursuit Patterns Index • Questionnaire: communication & social life	Hearing impairment: • ↓ mood & communication • Inconsistent effect on social engagement & ADLs
Chen, DS, *et al.* (2015)	n = 2,190 Age Range: 70-79 y.o.	Retrospective	Audiologic examination	• Short Physical Performance Battery(SPPB) • Gait speed • Self-report disability or nursing care	• Hearing loss associated w/: o ↓ SPPB, ↓ gait speed, ↑ disability or nursing care
Chia, E, *et al.* (2007)	n = 2,956 Ages: >49	Cross-sectional	Pure tone audiometry	SF-36	• Hearing Impairment: ↓ SF-36 PCS score. • Hearing aids did not improve SF-36 PCS score.
Dawes, P, *et al.* (2015)	n=164,770 Age Range: 40-69 y.o.	Experimental	• Digit Triplet Test (DTT) • Self-report	Self-report of general health, disability, social isolation, depression	• Hearing loss associated w/: o ↑ social isolation & depression • Hearing aids associated w/ ↑ social isolation
Grue, EV, *et al.* (2009)	n = 770 Ages: 75+	Retrospective	Hearing loss: quiet room to hear speech	InterRAI questionnaire	Hearing impairment & dual-sensory impairment associated w/ ↑ risk of IADL loss
Heyl, V, and Wahl, HW. (2012)	n = 430 Age Range: 75-94	Experimental	• Audiometric assessment • Self-report	• 10 items from classic ADL-IADL scales • Environmental Mastery Scale (EMS) from Ryff Scales of Psychological Well-Being (PWB)	Hearing Impairment: • ↓ADL difficulty & IADL difficulty • ↑ Independence • No difference in Environmental Mastery Scale of PWB
Hogan, A, *et al.* (2009)	n = 43,233 Ages: 55+	Retrospective	Self-report	SF-12	• Hearing impairment: ↓ PCS SF-12 scores • Hearing aids: ↑ PCS SF-12 scores
Hung, WW, *et al.* (2012)	n = 17,723 Ages: 65+	Retrospective	Self-report	• Self-report disability • Self-report ADL/IADL difficulty	Hearing loss not associated w/ disability in mobility or IADL/ADL tasks.
Loprinzi, PD, *et al.* (2013)	n = 1,145 Age NA	Retrospective	Audiometric assessment	Accelerometer	Subjects w/ hearing loss had similar physical activity to subjects w/ normal hearing

Social Theory (continued)					
Jagger, C, *et al.* (2005)	n = 1,579 Ages: 75+	Retrospective	Self-report	Activities of Daily Living (ADLs)	Hearing loss associated w/ ↑ risk of activity restriction
Kiely, KM, *et al.* (2013)	n = 1,611 Age Range: 65-103	Retrospective	Audiometric Assessment	• Centre for Epidemiological Studies Depression scale (CES-D) • Self-report difficulties w/ ADLs and IADLs • Adelaide activity profile (AAP)	Hearing loss & Dual-Sensory Loss: • ↑ depressive symptoms • ↑ ADL & IADL difficulty • No effect on social engagement
Lacerda, CF, *et al.* (2012)	n = 56 Age Range: 60-84	Experimental	Audiometric assessment	• SF-36 • Satisfaction with Amplification in Daily Life (SADL)	Post-hearing aid fitting: • ↑ SF-36 scores • ↑ SADL scores
Lin, MY, *et al.* (2004)	n = 6,112 (women) Ages: 69+	Retrospective	Audiometric assessment	National Health Interview Survey Supplement on Aging	Hearing Impairment & Dual-Sensory Impairment: ↑ risk of functional decline
Lopez, D, *et al.* (2011)	n = 5,354 Age Range: 76-81	Retrospective	Self-report	• Australian Department of Veterans' Affairs' Preventive Care Trial • SF-36	Hearing impairment associated w/ ↓ PCS SF-36 score
Lopez-Torres Hidalgo, J, *et al.* (2009)	n = 1,387 Age Inclusion: 65+	Cross-sectional	Self-report	• Hearing Handicap Inventory for the Elderly-Screening (HHIE-S) • Katz Index • Geriatric Depression Scale (GDS)	• Hearing loss associated w/: o >3 health conditions, ADL dependence, & depression • HA use associated w/ risk of ADL dependence
Lupsakko, TA, *et al.* (2005)	n = 601 Ages: 75+	Cross-sectional	Self-report hearing aid use	• Barthel index • Depression Scale Index (DSI)	Hearing aid users: • ↑ median income & ADL score • No difference w/ depression score
Mikkola, TM, *et al.* (2015)	n = 847 Age Range: 75-90 y.o.	Retrospective	Self-report	MMSE	Hearing loss not associated w/ MMSE scores
Pryce, H, and Gooberman-Hill, R. (2012)	n = 18 Age Range: 76-99	Cross-sectional	Self-report	• Observation • Interviews	• Individuals with hearing impairment: ↑ difficulty communicating, which led to ↓ social interaction • Hearing aids didn't ↑ social activities
Tomioka, K, *et al.* (2015)	n = 3,982 Age Range: 65-96 y.o.	Cross-sectional	• HHIE-S • Self-reported hearing loss	• 10-meter walk • Single-Leg Stace (SLS) • Geriatric Depression Scale (GDS) • International Physical Activity Questionnaire	• Hearing loss associated w/: o ↓ IADL and QOL scores o ↑ depression o ↓walking speed and standing balance.

Table 4. Literature support for the perceptual mechanism explaining the relationship between hearing loss and balance deficits.

Perceptual Theory					
Authors & Years	**Sample**	**Study Design**	**Hearing Assessment**	**Outcome Measures**	**Results**
Girard, SA, *et al.* (2014)	n = 8,728 (men) Age Range: 55-65	Retrospective	• Audiometric assessment • Number of years in a noisy workplace • Noise exposure level	Medical record: • Fall history • Injuries from a fall • Death after hospital admittance from a fall	Hearing loss and hospital admittance from a fall: • 93% injured • 10.7% admitted for a 2nd fall • 8% died after hospital admittance
Kanegaonkar, RG, *et al.* (2012)	n=20 Age Range: 23-44 y.o.	Experimental	History of hearing loss	• Clinical Test for Sensory Interaction in Balance (CTSIB): o Normal clinic room vs. Soundproof room	Only EO condition significant difference normal clinic vs. soundproof room
Lau, ST, *et al.* (2016)	n=16 Mean Age: OANH=69.9, OAHL=73.3	Experimental	pure-tone audiometry	• Single/Dual-task: o Walk o Word-recognition • Spatial expectation	• HL: overall ↓spatial expectation • No benefit of likely vs. unlikely spatial expectation for both NH and HL
Negahban, H, *et al.* (2017)	n=47 Mean Age: 67 y.o.	Experimental	pure-tone audiometry	• Modified Clinical Test of Sensory Interaction in Balance (CTSIB-M) • Center of Pressure (COP) sway	• ↑ COP sway of HA-off vs HA-on • No differences HA-off vs unaided w/ hearing loss • Benefit of HA: + correlation between time of HA acquisition and AP sway
Palmer, KT, *et al.* (2015)	n =5,915 men Mean Age: 39.9 y.o.	Retrospective	medical records	Occupational injury	• Ear problems significantly associated w/ injury consultation • Injury risk not associated w/ hearing loss
Vitkovic, J, *et al.* (2016)	n=97 Age Range: 21-80 y.o.	Experimental	pure-tone audiometry	• Modified Clinical Test of Sensory Interaction in Balance (CTSIB-M) • Center of Pressure (COP) sway	• Normal hearing: ↑ sway w/ sound • Hearing loss: no difference in sway w/ sound • Hearing aid: improved sway regardless of sound

Table 5. Literature support for the cognitive mechanism explaining the relationship between hearing loss and balance deficits.

Cognitive Theory					
Authors & Years	**Sample**	**Study Design**	**Hearing Assessment**	**Outcome Measures**	**Results**
Akyiğit, A, *et al.* (2014)	n = 338 (control: 214; dementia: 124) Mean Age: 82.9 y.o.	Experimental	Audiologic examination	• DSM-IV-TR • MMSE	Hearing loss associated w/ Alzheimer's dementia
Bazargan, M, *et al.* (2001)	n = 998 Age Range: 62-99	Cross-sectional	Self-Report	MMSE	Hearing impairment associated w/ cognitive deficits
Brink, MA & Stones, M. (2007)	n = 12,254 Age Inclusion: 65+	Retrospective	Self-report	Lawton Cognition Scale	Inconsistent effect of hearing impairment on cognition
Bruce, H, *et al.* (2017)	n=87 Mean Age: YA=21 y.o., OANH=65 y.o., OAHL=70 y.o.	Experimental	Pure-tone audiometry	• Single/Dual-task: o 1-back o Ankle plantarflexion amplitude or Hip extension amplitude	• Hearing loss: ↑ dual-task cost • No difference in ankle/hip amplitude between OANH or OAHL
Chia, E, *et al.* (2007)	n = 2,956 Age Inclusion: 49+	Cross-sectional	Audiometric assessment	SF-36	• Hearing Impairment: ↓ SF-36 MCS score • Hearing aids didn't improve SF-36 MCS score
Davis, A. (2003)	n = 2,466; 506; 351; 88 Age Range: 55-74	• Cross-sectional • Experimental	• Self-report • Audiometric assessment	• Four Alternate Auditory Feature (FAAF) test • Reading task with recall • Glasgow Hearing Aid Benefit Profile • Abbreviated Profile of Hearing Aid Benefit (APHAB) • Auditory Lifestyle and Demands (ALDQ)	Hearing loss & poor cognitive abilities: • ↓ FAAF score in unaided hearing tests • ↓ ALDQ scores • ↑ Hearing aid benefit
Dawes, P, *et al.* (2015)	n=164,770 Age range = 40-69 y.o.	Experimental	Digit Triplet Test (DTT); self-report	• Cognitive testing: o Reaction Time o Pairs matching o Fluid Intelligence	• Hearing loss associated w/ ↓ cognition • Hearing aids associated w/ ↑ cognitive performance

Cognitive Theory (continued)					
Authors & Years	**Sample**	**Study Design**	**Hearing Assessment**	**Outcome Measures**	**Results**
Dupuis, KM, *et al.* (2015)	n=301 Avg Age: 71	Experimental	• Audiologic examination • Self-report	Montreal Cognitive Assessment	Hearing loss associated w/ ↓ MCA scores
Edwards, JD, *et al.* (2017)	n=500 Age Range: 63–90 y.o.	Prospective	Pure-tone audiometry	• Self-report driving space, challenges, and difficulty • Useful Field of View (UFOV)	• Moderate/severe hearing loss: ↓ UFOV scores • Hearing loss not associated w/ driving mobility or driving cessation
Gatehouse, S, *et al.* (2003)	n = 50 Age: N/A	Experimental	Hearing aid users for at least 6 mo.	FAAF test	↓ FAAF scores
Glyde, H, *et al.* (2012)	n = 80 Ages: 7-89	Experimental	• Audiometric assessment • Real-ear insertion gain (REIG) for hearing aid users	• Listening in Spatialized Noise-Sentences test (LiSN-S) • COGNISTAT • The Speech, Spatial and Qualities of Hearing Scale (SSQ)	Hearing impairment associated w/: • ↓ LiSN-S scores • ↓ COGNISTAT scores • ↓ SSQ scores
Grue, EV, *et al.* (2009)	n = 770 Ages: 75+	Retrospective	Hearing loss: quiet room to hear speech	Cognitive Performance Scale (CPS)	• HL not associated w/ ↑ cognitive deficits risk • Dual-sensory impairment not associated w/ ↑ risk of cognitive deficits
Gurgel, RK, *et al.* (2014)	n = 4,463 Ages : 65+	Prospective	Audio amplifier during testing	• Dementia • Modified Mini-Mental Status Exam (3MS-R)	• Hearing loss associated w/: ↑ risk of dementia, ↑mental decline, ↓ 3MS-R score
Gussekloo, J, *et al.* (2005)	n=459 Age Inclusion: 85+ y.o.	Retrospective	Audiometer	• MMSE • 12-Word Learning Test • Letter Digit Coding Test • Stroop Test	• Hearing loss associated w/ ↓ MMSE scores • No association w/ HL & scores on memory and cognitive speed scores
Hallgren, M, *et al.* (2005)	n = 24 Age Range: 25-45 & 65-80	Experimental	Hearing aid users	• Hagerman Speech Test • Speech and Visual Information Processing System (SVIPS) tests • Self-report perceived listening effort	• Perceived effort: o Silence < Hagerman's noise < Speech. o ↑ w/o hearing aids • Hearing aids: o ↑ performance on Hagerman Speech Test o No benefit during SVIP tests o Benefit dependent on background noise
Heyl, V, and Wahl, HW. (2012)	n = 430 Age Range: 75-94 y.o.	Experimental	• Audiometric assessment • Self-report	• Counting Backwards • Animal-Naming • Revised version of Wechsler Adult Intelligence Scale (WAIS-R)	Cognitive functioning tests: No difference between hearing impaired group, visually impaired group, dual-impaired group, & unimpaired group.
Hogan, A, *et al.* (2009)	n = 43,233 Ages: 55+	Retrospective	Self-report	SF-12	• Hearing impairment: ↓ MCS SF-12 scores • Hearing aids: ↑ MCS SF-12 scores

Cognitive Theory (continued)

Authors & Years	Sample	Study Design	Hearing Assessment	Outcome Measures	Results
Harrison Bush, AL, *et al*. (2015)	n=894 Mean Age: 73.4	Retrospective	Audiometer	• Digit Symbol Substitution/Copy • Trail Making Test Part A/B • Letter and Pattern Comparison • Stroop Test • Digit and Spatial Span Test • HVLT	Hearing loss associated w/ ↓ cognitive abilities
Humes, LE. (2002)	n = 171 Age Range: 60-89	Experimental	Audiometric assessment	• CUNY Nonsense Syllable Test (NST) • Connected Speech Test (CST) • Speech Intelligibility Index (SII)	• Hearing aids during quiet & speech: o ↑ CUNY NST performance o ↑ CST performance • Hearing Loss = strongest predictor variable for SII performance
Koh, DH, *et al*. (2015)	n=46 Age Inclusion: 65+ y.o.	Experimental	Audiometer	• K-MMSE • TUG test • One-Leg Stance Test (OLST)	• Hearing loss associated w/ ↓ OLST time • Hearing loss not associated w/ TUG or K-MMSE
Larsby, B, *et al*. (2005)	n = 48 Age Range: 22-38 & 66-75	Experimental	Audiometric assessment	• Speech and Visual Information Processing System (SVIPS) tests • Borg's CR-10 scale	• Hearing impairment: ↑ reaction time during SVIPS • Hearing impairment and background noise: ↑ Borg CR-10 score
Larsby, B, *et al*. (2008)	n = 24 Age Range: 56-83	Experimental	• Audiometric assessment • HA fitting at least one year prior	• The Hagerman's test • Speech and Visual Information Processing System (SVIPS) • Self-rating: listening effort	• Hagerman's test & background noise: ↓ performance & ↑ perceived listening effort • SVIPS & background noise: ↓ performance & ↑ perceived listening effort
Lau, ST, *et al*. (2016)	n=16 Mean Age: OANH=69.9 OAHL =73.3	Experimental	Pure-tone audiometry	• Single/Dual-task: o Walk o Word-recognition • Spatial expectation	• HL and dual-task: o No difference word recognition scores o ↑ trunk/head pitch variability
Lin, FR, *et al*. (2011)	n = 347 Age Inclusion: 55+	Retrospective	Audiometric assessment	• MMSE • Free and Cued Selective Reminding Test (FCSRT) • Trail-Making A and B; Stroop • Letter and Category Fluency • American version of the National Adult Reading Test (AMNART) • Center for Epidemiologic Studies Depression (CES-D)	• Hearing impairment associated w/: o ↓ MMSE scores o ↓ FCSRT scores o ↓ Stroop test scores • Hearing impairment not associated w/: o Trail-making A & B scores o Letter & Category Fluency o AMNART scores o CES-D scores

Cognitive Theory (continued)					
Authors & Years	**Sample**	**Study Design**	**Hearing Assessment**	**Outcome Measures**	**Results**
Lin, FR, *et al.* (2013)	n = 1,984 Age Range: 70-79	Retrospective	Audiological testing	• Modified Mini-Mental State Examination (3MS) • Digit Symbol Substitution (DSS) • 20-item Centre for Epidemiological Studies Depression Scale • Dementia medication	• Hearing loss associated w/: o ↓ 3MS scores & ↓DDS test scores o No difference in Depression scores • Hearing aids associated w/: o ↓ 3 MS scores o No difference w/ DDS test scores • Hearing aids don't ↓ risk or rate of cognitive decline
Lin, MY, *et al.* (2004)	n = 6,112 (women) Age Inclusion: 69+	Retrospective	Audiometric assessment	MMSE	Hearing Impairment & Dual-Sensory Impairment: ↑ risk of functional decline
Lopez, D, *et al.* (2011)	n = 5,354 Age Range: 76-81	Retrospective	Self-report	• Australian Dept of Veterans' Affairs' Preventive Care Trial • SF-36	Hearing impairment associated w/ ↓ MCS SF-36 score
Lopez-Torres Hidalgo, J, *et al.* (2009)	n = 1,387 Ages: 65+	Cross-sectional	Self-report	Short Portable Mental Status Questionnaire	Hearing loss associated w/ cognitive impairment
Lunner, T. (2003)	n = 72; 17 Age Range: 33-89; 48-78	Experimental	Subjects w/ hearing loss & no prior hearing aid use	• Hagerman's Sentence Test • Reading Span Test • Rhyme judgment test	High cognitive functioning: ↑ performance on Hagerman's sentence test, reading span test, and rhyme judgment test both w/ & w/o hearing aids
Lupsakko, TA, *et al.* (2005)	n = 601 Ages: 75+	Cross-sectional	Self-report hearing aid use	MMSE	Hearing aid users: ↑ MMSE score
McCoy, SL, *et al.* (2005)	n = 24 Age Range: 66-81	Experimental	Audiometric assessment	Testing method taken from Miller and Selfridge (1950)	Hearing loss: • ↓ Recall of 1st & 2nd word in a 3-word sentence • No difference in 3rd word recall in 3-word sentence
Mikkola, TM, *et al.* (2015)	n = 847 Age Range: 75-90 y.o.	Retrospective	Self-report	• Self-report social and leisure activities • Impact on Participation and Autonomy (IPA) questionnaire • Self-report walking difficulty	Hearing loss associated w/ o ↑ difficulty w/ mobility o ↑ social isolation o ↓ participation in group activities
Moore, DR, *et al.* (2014)	n= 502,642 Mean Age: 56.5 y.o.	Retrospective	• Self-report • Digit Triplets Test (DTT)	• Fluid Intelligence • Prospective Memory • Visual Memory • Digit Span	↑ DTT scores associated w/ ↑ cognitive test scores.

Cognitive Theory (continued)					
Authors & Years	**Sample**	**Study Design**	**Hearing Assessment**	**Outcome Measures**	**Results**
Ng, EH, *et al.* (2013)	n = 26 Age Range: 32-65	Experimental	• Audiometric assessment • Hearing aid use 9.5+ hours daily	• Reading span (RS) test • Sentence-final Word Identification and Recall (SWIR)	An individual with hearing loss & high-working memory: • ↑ RS test performance • ↑ SWIR test performance
Pearman, A, *et al.* (2000)	n = 344 Age Range: 55-93	Experimental	• Audiometric assessment • Self-report	• MMSE • GDS • serial word recall test • generic health questionnaire	• Hearing impairment not associated w/ GDS or MMSE scores. • Hearing impairment associated w/ ↓ serial word recall test scores
Ronnberg, *et al.* (2011)	n = 2,756; 160 Average Age: 64; 75	Retrospective	Audiometric assessment	• Free recall of subject-performed tasks (SPTs) • Free recall of sentences test • Free word recall test • Vocabulary test • Fluency test	• Hearing loss negatively affected long-term memory: ↓ performance on the free recall of SPTs, free recall of sentences test, free-word recall test, a vocabulary test, & a fluency test. • Hearing loss had no effect on episodic or short-term memory, free recall of sentences test, & free-word recall test.
Rudner, M, *et al.* (2012)	n = 16; 30 Average Age: 63.5; 70	Experimental	• Audiometric assessment • Hearing aids fitted one year prior to study	• VAS for perceived listening effort • Letter monitoring task • Reading span test	Hearing loss and high working memory associated w/ ↓ VAS scores on the letter monitoring & reading span test
Shahidipour, Z, *et al.* (2013)	n = 47 (normal hearing = 24, mild to mod HL = 23) Age Range: 60-80 y.o.	Cross-sectional	Pure-tone and speech audiometry	•Dichotic Verbal Memory Test (DVMT) •MMSE	• Hearing Loss associated w/ ↓ DVMP • MMSE: no difference
Tomioka, K, *et al.* (2015)	n = 3,982 Age Inclusion: 65-96	Cross-sectional	HHIE-S; self-reported hearing loss	MMSE	Hearing loss associated w/ ↑ cognitive impairment
Tun, PA, *et al.* (2010)	n = 36 Age Range: 68-85	Experimental	Audiometric assessment	Reaction time while answering true/false Q's	Hearing impaired older adults = longer reaction time

Cognitive Theory (continued)					
Authors & Years	**Sample**	**Study Design**	**Hearing Assessment**	**Outcome Measures**	**Results**
van Hooren, SAH, *et al.* (2005)	n = 102 Age Inclusion: 60+	Experimental	Audiometric assessment	• Stroop Color-Word Test (SCWT) • Concept Shifting Task (CST) • Letter-Digit Substitution Test • Visual Verbal Learning Test • Verbal Fluency Test	One-year post-hearing aid fitting: ↑hearing ability , but no improvement on all cognitive tests
Wollesen, B, *et al.* (2017)	n=73 Mean Age: (NH=64, Mild HL=71, Mod/ Severe=78 y.o.)	Experimental	Pure-tone audiometry	•Single/Dual/Triple Task Conditions (STC/DTC/TTC): o Gait parameters o Holding cup w/ water o Stroop test	•Hearing loss associated w/: o Overall ↓ cadence, step length, and walking speed o ↓ step length DTC, but not TTC
Wu, Y-H, *et al.* (2016)	n=48 Mean Age: YANH=23.7 OAHL=69.9	Experimental	Pure-tone audiometry	•Dual-Task: o HINT (SNR) o Reaction Time (RT) easy/hard	•Hearing loss associated w/: o ↑ RT w/ hard task o No difference SNR scores easy vs hard task o Non-linear relationship between RT and SNR
Young Choi, A, *et al.* (2011)	n = 29 Average Age: 69.5 (experimental group) & 63.1 (control group)	Experimental	Audiometric assessment	• Visual Verbal Learning Test (VVLT) • Words-In-Noise (WIN) Test	Six-months post-hearing aid fitting: • ↑ VVLT scores • ↓ WIN test scores
Zekveld, AA, *et al.* (2007)	n = 30 Age Range: 24-72	Experimental	Audiometric assessment	• Cambridge Neuropsychological Test Automated Battery (CANTAB) • Search strategy for Spatial Working Memory	• Hearing loss not associated w/ CANTAB tests • Hearing loss associated w/ efficient search strategy for spatial working memory
Zekveld, AA, *et al.* (2011)	n = 74 Average Age: 61 (hearing impaired) & 55 (normal hearing)	Experimental	Audiometric assessment	• Hearing impairment associated w/: ↓ pupil adaptation in noise (cognitive overload), ↑ perceived listening effort, ↓ LDST scores • Hearing impairment not associated w/ ↓ TRT test scores • ↑ Word Voc test scores associated w/ ↑ LDST scores	• Pupillometry • Self-rating for perceived listening effort • Text Reception Threshold (TRT) test • Letter Digit Substitution Test (LDST) • Word Vocabulary (WordVoc) test

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CHAPTER 2

SIMULATED HEARING LOSS IN HEALTHY YOUNG ADULTS DOES NOT CHANGE REACTIVE BALANCE RESPONSES

Introduction

Epidemiological and emerging research evidence suggests a correlation between hearing loss and increased risk for falls, particularly in older adults. Individuals with hearing loss walk with slower gait speed, self-report poor physical mobility, fall more often compared to normal hearing individuals, and have increased COP sway measurements during quiet stance with background noise.[1-3] However, there is no clear evidence determining the effect of hearing impairment on balance. In a real-world setting, individuals with hearing loss are attending to auditory sounds, such as speech, while simultaneously attempting to stand, walk, or cross obstacles. It is plausible that when performing mundane postural tasks while attending to sounds, individuals with hearing loss are performing dual-tasks of and may have a higher risk of loss of balance and falling.

Incidences of hearing loss and balance problems are higher in older adults, and the consequences of loss of balance and falls have a major impact on this population. Studying the effect of hearing loss on balance control in older adults can be confounded and/or compounded by the multitude of age-related changes in the neuro-musckulosekeletal and cognitive systems that contribute to decreased balance control.[2,4] Therefore, we aimed to investigate the effect of hearing loss on the control of balance in healthy young subjects with intact neuro-musculoskeletal systems and simulated hearing loss. Various studies have simulated acute sudden hearing loss in both young and older adults with the goal to improve methods for audiological assessments of individuals with hearing loss, compare differences in sensorineural hearing loss to acute sudden hearing loss, and to improve hearing aid benefit.[5,6] However, no studies have investigated the effect of simulated hearing loss on the control of reactive balance. Research studies testing the control of reactive balance have a particular translational potential due to ability to create unexpected, similar to real-life loss of balance conditions in a controlled environment. A biomechanical outcome measure commonly used to analyze reactive balance is maximum Center of Pressure – Center of Mass (COP-COM) distance during compensatory steps.[7,8] Center of Pressure (COP) and Center of Mass (COM) interact closely with one another to maintain postural stability during both anticipatory stepping (ie. gait initiation) and compensatory stepping (ie. unexpected loss of balance).[9] The maximum COP-COM distance is considered an indicator of robustness in the balance control system.[10]

We created a novel auditory and balance dual-task paradigm to investigate the effect of hearing loss on balance control. To our knowledge, this is the first study to integrate a standardized audiology test with surface translation perturbations. Healthy, young adults with optimally functioning neuro-muscular and sensory systems performed an auditory task under normal hearing and simulated hearing loss conditions while simultaneously maintaining standing balance during unexpected surface translations. We hypothesized that sudden acute simulated hearing loss would negatively impact balance control manifested by a decreased maximum COP-COM distance during the first compensatory step, and increased reaction time for initiating the first compensatory step; particularly while performing the dual auditory-postural task compared to a no audio or normal auditory condition.

Methods

Subjects

Twenty-five young healthy young adults voluntarily agreed to participate in the study and provided written informed consent, approved by the institutional review board. All subjects were verbally screened prior to enrollment to ensure no auditory, vestibular, musculoskeletal, neurological, or cardiopulmonary conditions existed that would impair balance. Furthermore, subjects were excluded if they had a history of tinnitus, motion sickness/dizziness or were taking medications that affect balance.

Experimental Design

Subjects participated in two visits, consisting of: Visit 1) Memory and sensory screening, Visit 2) Balance testing: dual-task auditory and perturbation protocol.

Visit 1: *Memory and Sensory Screening*

Subjects participated in memory, auditory, visual, vestibular, somatosensory, and sensory integration screening to ensure no undiagnosed impairments existed. All screening procedures were performed by a licensed physical therapist. Subjects who passed memory and sensory screenings were invited back to participate in dual task auditory-balance testing. Subjects who did not pass all screening assessments were withdrawn by the principal investigator from the research study.

Visit 2: *Balance testing: Dual-Task auditory and perturbation protocol.*

Subjects were required to stand and maintain their balance following unexpected surface translations while simultaneously listening and repeating back sentences from the standardized audiology outcome measure, the Bamford-Kowal-Bench Speech-In-Noise (BKB-SIN) test, played through the headphones at 60 dBA.[11] These are simple sentences like, “The dog chased the cat.” A higher score on the BKB-SIN indicates a poorer performance. There were three auditory conditions: 1) no audio sound, no repeat back, resulting in the single task of maintaining balance; 2) normal hearing; and 3) simulated hearing loss.

Hearing loss was simulated using Adobe Audition. Five second clips of each sentence from the standardized audiology test, BKB-SIN, were uploaded into the program. The BKB-SIN consists of a target voice and multi-talker babble/noise. The voice and the multi-talker babble were separated from 1 track into 2 separate tracks with 1 track constituting the target voice and 1 track constituting the multi-talker babble. The decibel (dB) levels were manipulated at particular frequencies associated with moderate hearing loss. Moderate hearing loss values of decibel loss per frequency were obtained from The National Institute for Occupational Safety and Health (NIOSH) Hearing Loss Simulator.[12] Moderate hearing loss was simulated by applying a Fast Fourier Transform (FFT) filter (Logarithmic scale, FFT size: 2048, Blackman window) in Adobe Audition to the separated track of the voice, according to previous research simulating hearing loss.[5,6] (Figure 1)

Subsequently, the voice and babble/noise were recombined in one file that maintained the Speech-in-Noise (SIN) ratio associated with each sentence of the BKB-SIN test. A total of 3 lists each containing 8 short sentences were manipulated to simulate hearing loss; the other 3 lists of the BKB-SIN were used in their original state. No sentence was heard more than one time by each subject in the study. Subjects used Bose® QuietComfort 35 wireless headphones to listen to sentences and limit any additional environmental noise during testing.

A 12 camera Motion Analysis System capture system collected kinematic data from 54 reflective markers placed on anatomical landmarks of the body. The V-gait treadmill system (by Motek Medical) with two separate force plates mounted underneath each belt was used to deliver perturbations and record force data. Perturbations consisted of backward surface translations to simulate a real-word unexpected event that creates a loss of balance in the forward direction. Three levels of balance conditions were delivered at accelerations: Level “0” (0 m/s^2), with no backward surface translations, resulting in the single task of listening and repeating back the sentence; Level “1” backward surface translations at acceleration of 2m/s^2; and Level “2” backward surface translations at acceleration of 5m/s^2. Subjects were instructed to “do whatever is natural to you to keep your balance” and encouraged to guess every time they heard babble. When the treadmill accelerated backward, subjects experienced a forward loss of balance requiring 1 or more compensatory steps forward to maintain their balance. An overhead harness system equipped to support up to 181 kg was in place to preventing injury or a fall to the floor.

Subjects were initially introduced to each single task: 1) listening and repeating back sentences from the BKB-SIN test, in sitting with and without simulated hearing loss and, 2) responding to surface translations at all levels in standing. After introduction to the two tasks, subjects performed a randomized sequence of single- and dual-tasks with combinations of balance and auditory conditions.

Single- and dual-task auditory-balance conditions were randomized to control for a learning effect with the BKB-SIN and reactive balance ability.[13,14] Randomization occurred in a data excel file according to perturbation level, auditory condition, and Signal-To-Noise (SNR) level; resulting in a total of 64 trials per subject. Conditions were then divided into 4 sets to allow subjects to have a mental and physical break between bouts of tasks (Table 1).

Outcome measures

Primary outcome measures were: performance on the BKB-SIN, the maximum COP-COM distance during the first compensatory step, and the reaction time for initiating the first compensatory step.

Maximal distance between COP and COM has been documented in the literature as a stability measure.[8] In gait initiation, the dissociation of COP from COM (COP-COM) is a requirement for the first step and is considered a measure of robustness of balance control system; the larger the COP-COM distance, the more robust the balance control is considered to be.[10] Therefore, COP-COM was chosen as an appropriate measure of postural stability.

Data Processing

All kinematic and force data were processed using MATLAB. COP was measured using ground reaction forces from the force plate data and COM was extrapolated using the sacral marker as described in Yang and Pai, 2014 [15]. Baseline values for COP-COM were calculated in the 30 frames before the surface translation (0.25 seconds). The maximum COP-COM distance during the first compensatory step was calculated within the window of time from surface perturbation to completion of the first compensatory step, indicated by placement of the stepping leg heel on the force plate. Both heel marker data and force data were used to confirm initiation and completion of reactive step. All maximum COP-COM distances were normalized to subject height. Reaction time was determined as the time in milliseconds (ms) from start of surface perturbation to the point in time when the heel marker of the stepping foot transitioned from accelerating backward while receiving the surface translation to accelerating forward in the opposite direction while responding with a compensatory step. Calculation of reaction time was assessed within the window from surface perturbation initiation to maximum acceleration of the heel marker during first compensatory forward step.

Data Analysis

The BKB-SIN scores, maximum COP-COM distance during the first compensatory step, and reaction time values from the 8 trials of same combination of auditory-balance conditions were averaged by person, resulting in an average outcome score per combination of conditions. For each outcome measure – BKB-SIN, maximum COP-COM distance, and reaction time – repeated measures ANOVA models were run in Stata 13.1 using auditory condition, and balance condition as predictors. All independent variables were coded as categorical with time, auditory, and balance conditions designated as repeated measures variables.

Results

Twenty-five subjects participated in Visit 1, however only 19 subjects completed Visit 2. Three subjects were lost to follow-up, 2 subjects did not pass hearing screening in Visit 1, and 1 subject did not pass vision screening in Visit 1. Baseline characteristics and results of memory and sensory screenings for the 19 subjects are presented in Table 2.

The overall mean BKB-SIN score under the normal auditory condition was 7.7±1.2 dB (Hearing Level) HL and under the simulated hearing loss condition was 16.8±2.5 dB HL. The overall mean BKB-SIN scores under the normal hearing condition with perturbation Level 0 was 8.1±0.8 dB HL, with Level 1 was 7.7±1.3 dB HL, and with Level 2 was 7.2±1.6 dB HL; the overall mean BKB-SIN scores under the simulated hearing loss condition with perturbation Level 0 was 17.9±2.0 dB HL, with Level 1 was 16.2±2.7 dB HL, and with Level 2 was 16.5±2.7 dB HL. Young adults performed significantly worse on the BKB-SIN under the simulated hearing loss condition compared to the normal hearing condition at each surface translation Level 0, 1, and 2 . There was a significant main effect for hearing condition ($p<0.001$); however, there was no significant interaction between hearing condition and perturbation level (Figure 2).There was no significant difference in BKB-SIN scores at surface translation Levels 0, 1, or 2 under Normal Hearing nor at surface translation Levels 0, 1, or 2 under Simulated Hearing Loss condition ($p>0.05$) (Figure 2).

The average for maximum COP-COM distance during the no repeat back condition was 8.9±2.4 centimeters (cm) at Level 1, and 19.0±4.5 cm at Level 2. The average for maximum COP-COM distance during the normal hearing condition was 9.0±2.2 cm at Level 1, and 18.6±4.1 cm at Level 2. The average for maximum COP-COM distance during the simulated hearing loss condition was 9.0±1.9 cm at Level 1, and 18.8±5.3 cm at Level 2. Maximum COP-COM distance was significantly different between surface translation Level 1 and Level 2 ($p<0.001$). However, maximum COP-COM distance was not significantly different between no repeat back, normal hearing, and simulated hearing loss conditions during Level 1 or Level 2 surface translations ($p=0.9496$) (Figure 3). There was no significant interaction between hearing condition and perturbation level.

The average reaction time during the no repeat back condition was 301±51 milliseconds (ms) at Level 1, and 216±14 ms at Level 2. The average reaction time during the normal hearing condition was 308±50 ms at Level 1, and 214±18 ms at Level 2. The average reaction time during the simulated hearing loss condition was 292±55 ms at Level 1, and 210±19 ms at Level 2. Young adults had a significantly faster average reaction time during surface translation Level 2 compared to Level 1 ($p<0.001$). However, reaction time was not significantly different between no repeat back, normal hearing, and simulated hearing loss conditions during Level 1 or

Level 2 surface translations ($p>0.6551$) (Figure 4). There was no significant interaction between hearing condition and perturbation level.

Discussion

The results suggest subjects performed significantly worse on the BKB-SIN under the simulated hearing loss condition. Subjects' maximum COP-COM increased as the perturbation level increased and reaction time decreased as the perturbation level decreased. However, acute sudden simulated hearing loss did not significantly affect the reactive balance outcome measures investigated in this study: maximum COP-COM distance during the first compensatory step, nor the reaction time to initiate the first step. These results suggest that hearing loss may not have a major impact during dual-task situations for young healthy adults.

The results of our study suggest an acute, sudden hearing loss alone may not result in reactive balance deficits in young adults with optimally functioning systems. Similar research with young adults performing single and dual listening-reactive postural task found a significant difference in listening performance, but not reactive balance, during quiet and noisy single- and dual-task conditions, suggesting young adults may have the resources to allocate attention between both tasks.[16]

One explanation could be that young adults typically display flexibility during cognitive-balance dual-task conditions, often performing well on both tasks in cognitive-balance dual-task conditions.[17] Young adults have also been known to prioritize one task over the other, particularly prioritization of the cognitive task when the postural task is not perceived as a threat.[18] In our study, young adults may have prioritized the postural task due to the perturbation levels causing compensatory steps to regain balance. Lastly, young adults may over-perform during dual-task conditions when the single-task is automatic and requires little to no attention or executive functioning to execute the task, leading to confounded results.[19] Our results may, therefore, not reveal the true impact of hearing loss on postural control.

The results from the maximum COP-COM distance during the first reactive step coincide with the literature on surface translations and compensatory stepping, in which a treadmill or platform acceleration of 2.0m/s^2 is fast enough to induce a stepping strategy; thus, leading to the required separation of COP and COM to observe a COP-COM maximum distance.[20] Increased separation of COP-COM during the first compensatory step at Level 2 compared to Level 1 surface translations further supports our analysis. In addition, our results for reaction time during the initial compensatory step coincide with the literature on kinetic and kinematic reaction time, particularly during dual-task conditions, for healthy young adults without balance impairments.[7]

One limitation of our study is that sudden, acute simulated hearing loss cannot recreate the neurodegenerative processes that accompany age-related sensorineural hearing loss, also known as presbycusis.[21] Recent evidence has linked presbyscusis to greater cognitive decline compared to individuals with normal hearing, as well as with neurodegenerative diseases associated with aging, such as Alzheimer’s and Parkinson’s disease.[22,23] Anatomical and functional brain changes occurring independent of age-effects are associated with not only with auditory processing, but also with attention and emotional processing, and have been identified in individuals with age-associated hearing loss.[24] Therefore, manipulating auditory input to simulate hearing loss may not address the full scope of brain function changes occurring in addition to or as a result of hearing loss, particularly with the older adult population.

Our attempt to simulate hearing loss in young healthy adults was unable to elucidate the contribution of hearing loss to balance deficits observed in the older adult populations with hearing loss. Currently, limited research exists describing an underlying mechanism that explains how and why individuals with hearing loss fall more often than individuals with normal hearing, therefore further research is needed. Potential mechanisms to explain the relationship between hearing loss and increased risk for falls includes a: 1) Physiological: various physiological theories exists, such as a shared blood supply of the cochlea and vestibular system or a gene plays a role in the association between hearing loss and balance deficits;[25] 2) Social: the vicious cycle occurs of social isolation due to difficulty hearing and communicating, decreased physical activity, leading to weakness and increased risk for falls;[26] 3) Perceptual: hearing loss may create an incomplete or inaccurate representation of environmental sounds (ie. the proximity a fire truck's siren), putting the individual at risk of unexpected events that could lead to a fall;[27] 4) Cognitive: an individual with hearing loss is constantly performing a dual-task of maintaining balance while processing environmental sounds, such as speech, thus dividing the individual's attention and increasing the risk of falling.[2,28]

Conclusion

In young healthy adults, simulated hearing loss does not negatively impact postural control. Young health adults either prioritize the postural task or simultaneously respond to balance perturbations while attending to auditory task due to sufficient redundancy in the system. Further research is needed to determine whether a cause-and-effect relationship exists between hearing loss and balance deficits and, if so, which underlying mechanisms play a key role. Learning more about the underlying mechanisms will help create clinical assessment tools and interventions for individuals, particularly older adults, with hearing loss and balance deficits.

Acknowledgments

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Table 1. The Randomization Table illustrating the number of trials randomized to different combinations of auditory and surface translations conditions resulting in a total of 64 trails and one set of 30 seconds of quiet stance.

Randomization Table

			Surface Translation Level		
			Level 0	*Level 1*	*Level 2*
Auditory Condition	*Repeat Back*	*Simulated Hearing Loss*	8	8	8
		Normal Hearing	8	8	8
	No Repeat Back	*No Audio*	30sec	8	8
			16	24	24
				64 trials + 30sec Quiet Stance	

Table 2. Average baseline characteristic values and scores among the young, healthy adults.

Baseline Characteristics	
Number of Subjects (n)	19
Age (yrs) (mean ± SD)	27.2 ± 3.1
Height (cm) (mean ± SD)	170.3 ± 9.0
Weight (kg) (mean ± SD)	73.2 ± 11.1
Gender (%)	
Male	42%
Female	58%
Race (%)	
White	63%
Asian	26%
Black	11%
Hand Dominance (%)	
Right	95%
Left	5%

Cognitive and Sensory Screening	
Cognitive	
Word Span Test (mean ± SD)	8.5 ± 1.9
Auditory	
Cerumen Impaction	Negative
Pure-tone threshold	< 20 dB HL at 500-4,000 Hz
Speech-In-Noise (mean % ± SD)	82% ± 8%
Visual	
Eye Chart (mean score)	20/15
Somatosensory	
Ankle Joint Position (%)	100%
Tuning Fork (L foot) (mean ± SD)	7.9 ± 0.2
Tuning Fork (R foot) (mean ± SD)	7.9 ± 0.3
Vestibular	
Dix-Hallpike Maneuver	Negative signs/symptoms
Dynamic Visual Acuity (mean line difference)	0
Vestibular/Ocular Motor Screening	Negative signs/symptoms
Sensory Integration	
Clinical Test of Sensory Interaction and Balance	6/6 conditions

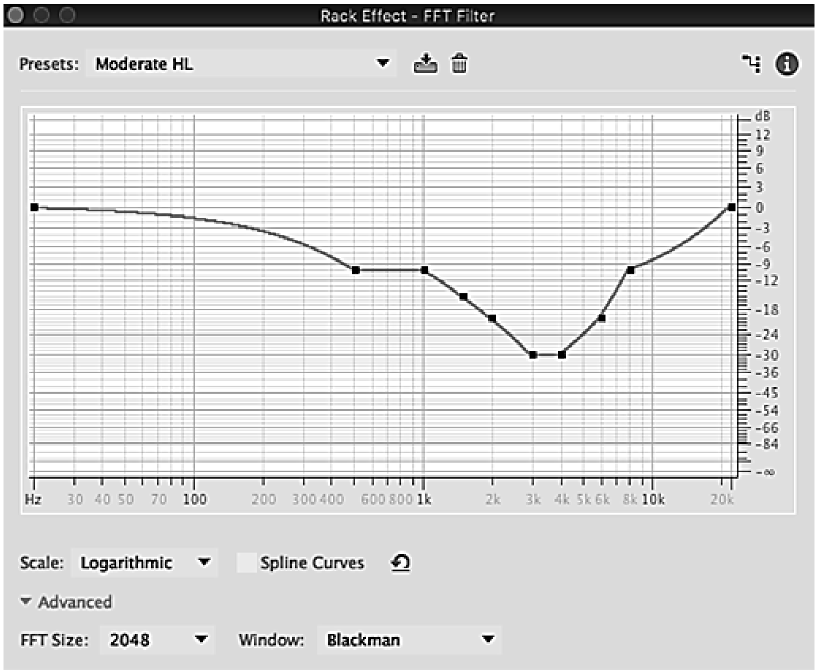

Figure 1. Created in Adobe Audition and used to simulate hearing loss according to defined moderate hearing loss.

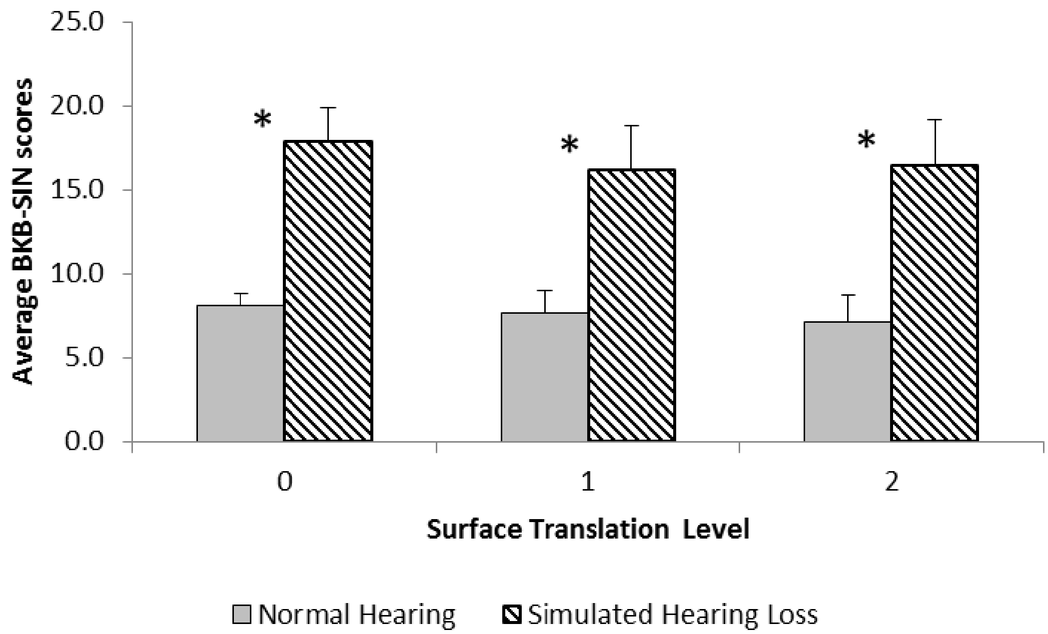

Figure 2. Group averages + stdv for BKB-SIN scores under the normal hearing condition (grey bars) and simulated hearing loss condition (hashed bars) at the three levels of surface translations perturbation: Level 0 = no perturbation of $0 m/s^2$, no repeat back, resulting in single task of maintaining balance; Level 1 = surface translation at $2 m/s^2$; and Level 2 = surface translation at $5 m/s^2$. Combinations of Hearing Conditions and Levels 1 and 2 resulted in dual-task conditions of repeating back sentences while maintaining standing balance in response to surface translation perturbations. The higher the score on BKB-SIN, the lower the performance on the auditory task of repeating back sentences.

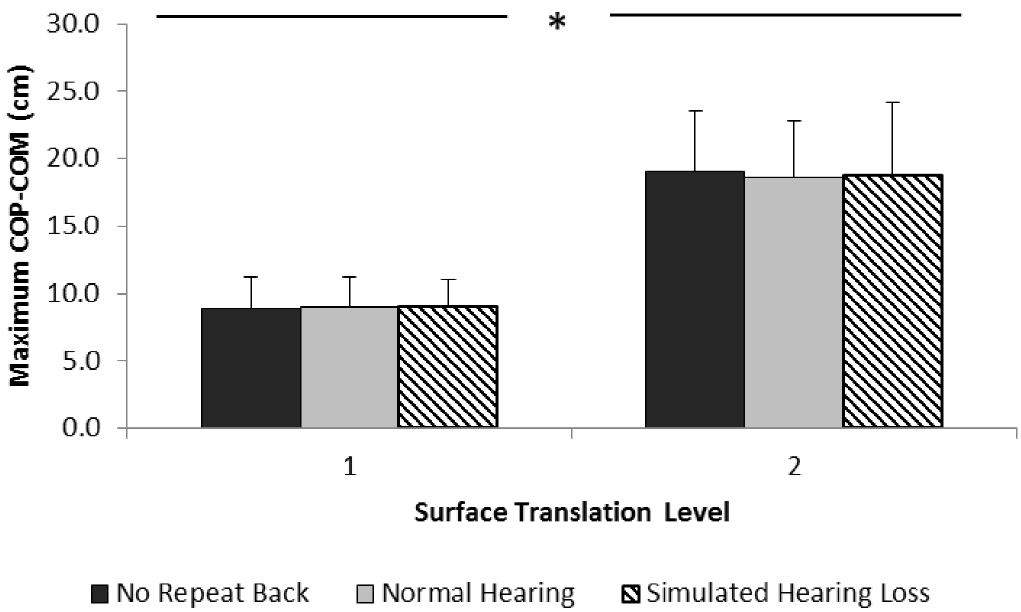

Figure 3. Group averages + stdv of Maximum COP-COM difference (cm) during the first compensatory step in response to surface translations at Level 1 = 2m/s^2 (left bars) and at Level 2 = 5m/s^2 (right bars), under the no repeat back condition (black bars), normal hearing condition (grey bars), and simulated hearing loss condition (hashed bars).

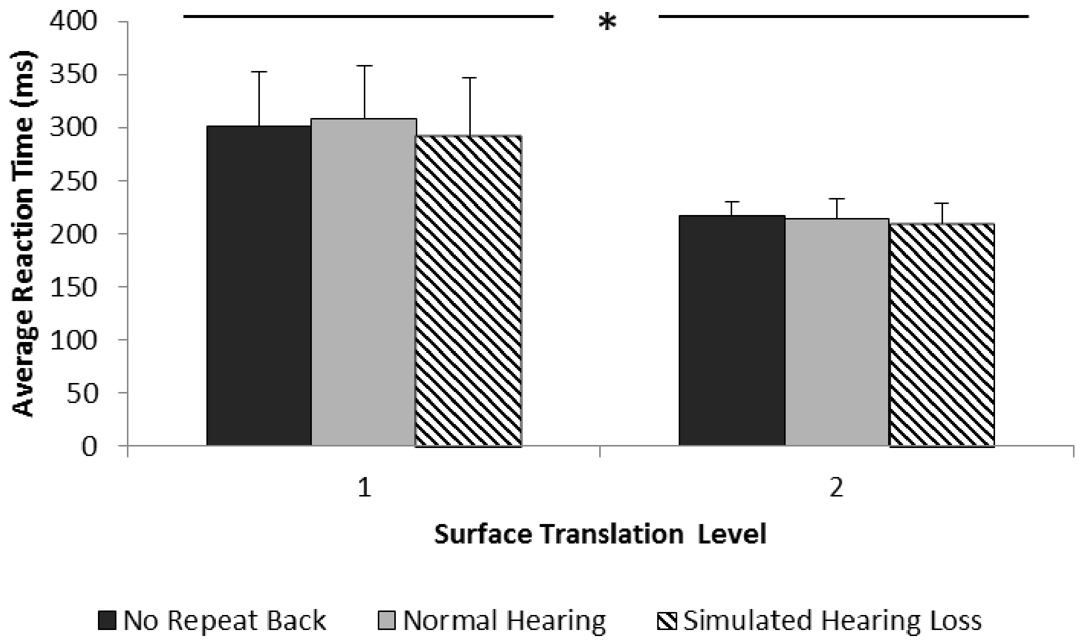

Figure 4. Group averages + stdv of Reaction time (ms) to generate the first compensatory step in response to surface translations at Level 1 = $2 m/s^2$ (left bars) and at Level 2 = $5 m/s^2$ (right bars) , under the no repeat back condition (black bars), normal hearing condition (grey bars), and simulated hearing loss condition (hashed bars).

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CHAPTER 3

SIMULATED HEARING LOSS IN HEALTHY YOUNG AND OLDER ADULTS RESULTS IN POOR REACTION TIME IN OLDER ADULTS

Introduction

Recently, new emerging research has unveiled a correlation between hearing loss and increased risk for falls, particularly in older adults (Agmon, Lavie, & Doumas, 2017). Research has shown individuals with hearing loss walk slower, have a higher incidence of frailty, fall more frequently, undergo hospital admission more often, and have increased Center of Pressure (COP) sway in a noisy environment compared to normal hearing individuals (Çakmur, 2015; Pope, Gallun, & Kampel, 2013; Sogebi, Oluwole, & Mabifah, 2015; Viljanen et al., 2009; Vitkovic, Le, Lee, & Clark, 2016). However, minimal evidence provides a clear cause-and-effect explanation for the observed association between hearing impairment and postural control (Agmon et al., 2017). Several theories exist as to how and why hearing loss affects postural control (Jiam, Li, & Agrawal, 2016). One plausible theory suggests individuals with hearing loss are performing a dual-task when standing or walking while attending to sounds, requiring divided attention or increased cognitive resources, leading to a higher risk of loss of balance and falling (Viljanen et al., 2009). The risk of falling increases as age increases due to a global decline of sensory and motor function (Bolger, Ting, & Sawers, 2014; Brauer, Woollacott, & Shumway-Cook, 2002; Hyodo et al., 2012; Jacobs, Dimitrova, Nutt, & Horak, 2005).

Various studies have simulated hearing loss, in both young and older adults with normal hearing to improve audiological assessments for individuals with hearing loss, to compare the differences between sensorineural hearing loss and acute sudden hearing loss, and to improve hearing aid technology (Bacon, Opie, & Montoya, 1998; Buus & Florentine, 1985; Hornsby, Johnson, & Picou, 2011; Korhonen & Kuk, 2008; McPherson, McMahon, Wilson, & Copland, 2012; Moore, Vickers, Glasberg, & Baer, 1997; Stone & Moore, 1999). Although these studies are beneficial to both audiologists and individuals with hearing loss, no studies have investigated the effect of simulated hearing loss outside of an audiological setting nor have any studies investigated the effect of simulated hearing loss on the control of reactive balance. Laboratory experiments investigating reactive balance have the ability to create unexpected, real-life loss of balance in a safe and controlled environment (Horak, Dimitrova, & Nutt, 2005). Biomechanical outcome measures commonly utilized in research settings to assess reactive balance abilities include: maximum Center of Pressure (COP)-Center of Mass (COM) distance during compensatory steps and reaction time to initiate the first compensatory step (Burleigh, Horak, & Malouin, 1994; Horak et al., 2005; Kanekar & Aruin, 2014; Mansfield, Peters, Liu, & Maki, 2010; McIlroy & Maki, 1996). Maximal displacements of COP and COM have been individually documented in the literature as a stability measures. Another important stability measure is the stability margin, the difference between COP and COM (COP-COM) (Jacobs et al., 2005; Kanegaonkar, Amin, & Clarke, 2012; Santos, Kanekar, & Aruin, 2010; Winter, 1979). COP and COM interact closely with one another to maintain postural stability during both anticipatory stepping, i.e. gait initiation, and compensatory stepping, i.e. unexpected loss of balance (Shumway-Cook & Woollacott, 2007). The COP-COM maximum distance is considered an indicator of robustness in the balance control system (Corriveau, Hébert, Prince, & Raîche, 2001; Jančová, 2008; Lafond, Duarte, & Prince, 2004; Papa, Foreman, & Dibble, 2015; Winter, 1995; Winter, Prince, Frank, Powell, & Zabjek, 1996).

Reaction time is an important fall predictor outcome measure in which fall risk is associated with a slower reaction time (Lord & Fitzpatrick, 2001). In fact, reaction time may be a better assessment tool to determine fall risk compared to walking speed (van den Bogert, Pavol, & Grabiner, 2002). Walking speed is considered the “6th vital sign” of fall risk in older adults (Fritz & Lusardi, 2009).

We created a novel auditory and balance dual-task paradigm to investigate the effect of hearing loss on balance control. Healthy, young and older adults with audiometrically assessed normal hearing abilities performed an auditory task under normal hearing and simulated hearing loss conditions while simultaneously maintaining standing balance during unexpected surface translations. We hypothesized that sudden acute simulated hearing loss would have a greater negative impact on postural control of older adults compared to younger adults; in which the older adults will have decreased maximum COP-COM distance during the first compensatory step, and increased reaction time for initiating the first compensatory step. Furthermore, we hypothesized that both young and older adults will perform worse during the acute, simulated hearing loss condition compared to the normal hearing condition or no repeat back audio condition.

Methods

Subjects

Twenty-five healthy, young adults and 33 healthy, older adults voluntarily agreed to participate and provided Institutional Review Board (IRB)-approved informed consent. All participants were phone screened prior to enrollment to ensure no sensory, auditory, health conditions or balance impairments existed that would restrict exercise ability or confound results of the study. Subjects were excluded if the subject had a history of tinnitus (ringing in the ears), history of motion sickness/dizziness, body weight over 400 lbs, or currently taking medications that affect balance.

Experimental Design

Subjects participated in two visits, consisting of: Visit 1) Cognitive and sensory screening, Visit 2) Balance testing: dual-task auditory and perturbation protocol.

Visit 1: *Cognitive and Sensory Screening*

Participants participated in cognitive, auditory, visual, vestibular, somatosensory, and sensory integration screening to ensure no undiagnosed impairments existed (Table 1). All screening procedures were performed by a licensed physical therapist. Cognitive testing consisted of a Mini Mental State Examination (MMSE) and a Word Span Test (MMSE $\geq$ 24; Word Span Test $\geq$ 4). Speech-in-noise ability has been associated with working memory ability (Ronnberg, Rudner, Foo, & Lunner, 2008); therefore, we wanted to ensure working memory was intact. Hearing screening consisted of a pure-tone threshold testing, speech-in-noise screening, and checking for earwax impaction. Participants underwent pure-tone threshold conduction hearing screening using a MAICO MA 25 portable audiometer and were required to hear within standard auditory threshold (20-55dB HL at 500-4000Hz) guidelines recommended for audiometry screening (Fausti, Wilmington, Helt, Helt, & Konrad-Martin, 2005; Walker, Cleveland, Davis, & Seales, 2013). Participants underwent Speech-In-Noise screening through an online screening test (SIN $\geq$ 60% (Hear-It, 1999). Adults with normal hearing thresholds who fail speech in noise tests may either have excessive difficulty hearing speech in a noisy environment or have a “hidden” hearing loss that cannot diagnosed yet through pure tone audiometry and were excluded from the study (Anderson, Parbery-Clark, Yi, & Kraus, 2011; Liberman, Epstein, Cleveland, Wang, & Maison, 2016). Presence of earwax impaction was determined using a sound probe; inability to view the eardrum would exclude an individual from participating in the study. Visual testing consisted of subjects reading a standard eye chart, in which subjects must have at least 20/20 vision. Vestibular testing consisted of a Dix-Hallpike Maneuver, Vestibular-Occulomotor Screening (VOMS) test, and the Dynamic Visual Acuity (DVA) test to ensure no peripheral or central vestibular disorders (no objective signs or reported symptoms of vestibular disorders) (Demer, Honrubia, & Baloh, 1994; Hoffman, Einstadter, & Kroenke, 1999; Labuguen, 2006; Mucha et al., 2014). Somatosensory testing consisted of joint position testing and tuning fork testing using a 128-Hz tuning fork to rule out impairments with sense of touch (5/5 joint position; Young Adults: ≥5.5, Older Adults: ≥4.5 tuning fork) (Boyle & Negus, 1998; Kästenbauer, Sauseng, Brath, Abrahamian, & Irsigler, 2004). Sensory integration testing was performed using a standard clinical outcome measure known as the Clinical Test of Sensory Interaction and Balance (CTSIB) (maintain balance for 30sec for all 6 conditions) (Shumway-Cook & Horak, 1986). Participants who passed cognitive and sensory screenings were invited back to participate in dual-task auditory and balance testing; participants who failed cognitive and sensory screenings were withdrawn by the Principal Investigator.

Visit 2: *Balance testing: Dual-Task auditory and perturbation protocol.*

The auditory task selected was the standardized audiology test Bamford-Kowal-Bench Speech-In-Noise-Test (BKB-SIN) consisting of a target voice and multi-talker babble. Five second clips of each sentence from the BKB-SIN were uploaded and the voice and the multi-talker babble were separated into 2 separate tracks: Track 1) the target voice and, Track 2) the multi-talker babble. Hearing loss was simulated using the Fast Fourier Transform (FFT) filter (Logarithmic scale, FFT size: 2048, Blackman window) in Adobe Audition, allowing specific decibel (dB) levels to be manipulated at particular frequencies associated with moderate hearing loss (Cruickshanks et al., 1998; Hornsby et al., 2011; Korhonen & Kuk, 2008; McPherson et al., 2012).

After initial introduction to the auditory and the balance single-tasks, subjects performed a randomized single- and dual-task test with a balance condition and an auditory condition. Subjects were required to stand and maintain their balance with and without surface translations while simultaneously listening and repeating back auditory sentences from the BKB-SIN through headphones at 60 dBA (Wilson, McArdle, & Smith, 2007). Subjects performed the auditory test while wearing Bose® QuietComfort 35 wireless headphones to limit additional environmental noise and performed the task under a normal hearing and simulated hearing loss condition. Subjects were also encouraged to guess every time they heard babble through the headphones.

Backward surface translations were delivered and led to the subject experiencing to a forward loss of balance, while he or she simultaneously either repeated or prepared to repeat the sentence. Surface translations were delivered through the treadmill dual-belts at acceleration Level 0 (0 m/s^2), Level 1 (2m/s^2), or Level 2 (5 m/s^2). The loss of balance required participants to take 1 or more compensatory steps forward to maintain their balance. An overhead harness system equipped to support up to 181 kg was in place to prevent participants from hitting the ground if a fall would occur.

A 12 camera Motion Analysis System collected kinematic data from 54 reflective markers placed on anatomical landmarks of the body. The V-gait treadmill system by Motek

Medical containing 2 separate force plates mounted underneath each belt was used to deliver surface translations and record force data.

Single- and dual-task auditory-balance conditions were randomized to control for a learning effect with the BKB-SIN and reactive balance ability (Cainer, James, & Rajan, 2008; Lussier, Gagnon, & Bherer, 2012; Mansfield, Peters, Liu, & Maki, 2007). Randomization occurred in a data excel file according to perturbation level, hearing condition, and Signal-To-Noise (SNR) level; resulting in a total of 64 conditions (Table 2). Conditions were then divided into 4 sets to allow participants to have a mental and physical break between bouts of performing the dual-task.

Outcome measures

Primary outcome measures were: performance on the BKB-SIN, COP-COM maximal distance during the first compensatory step, and reaction time for initiating the first compensatory step.

The BKB-SIN is a Speech-In-Noise outcome measure developed to test SNR loss for individuals with hearing impairment, as well as the necessity of a hearing aid (Beck & Nilsson, 2013; Bench, Kowal, & Bamford, 1979). The score provides the SNR required for the test-taker repeat the full sentence accurately 50% of the time, with a higher score indicating a poorer performance (ETYMOTIC, 2018).

During compensatory stepping there is an initial COM displacement followed by COP displacement that regains balance equilibrium and maintains postural stability (Henry, Fung, & Horak, 2001; Horak et al., 2005; Santos et al., 2010; Winter, Patla, & Frank, 1990; J. Yang, Winter, & Wells, 1990). Because COP and COM are interconnected during compensatory stepping, the maximum COP-COM distance was chosen as the most appropriate outcome measure of postural stability during unexpected compensatory steps (Kanekar & Aruin, 2014).

Data Processing

All kinematic and force data were processed using MATLAB. COP was measured using ground reaction forces from the force plate data and COM was extrapolated using the sacral marker (F. Yang & Pai, 2014). Baseline values for COP-COM were calculated in the 30 frames before the surface translation (0.25 seconds). The maximum COP-COM distance during the first compensatory step was calculated within the window of time from surface perturbation to completion of the first compensatory step, indicated by placement of the stepping leg heel on the force plate. Both heel marker data and force data were used to confirm initiation and completion of reactive step. All maximum COP-COM distances were normalized to subject height. Reaction time was determined as the time in milliseconds (ms) from start of surface perturbation to the point in time when the heel marker of the stepping foot transitioned from accelerating backward while receiving the surface translation to accelerating forward in the opposite direction while responding with a compensatory step. Calculation of reaction time was assessed within the window from surface perturbation initiation to maximum acceleration of the heel marker during first compensatory forward step.

Data Analysis

For each individual, the values for each outcome (i.e., reaction time and maximum COP-COM difference) for each variation set—3 hearing levels X 3 perturbation levels in a randomized sequence—were averaged, resulting in an average outcome score by variation set (each individual has 8 variation sets). For each outcome, repeated measures ANOVA models were run in Stata 13.1 using group, time, hearing levels, perturbation levels, and a hearing-by-perturbation levels interaction term as predictors. The between-subjects error terms was designated as “subject in age group”, “subject” was designated as the variable representing the lowest unit in the between-subjects error term, and “age group” was designated as the variable for computing the pooled covariance matrix.

Results

Twenty-five young adults and 33 older adults participated in Visit 1. For the young adults, three subjects did not pass Visit 1 screening and three were lost to follow-up. For the older adults, one subject was lost to follow-up, eleven subjects did not pass Visit 1 screening, and one subject withdrew from the study. A total of nineteen young adults and twenty older adults completed both study visits, their baseline characteristics and screening tests scores are provided in Table 3.

For the young adults, the overall mean BKB-SIN scores under the normal hearing condition with perturbation Level 0 was 8.1±0.8 dB Hearing Level (HL), with Level 1 was 7.7±1.3 dB HL, and with Level 2 was 7.2±1.6 dB HL; the overall mean BKB-SIN scores under the simulated hearing loss condition with perturbation Level 0 was 17.9±2.0 dB HL, with Level 1 was 16.2±2.7 dB HL, and with Level 2 was 16.5±2.7 dB HL. For the older adults, the overall mean BKB-SIN scores under the normal hearing condition with perturbation Level 0 was 8.0±0.6 dB HL, with Level 1 was 8.2±1.7 dB HL, and with Level 2 was 8.5±1.0 dB HL; the overall mean BKB-SIN scores under the simulated hearing loss condition with perturbation Level 0 was 18.3±1.4 dB HL, with Level 1 was 14.9±2.3 dB HL, and with Level 2 was 16.5±1.8

dB HL. Both young and older adults performed significantly worse on the BKB-SIN under the simulated hearing loss condition compared to the normal hearing condition at each surface translation Level 0, 1, and 2 ($p<0.001$), but no difference existed between groups ($p>0.05$) (Figure 1).

For the young adults, the average maximal COP-COM distance for the no audio condition at Level 1 was 8.9±2.4 cm and with Level 2 was 19.0±4.5 cm, for the normal hearing condition at Level 1 was 9.0±2.2 cm and with Level 2 was 18.6±4.1 cm, for the simulated hearing loss condition at Level 1 was 9.0±1.9 cm and with Level 2 was 18.8±5.3 cm. For the older adults, the average maximal COP-COM distance for the no audio condition at Level 1 was 9.1±1.5 cm and with Level 2 was 18.9±2.1 cm, for the normal hearing condition at Level 1 was 9.0±1.4 cm and with Level 2 was 18.5±2.0 cm, for the simulated hearing loss condition at Level 1 was 8.9±1.3 cm and with Level 2 was 18.4±2.1 cm. Both young and older adults had significantly worse average maximum COP-COM distance during Level 2 surface translation compared to Level 1 ($p<0.001$), but no interaction existed between surface translation level and auditory condition for maximum COP-COM distance ($p>0.05$) (Figure 2).

For the young adults, the reaction time for the no audio condition was 301±51 ms at Level 1 and was 216±14 ms at Level 2; for the normal hearing condition it was 308±50 ms at Level 1 and was 214±18 ms at Level 2; for the simulated hearing loss condition it was 292±55 ms at Level 1 and was 210±19 ms at Level 2. For the older adults, the reaction time for the no audio condition at Level 1 was 294±43 ms and with Level 2 it was 226±18 ms; for the normal hearing condition at Level 1 was 298±42 ms and with Level 2 it was 230±20 ms, for the simulated hearing loss condition at Level 1 was 283±37 ms and with Level 2 was 224±20 ms. For reaction time, the results of repeated measure ANOVA indicated that after controlling for all other variables in the model, there were significant main effects for perturbation levels and hearing levels, and a significant interaction effect for perturbation by hearing levels; time was not significant. Both young and older adults had significantly shorter reaction time in response to the Level 2 surface translation compared to the Level 1 ($p<0.001$) (Figure 3). A significant interaction was found between perturbation level and auditory condition for reaction time ($p=0.0191$). Across all participants, the marginal means scores indicated that the longest reaction times were recorded for perturbation level 1 with normal hearing and shortest for perturbation level 2 with hearing loss. Differences in reaction times between level 1 and level 2 perturbations, by hearing condition and group, illustrate the interaction effect ($p<0.05$) (Figure 4).

Discussion

The results suggest both young and older adults perform significantly worse on the BKB-SIN under the simulated hearing loss condition, thus confirming successful simulation of hearing loss. Sudden, acute simulated hearing loss significantly worsened BKB-SIN scores to comparable levels documented in the literature values of moderate-to-severe SNR experienced by individuals with hearing loss (Cruickshanks et al., 1998; ETYMOTIC, 2018). Typically during simulated hearing loss studies, adults are grouped together based on similar hearing thresholds regardless of age (Stone & Moore, 1999). Because we screened all adults to be at similar hearing thresholds, our results of no group differences on performance of the BKB-SIN are as expected.

The maximum COP-COM distance increased as the perturbation level increased. Our results for maximum COP-COM distance during the first reactive step are aligned with the literature on COP-COM values during surface translations and compensatory stepping, in which a platform acceleration of $2.0 m/s^2$ is fast enough to induce a stepping strategy; our values for maximum COP-COM during perturbation Level 1 are larger compared to literature values of COP-COM during static stance. (Henry et al., 2001; Horak et al., 2005; McIlroy & Maki, 1996; Santos et al., 2010; Winter et al., 1990; J. Yang et al., 1990). This is to be expected because a stepping strategy requires greater dissociation of COP from COM compared to quiet standing control strategy in which the two variables track closely together (Woollacott & Shumway-Cook, 2002). Our COP-COM results during perturbation Level 2 matched closely with literature values of COP-COM results using the tether-release model (Henry et al., 2001; Horak et al., 2005; McIlroy & Maki, 1996; Papa et al., 2015; Santos et al., 2010; Winter et al., 1990; J. Yang et al., 1990). However, because there were no significant differences in maximum COP-COM distance between young and old adults, or between normal and simulated hearing loss conditions, the maximum COP-COM distance may not be an effective standalone outcome measure for identifying individuals with hearing loss and balance deficits.

Our reaction time results during the initial compensatory step corresponds with the literature on kinetic and kinematic reaction time, particularly during dual-task conditions, for both normal, healthy young and older adults (Brauer et al., 2002; Luchies et al., 2002). A similar study performed by Troy and Grabiner (2006) assessed that change in trajectory for the lateral malleoli marker, indicating reaction time, occurred at ~200ms during a slip on a platform in young adults. Our result that healthy, young and older adults do not have a difference in reaction time during no repeat back and normal hearing conditions also aligns with literature on healthy young and older adults, particularly during dual-task conditions (Brown, Shumway-Cook, & Woollacott, 1999; Luchies et al., 2002). Overall, the reaction time decreased between perturbations at Level 1 and 2, as it would be expected, because a faster perturbation level requires a shorter reaction time to initiate a compensatory step. Reaction time was shown to decrease in young adults as speed of surface translation increased (Bhatt, Wening, & Pai, 2005). However, the pattern of change in reaction time we observed was not similar in young and older adults or across hearing conditions. Older adults responding to perturbations under simulated hearing loss did not decrease their reaction time just as much as they did under no repeat back or normal hearing condition (smaller difference between level 2 and level 1 noted in figure 4).

Young adults were able to adapt their initiation, decrease reaction time, execute compensatory stepping, and increase the maximum COP-COM distance, as they responded to progressively more challenging perturbations of balance. They did so regardless of hearing conditions, confirming that either they have enough redundancy in the balance control system or that the dual-task was not challenging enough to affect their balance control. In older adults with simulated hearing loss, attending to auditory input does not change the execution of compensatory steps (no significant changes in COP-COM) but attending to auditory input may play a role in ability to modulate the initiation of reactive balance strategies during challenging dual-task situations (significant smaller difference in reaction time responding to level 2 vs level 1 perturbations under simulated hearing loss compared to no repeat back and normal hearing conditions).

Additional research is beginning to test various types of adults in dual-task conditions of a balance-auditory/cognitive task to determine whether individuals with hearing loss have poorer balance measures (Bruce et al., 2017; Lau, Pichora-Fuller, Li, Singh, & Campos, 2016). One study, performed by Bruce et al. (2017) observed the effect of a reactive balance task and a simultaneous cognitive task in the presence of babble. Their results found that healthy young adults have more flexibility responding to dual cognitive-balance tasks, while healthy older adults prioritize their balance over the auditory task (Bruce et al., 2017). Another study by Lau et al. (2016) determined that both older adults with normal hearing and older adults with hearing loss prioritize their balance over the listening task. Due to the variability of the outcome measures in the dual-task studies, not enough evidence exists to confirm or deny whether cognitive processing plays a role in causing older adults with hearing loss to have balance deficits. More research is needed to determine whether dual-task conditions requiring cognitive resources negatively affect balance control for older adults with hearing loss.

One limitation to our study is that males and females were not equally represented in the older adult group; we had 75% females and 25 % males. However, hearing loss is more common in males compared to females (Lin, Niparko, & Ferrucci, 2011). Males who reported symptoms of hearing loss during the phone screen or males who did not pass audiometer testing at Visit 1 were not eligible to participate in Visit 2, limiting the number of males eligible to participate in Visit 2.

Conclusion

Sudden, acute simulated hearing loss negatively affects auditory scores for both young and older adults. The execution of compensatory stepping response seems to be maintained with aging under all conditions of hearing. However, the ability to modulate reaction time and initiate compensatory steps to regain balance as a function of challenge difficulty is negatively impacted by simulated hearing loss in older adults. Individuals with hearing loss may be at greater risk of falling compared to individuals with normal hearing due to age-related cognitive and neurodegenerative changes associated with hearing loss (Mudar & Husain, 2016). Further research is needed to clarify whether a single or multi- cause and effect relationship exists between hearing loss and balance deficits.

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Table 1. The cut-off scores are provided for the Cognitive and Sensory screening assessments performed during Visit 1.

Cognitive and Sensory Screening: Cut-off Scores

Cognitive	
MMSE	24
Word Span Test	4
Auditory	
Cerumen Impaction	Visible eardrum
Pure-tone threshold	Young adults: < 20 dB HL at 500-4,000 Hz Older adults: < 30 dB HL at 500-4,000 Hz
Speech-In-Noise	60%
Visual	
Eye Chart (mean score)	20/20
Somatosensory	
Ankle Joint Position	100%
Tuning Fork	Young adults: 4.5 Older adults: 5.5
Vestibular	
Dix-Hallpike Maneuver	Negative signs/symptoms
Dynamic Visual Acuity	3 lines above baseline vision
Vestibular/Ocular Motor Screening	Negative signs/symptoms
Sensory Integration	
Clinical Test of Sensory Interaction and Balance	6/6 conditions

Table 2. The Randomization Table illustrating the number of trials randomized to different combinations of auditory and surface translations conditions resulting in a total of 64 trails and one set of 30 seconds of quiet stance.

<table>
<tr><td colspan="6">Randomization Table</td></tr>
<tr><td colspan="3" rowspan="2"></td><td colspan="3">Perturbation Level</td></tr>
<tr><td>Level 0</td><td>Level 1</td><td>Level 2</td></tr>
<tr><td rowspan="3">Auditory Condition</td><td rowspan="2">Repeat Back</td><td>Simulated Hearing Loss</td><td>8</td><td>8</td><td>8</td></tr>
<tr><td>Normal Hearing</td><td>8</td><td>8</td><td>8</td></tr>
<tr><td>No audio</td><td>No Audio</td><td>30sec</td><td>8</td><td>8</td></tr>
<tr><td colspan="3" rowspan="2"></td><td>16</td><td>24</td><td>24</td></tr>
<tr><td colspan="3">64 conditions
+ 30sec Quiet Stance</td></tr>
</table>

Table 3. Average baseline characteristic values and scores among the young and older adults.

Baseline Characteristics		
	Young Adults	**Older Adults**
Number of Participants (n)	19	20
Age (yrs) (mean ± SD)	27.2 ± 3.1	68.7 ± 4.3
Height (cm) (mean ± SD)	170.3 ± 9.0	163.6 ± 8.2
Weight (kg) (mean ± SD)	73.2 ± 11.1	70.5 ± 19.0
Gender (%)		
Male	42%	25%
Female	58%	75%
Race (%)		
White	63%	95%
Asian	26%	5%
Black	11%	0%
Cognitive and Sensory Screening		
Cognitive		
MMSE (mean ± SD)	28.9 ± 1.2	28.9 ± 1.4
Word Span Test (mean ± SD)	8.5 ± 1.9	6.6 ± 2.5
Auditory		
Cerumen Impaction	Negative	Negative
Pure-tone threshold	<20 dB HL at 500-4,000 Hz	<30 dB HL at 500-4,000Hz
Speech-In-Noise (mean % ± SD)	82% ± 8%	73% ± 7%
Visual		
Eye Chart (mean score)	20/15	20/20
Somatosensory		
Ankle Joint Position (%)	100%	100%
Tuning Fork (L foot) (mean ± SD)	7.9 ± 0.2	6.8 ± 0.9
Tuning Fork (R foot) (mean ± SD)	7.9 ± 0.3	6.7 ± 1.1
Vestibular		
Dix-Hallpike Maneuver	Negative signs/symptoms	Negative signs/symptoms
Dynamic Visual Acuity (mean line difference)	0	0.7 ± 0.8
Vestibular/Ocular Motor Screening	Negative signs/symptoms	Negative signs/symptoms
Sensory Integration Clinical Test of Sensory Interaction and Balance (CTSIB)	6/6 conditions	6/6 conditions

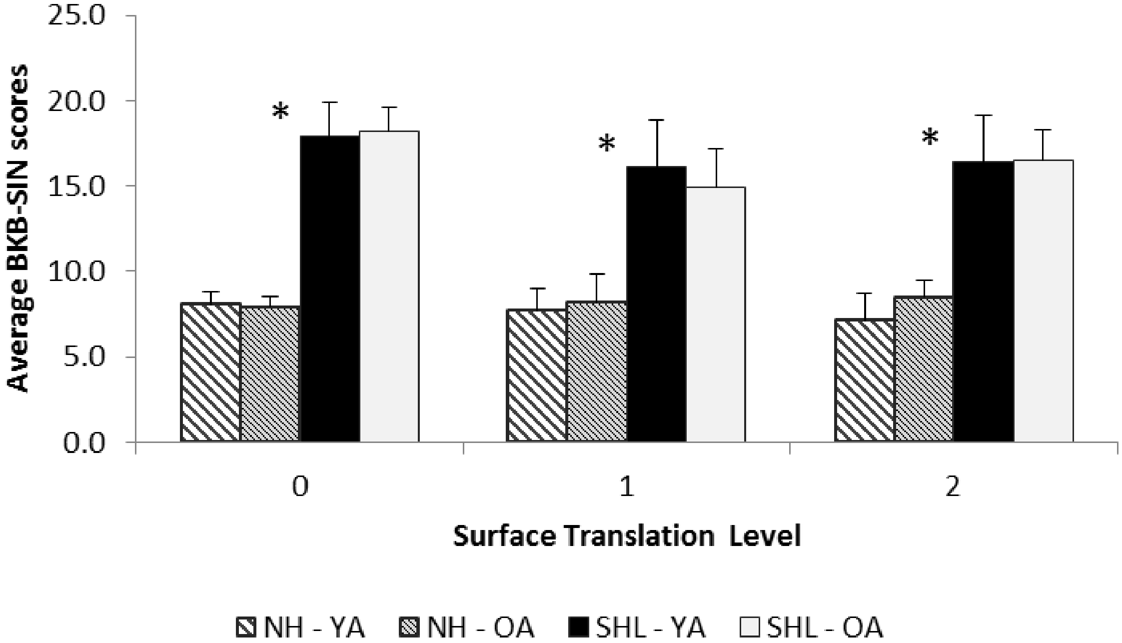

Figure 1. Group averages + stdv for BKB-SIN scores under the Normal Hearing (NH) condition (hashed bars) of Young Adults (YA) and Older Adults (OA) and Simulated Hearing Loss (SHL) condition (colored bars) at the three levels of surface translations perturbation: Level 0 = surface translation at 0m/s^2, resulting in single task of repeating back; Level 1 = surface translation at 2m/s^2, and Level 2 = surface translation at 5m/s^2. Combinations of Hearing Conditions and Levels 1 and 2 resulted in dual-task conditions of repeating back sentences while maintaining standing balance in response to surface translation perturbations. The higher the score on BKB-SIN, the lower the performance on the auditory task of repeating back sentences.

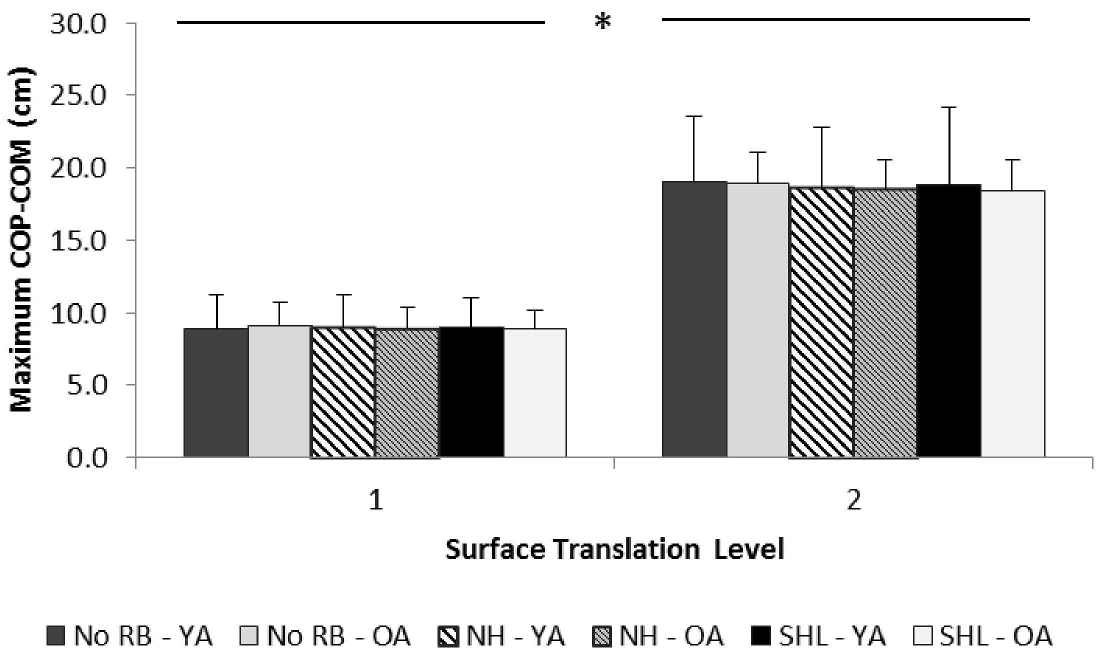

Figure 2. Group averages + stdv of Maximum COP-COM difference (cm) during the first compensatory step in response to surface translations at Level 1 = 2m/s^2 (left bars) and at Level 2 = 5m/s^2 (right bars), under the No repeat back condition (grey bars) , normal hearing condition (hashed bars) and simulated hearing loss condition (black and white bars). No RB = No Repeat Back/Audio condition, NH = Normal Hearing condition, SHL = Simulated Hearing Loss condition, YA = Young Adults, OA = Older Adults.

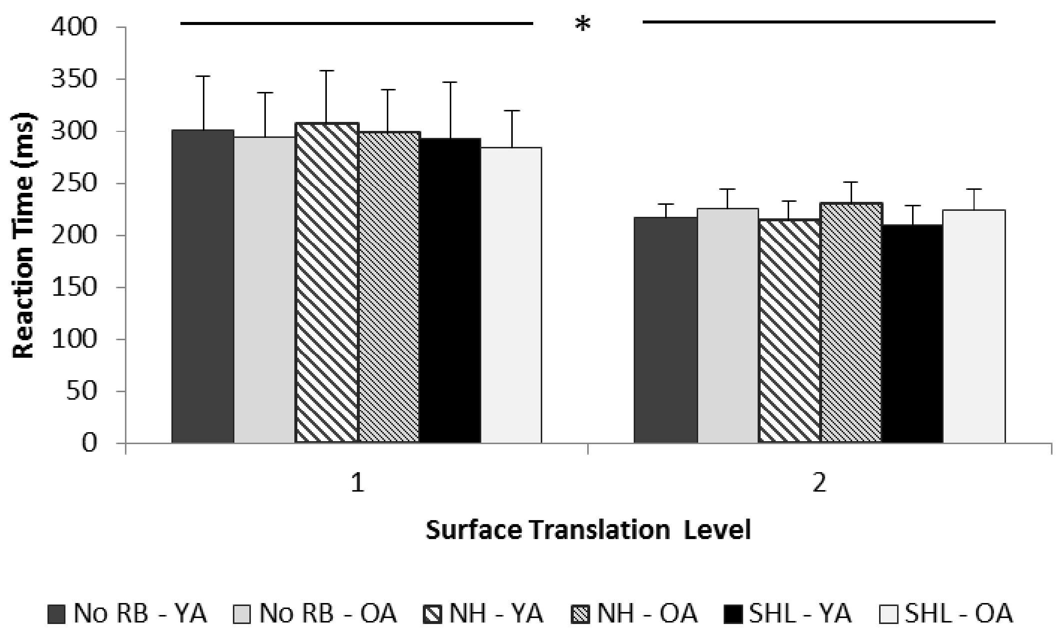

Figure 3. Group averages + stdv of Reaction Time (ms) during the first compensatory step in response to surface translations at Level 1 = $2m/s^2$ (left bars) and at Level 2 = $5m/s^2$ (right bars), under the No repeat back condition (grey bars) , normal hearing condition (hashed bars) and simulated hearing loss condition (black and white bars). No RB = No Repeat Back/Audio condition, NH = Normal Hearing condition, SHL = Simulated Hearing Loss condition, YA = Young Adults, OA = Older Adults.

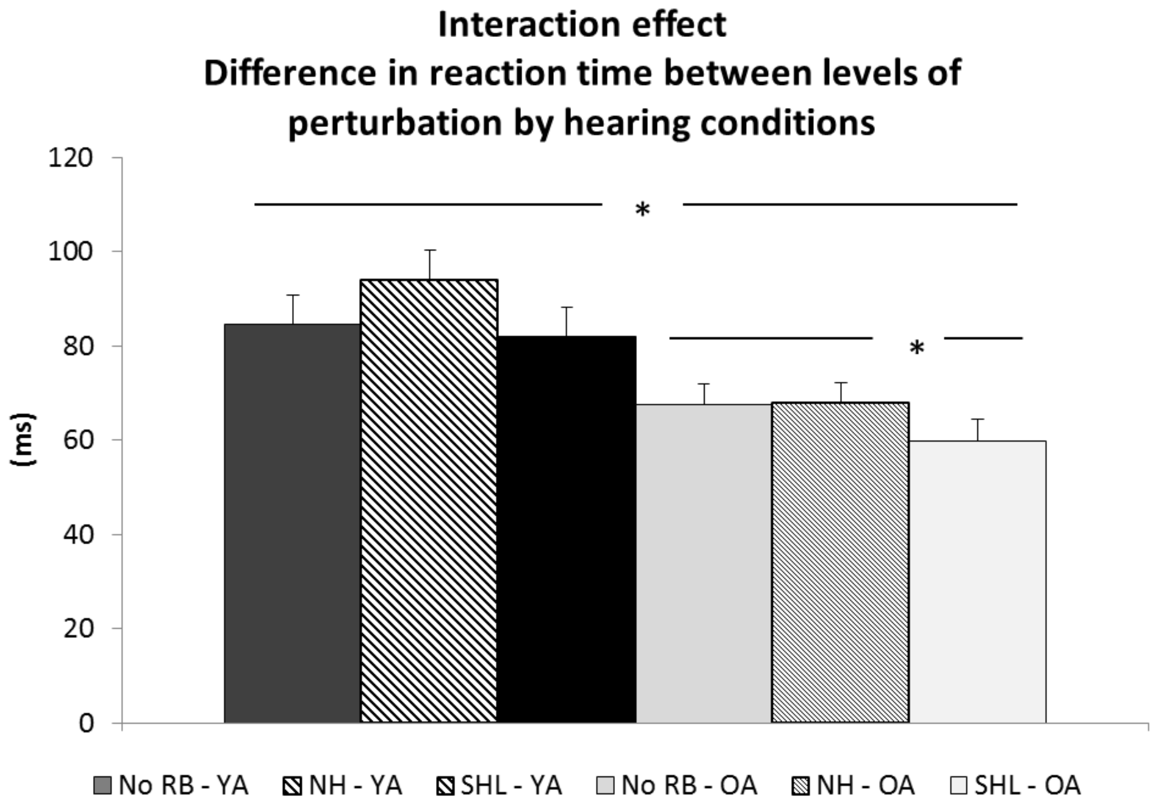

Figure 4. Group averages + stdv of differences between Reaction Time (ms) at Level 1 = $2m/s^2$ and at Level 2 = $5m/s^2$, under the No repeat back condition (right bars - dark grey and large hashed bars) , normal hearing condition (middle bars - black and medium grey bars) and simulated hearing loss condition (left bars - small hashed and light grey bars). No RB = No Repeat Back/Audio condition, NH = Normal Hearing condition, SHL = Simulated Hearing Loss condition, YA = Young Adults, OA = Older Adults.

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CHAPTER 4

HEARING AIDS DO NOT IMPROVE REACTIVE BALANCE IN OLDER ADULTS WITH HEARING LOSS

Introduction

Hearing loss, particularly with older adults, has been linked to many negative consequences, such as difficulty with communication, depression, and increased risk for dementia (Brink & Stones, 2007; Dawes et al., 2015; Heyl & Wahl, 2012). Recent epidemiological research has also linked hearing loss in older adults to increased risk for falls. Individuals with hearing loss have been found to have slower gait speed compared to normative values, report increased difficulties with Activities of Daily Living (ADLs), have injurious falls more often, and have worse standing balance in noisy environments compared to individuals with normal hearing (Grue, Kirkevold, & Ranhoff, 2009; Sogebi, Oluwole, & Mabifah, 2015; Viljanen et al., 2009; Vitkovic, Le, Lee, & Clark, 2016).

Hearing aids are the most common treatment for hearing loss. It would follow that advances in hearing aid technology – including digital and frequency-modulated systems that allow for custom tuning of devices based on individual needs and miniaturization of hearing aids that them less noticeable – would translate into hearing aids being widely used (McCormack & Fortnum, 2013; Thibodeau, 2010). However, many older adults who would benefit from hearing aids do not wear hearing aids (Lupsakko, Kautiainen, & Sulkava, 2005). Less than 4% of older adults with sensorineural or conductive hearing loss wear hearing aids, which equals an estimated 22.9 million Americans. Reasons for low adoption rates of hearing aids include the high cost and the necessary adaptation period for neuroplasticity to take place before noticing a benefit in hearing ability (Castiglione et al., 2016; Cox & Alexander, 1992; Lupsakko et al., 2005). Lupsakko et al. (2005) found that 25% of subjects who had hearing aids did not wear their listening devices regularly for subjective reasons, such as considering the device: i) unnecessary / not providing benefit (42% of subjects), ii) hard to use (21% of subjects) or iii) defective (17% of subjects). Further confounding the hearing aid controversy is the evidence that suggests high working memory may influence listening abilities and may reduce the benefit of a hearing aid (Chia et al., 2007; Hallgren, Larsby, Lyxell, & Arlinger, 2005; Lupsakko et al., 2005; Meister, Walger, Brehmer, von Wedel, & von Wedel, 2008; Ng, Rudner, Lunner, Pedersen, & Ronnberg, 2013). Promising recent evidence, on the contrary, suggests that hearing aids, as part of a comprehensive rehabilitation program, may improve listening ability and slow or prevent cognitive decline for older adults with hearing loss (Mudar & Husain, 2016).

Even though hearing aids do not completely restore speech understanding, particularly in a noisy environment, the literature suggests hearing aids may still improve communication, decrease depressive symptoms, and may allow listening tasks to be less cognitively taxing; thereby preventing mental fatigue, information overload, and cognitive decline (Arlinger, 2003; Lupsakko et al., 2005; Young Choi, Shim, Lee, Yoon, & Joo, 2011). Furthermore, hearing aids may even improve physical functioning and increase independence performing ADLs (Hogan, O'Loughlin, Miller, & Kendig, 2009; Lupsakko et al., 2005). Although this evidence is insightful, to date no experimental studies tested whether a decrease in cognitive load is the reason why a hearing aids improve physical functioning (Lin et al., 2013; Vitkovic et al., 2016). In a real-world setting, older adults with hearing loss are constantly attending to sounds, such as talking, while simultaneously attempting to stand, walk, or cross obstacles. It is possible that when performing balance tasks while attending to sounds, like standing or walking, older adults with hearing loss are performing dual-tasks of and may have a higher risk of loss of balance and falling (Bruce et al., 2017).

Because hearing aids improve sound recognition, it is plausible that hearing aids may provide a possible solution to improving standing or walking balance for older adults with hearing loss by easing difficulty of the listening task (Vitkovic et al., 2016). However, the impact of hearing aids on balance control – whether hearing aids may improve balance and through what mechanism(s) – is not fully understood (Negahban & Nassadj, 2017).

To our knowledge, there is a paucity of evidence regarding the effect of hearing loss on ability to regain balance after a stumble, trip, slip, or near loss of balance. Few studies have investigated the effect of hearing loss on reactive balance, and whether hearing aids may improve control of reactive balance, in an older adult population (Bruce et al., 2017). Experiments assessing the control of reactive balance are particularly translational due to the creation of unexpected, "real-life" loss of balance situations in a controlled environment. Biomechanical outcome measures commonly used to analyze reactive balance are: reaction time to generate the compensatory step after a loss of balance, and the relationship between the Center of Pressure (COP) and Center of Mass (COM) during compensatory steps (Burleigh, Horak, & Malouin, 1994; Horak, Dimitrova, & Nutt, 2005; Kanekar & Aruin, 2014; McIlroy & Maki, 1996). The maximum COP-COM distance is considered a robust indicator of balance control ability in the human body (Corriveau, Hébert, Prince, & Raîche, 2001; Jančová, 2008; Lafond, Duarte, & Prince, 2004; Papa, Foreman, & Dibble, 2015; Winter, 1995; Winter, Prince, Frank, Powell, & Zabjek, 1996).

We created a novel auditory and balance dual-task paradigm to investigate the effect of hearing loss on reactive balance control in older adult hearing aid users, and whether hearing aids improve reactive balance control. Older adults with hearing loss, who were regular hearing aid users, performed an auditory test under normal hearing (hearing aid) and hearing loss (no hearing aid) conditions while simultaneously maintaining standing balance during unexpected surface translations. We intend to test whether a cognitive mechanism explains why hearing aids improve balance for older adults with hearing loss. We theorize older adults with hearing loss who wear a hearing aid do not need to attend as closely to unheard speech or sounds and can dedicate more resources to balance control. We hypothesized hearing aids would improve reactive balance control for older adults with hearing loss shown by an increased maximum COP-COM distance during the first compensatory step, decreased reaction time for initiating the first compensatory step, and an improved score on the auditory test. Furthermore, we hypothesized older adults with hearing loss would have improved scores on the single-task condition compared to the dual-task condition.

Methods

Subjects

Twenty-three older adults with diagnosed bilateral hearing loss and fitted with hearing aids in both ears participated in the research study. All participants were verbally screened to ensure no vestibular, neuromuscular, or cardiovascular conditions existed that would impair balance or restrict exercise. Participants received an initial phone screen that included a verbal explanation of the research study prior to visiting the lab. All subjects voluntarily agreed to participate through Institutional Review Board (IRB) approved written informed consent.

Subjects were excluded if they had a history of motion sickness or dizziness, inability to stand independently for at least 1 minute, body weight over 181 kg, a history of falls (>3 falls within the past 6 months), or currently taking medications that affect balance. Subjects were also excluded if they had not used a hearing aid more than 2 months, had not underwent a hearing aid fitting within the past year, or did not use the hearing aid for at least 8 hours per day. Due to the requirement that subjects were fitted for hearing aids prior to enrollment, proof of hearing loss through an audiogram was not required.

Experimental Design

Subjects participated in two visits, consisting of: Visit 1) Cognitive and sensory screening, Visit 2) Balance testing: Dual-Task auditory and perturbation protocol.

Visit 1: *Cognitive and Sensory Screening*

Cognitive and sensory screenings were used to rule-out any additional undiagnosed impairments. All screening was performed by a licensed healthcare professional. Participants who passed cognitive and sensory screenings were eligible to participate in dual-task auditory and balance testing; participants who failed cognitive and sensory screenings were withdrawn the study.

Visit 2: *Dual-Task auditory and balance testing protocol.*

Protocol

Participants were first introduced to the balance and auditory tasks and were instructed to "do whatever is natural to you to keep your balance" and encouraged to guess every time they heard babble. Subjects were asked to stand and maintain their balance while simultaneously listening and repeating back auditory sentences from the Bamford-Kowel-Bench Speech-In-Noise (BKB-SIN), a standardized audiology test, played through the speakers at 60 dBA (Wilson, McArdle, & Smith, 2007). There were three auditory conditions: 1) no repeat back resulting in the single-task of maintaining balance; 2) hearing aid condition in which participants performed dual-tasks while wearing their hearing aids (equivalent to their "normal hearing") and 3) no hearing aid condition in which participants performed the dual-tasks without their hearing aids (equivalent to "hearing loss"). A higher score on the BKB-SIN indicates a poorer performance. Only 2 subjects required visual cues when hearing aids were not worn.

Surface translations were administered in the backward direction while the subjects simultaneously repeated back sentences (dual-task conditions) or did no repeat back sentences (single-task conditions). When the treadmill accelerated backward, participants experienced a forward loss of balance requiring participants to take 1 or more steps forward to maintain their balance. Surface translations were administered using the treadmill dual-belts at the following accelerations: Level 0 (0 m/s^2) no backward surface translations, resulting in the single task of listening and repeating back the sentence; Level 1 (2m/s^2) or Level 2 (5 m/s^2), resulting in dual-tasks of repeating back sentences while responding to balance perturbations. An overhead harness system able to support 181 kg was attached to each subject to prevent the subject from hitting the ground if a fall would occur.

A 12 camera Motion Analysis System capture system collected kinematic data from 54 reflective markers placed on anatomical landmarks of the body (reference). The V-gait treadmill system (by Motek Medical) with two separate force plates mounted underneath each belt was used to deliver surface perturbations and record force data.

Single- and dual-task auditory-balance conditions were randomized to control for a learning effect with the BKB-SIN and reactive balance ability (Cainer, James, & Rajan, 2008; Lussier, Gagnon, & Bherer, 2012; Mansfield, Peters, Liu, & Maki, 2007). Randomization occurred according to perturbation level, hearing condition, and Signal-To-Noise (SNR) level; resulting in a total of 64 conditions (Table 1). Subjects performed half of testing with their hearing aid and half of testing without their hearing aid. The order of hearing aid use (first or second) was randomized among subjects based on the order the subjects participated in Visit 2.

Outcome measures

Primary outcome measures included: BKB-SIN performance, maximum Center of Pressure (COP) - Center of Mass (COM) (COP-COM) distance during the first compensatory step, and reaction time for initiating the first compensatory step.

The BKB-SIN is a Speech-In-Noise (SIN) outcome measure developed to test an individual's ability to hear speech in noise, which determines the necessity and benefit of a hearing aid (Beck & Nilsson, 2013; Bench, Kowal, & Bamford, 1979). The sentences and words are easily understandable for all ages, making the test appropriate for both adults and children with severe hearing impairment (Dodd-Murphy & Ritter, 2013; Gifford, Dorman, Shallop, & Sydlowski, 2010). These are simple sentences like, "The apple pie was good." Each sentence has a specific speech-to-noise ratio. The BKB-SIN consists of 1 main talker (target voice/signal) and a 4-person multi-talker babble (background noise/distractors). The score of BKB-SIN reveals the signal to noise ratio of the target voice required for an individual to get the full sentence correct 50% of the time (ETYMOTIC, 2018). The higher the BKB-SIN score, the lower the performance on the auditory test.

In response to a loss of balance forward, a forward compensatory step is required to regain balance. During the first compensatory step, COM is displaced forward, while COP is first displaced posteriorly and laterally under the stance leg (Kanekar & Aruin, 2014; Martin et al., 2002). This dissociates the two variables, creating a distance between COP and COM which eventually will be reconciled when the first step is completed to regain equilibrium and maintain postural stability (Henry, Fung, & Horak, 2001; Horak et al., 2005; Santos, Kanekar, & Aruin, 2010; Winter, Patla, & Frank, 1990; J. Yang, Winter, & Wells, 1990). Maximum COP-COM distance is a stability measure that creates a stability margin for each individual (Jacobs, Dimitrova, Nutt, & Horak, 2005; Kanekar & Aruin, 2014; Papa et al., 2015; Santos et al., 2010; Winter, 1979). Quiet standing analyses of COP-COM suggest good reliability in the Antero-Posterior (AP) direction (Corriveau et al., 2001). Due to the interplay between COP and COM, COP-COM was chosen as an appropriate measure of postural stability during unexpected backward surface translations that force forward compensatory steps to occur.

Data Processing

All kinematic and force data were processed using MATLAB. COP was measured using ground reaction forces from the force plate data and COM was extrapolated using the sacral marker as described in F. Yang and Pai (2014). Baseline values for COP-COM were calculated in the 30 frames before the surface translation (0.25 seconds). The maximum COP-COM distance during the first compensatory step was calculated within the window of time from surface perturbation to completion of the first compensatory step, indicated by placement of the stepping leg heel on the force plate. Both heel marker data and force data confirmed initiation and completion of reactive step. All maximum COP-COM distances were normalized to subject height. Reaction time was determined as the time in milliseconds (ms) from start of surface perturbation to the point in time when the heel marker of the stepping foot transitioned from accelerating backward while receiving the surface translation to accelerating forward in the opposite direction thus initiating a compensatory step. Calculation of reaction time was assessed within the window from surface perturbation initiation to maximum acceleration of the heel marker during first compensatory forward step.

Data Analysis

For each individual, the values for each outcome (i.e. BKB-SIN scores, reaction time, and max COP-COM diff) for each variation set—3 hearing conditions X 3 perturbation levels in a randomized sequence—were averaged, resulting in an average outcome score by combination of conditions. For each outcome, repeated measures ANOVA models were run in Stata 13.1 using time, hearing conditions, and perturbation levels as predictors.

Results

Twenty-three participants participated in Visit 1, however one subject did not pass the screening and two subjects withdrew from the research study. Twenty subjects were included in Visit 2 for dual-task auditory and balance testing. The baseline characteristics of the subjects were the following: 45% of subjects were female and 55% of subjects were male (9 F, 11 M); the mean age of subjects was 73.2±9.1 years; 95% of subjects were white, 0% of subjects were Asian, and 5% of subjects were black (Table 2). Cognitive and sensory screening results were also included (Table 2). Upon review of data, one subject was excluded from analysis as the subject completed less than half of trials due to a fall in the harness and inability to complete Level 2 perturbations. Three subjects had two outlier values for max COP-COM and reaction time more than 3.5 standard deviations from group average; these outlier values were replaced with the value for group mean at that combination of conditions. .

The overall mean BKB-SIN score under the hearing aid condition was 12.5±3.2 dB Hearing Level (HL) and under the no hearing aid condition was 19.5±3.9 dB HL. The mean BKB-SIN score in the hearing aid condition at perturbation Level 0 was 12.3±3.5 dB HL, Level 1 was 13.4±3.1 dB HL, and Level 2 was 11.8±2.9 dB HL. The mean BKB-SIN score in the no hearing aid condition at perturbation Level 0 was 20.6±2.8 dB, at Level 1 was 19.2±3.6 dB HL, and Level 2 was 18.6±5.4 dB HL. There was a significant main effect for hearing condition ($p<0.0001$), but not for perturbation level ($p=0.7804$). There was no significant interaction between hearing condition and perturbation level ($p>0.05$) (Figure 1).

The average maximum COP-COM distance for the no repeat back condition at Level 1 was 6.9±0.8 cm and at Level 2 was 17.4±2.8 cm; for the hearing aid condition at Level 1 was 7.0±0.8 cm and at Level 2 was 17.1±2.8 cm; for the no hearing aid condition at Level 1 was 6.3±1.1 cm and at Level 2 was 17.4±3.4 cm. There was a significant main effect for perturbation level ($p<0.001$), but not for hearing condition ($p= 0.9417$). There was no significant interaction between hearing condition and perturbation level ($p>0.05$) (Figure 2).

The reaction time for the no audio condition at Level 1 was 312±21 ms and at Level 2 was 238±20 ms; for the hearing aid condition at Level 1 was 303±30 ms and at Level 2 was 241±24 ms; for the no hearing aid condition at Level 1 was 314±48 ms and at Level 2 was 248±39 ms. There was a significant main effect for perturbation level ($p<0.001$), but not for hearing condition ($p= 0.1029$). There was no significant interaction between hearing condition and perturbation level ($p>0.05$) (Figure 3).

Discussion

The results suggest subjects performed significantly worse on the BKB-SIN under the no hearing aid condition, subjects' maximum COP-COM distance increased, and reaction time decreased as the balance perturbation level increased. While hearing aids improve speech comprehension in noise, in this study hearing aids did not necessarily improve reactive balance ability for older adults with hearing loss.

To our knowledge, this is the first research study to utilize reactive balance outcome measures of maximum COP-COM distance and reaction time, in conjunction with a standardized audiology test, BKB-SIN. Although few studies have observed maximum COP-COM distance as an outcome measure, our results are comparable to previously reported values whereas maximum COP-COM distance was averaging 3.5cm in static standing balance and 16cm with loss of balance requiring compensatory steps (Horak et al., 2005; Papa et al., 2015). Our values for maximum COP-COM distance during level 1 perturbations are higher than results reported for static stance, (6.7cm vs. 3.5 cm), and to be expected at the 2m/s^2 perturbation (McIlroy & Maki, 1996). Our results for maximum COP-COM distance during level 2 perturbations align very well with reported values from other compensatory steps perturbations (17 cm vs 16 cm) (Papa et al., 2015). Because there was no significant difference in maximum COP-COM distance between conditions of no repeat back, hearing aids, and no hearing aids, these results suggest that older adults with hearing loss maintain the ability to execute compensatory reactive balance stepping strategy and prioritize balance task over the auditory task.

Our results of step-onset reaction time in response to level 1 perturbations averaging ~310 ms indicate slower reaction time compared to values from older adults with normal hearing responding to unexpected similar magnitude perturbations previously reported in the literature, averaging ~250 ms (Patel & Bhatt, 2015). These results are similar to literature values of step-onset time in older adults with balance impairments averaging ~280ms (Brauer, Woollacott, & Shumway-Cook, 2002). Only in response to more challenging perturbations at level 2, the reaction time in older adults with hearing loss was comparable to values reported in the literature (Brauer et al., 2002; Patel & Bhatt, 2015). However, there was no modulation of stepping reaction time as a function of auditory condition and hearing aids did not improve stepping reaction time. These results suggest older adults with hearing loss have slower ability to initiate reactive balance compared to older adults with normal hearing and these balance impairments may be similar to a typical older adult faller.

To our knowledge only one study prior to ours has investigated reactive balance in older adults with and without hearing loss; however, the study used ankle and hip strategy as an outcome measure versus compensatory steps (Bruce et al., 2017). The study, performed by Bruce et al. (2017) determined older adults with hearing loss performed significantly worse on a cognitive task during background noise of babble, but had no difference in reactive balance outcomes. These results suggest dual-task interference with the older adults with hearing loss, who prioritize maintaining balance over cognitive task performance (Bruce et al., 2017). This study, however, did not observe the effect of hearing aids on control of reactive balance. Evidence as to whether hearing aids improve balance control is mixed (Agmon, Lavie, & Doumas, 2017; Negahban & Nassadj, 2017; Vitkovic et al., 2016). Hearing aids have been shown to both improve and worsen COP sway in white noise (Rumalla, Karim, & Hullar, 2015; Vitkovic et al., 2016). Inconsistent results could be due to variable low-level background sounds in different laboratory environments, in which certain sounds could be picked-up more readily by the hearing aid compared to no hearing aid; these certain sounds could create an inadvertent distraction to hearing aid users and confound results (Vitkovic et al., 2016). Therefore, the results of our study may mask the reality that hearing aids may improve balance, particularly during dual-task conditions.

One limitation to our study was not including older adults with normal hearing as a comparison group. Results of studies comparing balance control of older adults with normal hearing to older adults with hearing loss, however, have been mixed. For instance, gait parameters have been found to be worse in some participants with hearing loss, but unaffected in other participants (Edwards et al., 2016; Kamil et al., 2016; Wollesen et al., 2018). These results suggest hearing aids may improve balance in certain scenarios and in certain groups of hearing aid users but future research should be performed to determine which outcome measures and testing conditions are better suited and more sensitive to capture these effects.

Conclusion

As expected, hearing aids improved speech recognition in noise, but did not significantly changed the ability of older adults with hearing loss to initiate and execute the first compensatory step in response to balance perturbations. Further research needs to be done to determine whether a cause and effect relationship exists between hearing loss and balance deficits and, if so, whether hearing aids improve reactive balance control by decreasing listening effort and/or cognitive processing.

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Table 1. The randomization table illustrating the number of trials randomized to different combinations of auditory and surface translations conditions resulting in a total of 64 trails and one set of 30 seconds of quiet stance.

Randomization Table

			Perturbation Level		
			Level 0	*Level 1*	*Level 2*
Auditory Condition	*Repeat Back*	*Hearing Aid*	8	8	8
		No Hearing Aid	8	8	8
	No Repeat Back	*No Audio*	30sec	8	8
			16	24	24

64 conditions
+ 30sec Quiet Stance

Table 2. Average baseline characteristic values and scores among the young and older adults.

Baseline Characteristics	
Number of Participants (n)	20
Age (yrs) (mean ± SD)	73.2 ± 9.1
Height (cm) (mean ± SD)	169.4 ± 9.0
Weight (kg) (mean ± SD)	74.0 ± 16.6
Gender (%)	
Male	55%
Female	45%
Race (%)	
White	95%
Asian	0%
Black	5%
Initial Hearing Loss Diagnosis	52.3 ± 22.5

Cognitive and Sensory Screening	
Cognitive	
MMSE (mean ± SD)	28.6 ± 1.6
Word Span Test (mean ± SD)	5.5 ± 1.7
Auditory	
Cerumen Impaction	Negative
Visual	
Eye Chart (mean score)	20/20
Somatosensory	
Ankle Joint Position (%)	100%
Tuning Fork (L foot) (mean ± SD)	6.5 ± 1.0
Tuning Fork (R foot) (mean ± SD)	6.6 ± 1.1
Vestibular	
Dix-Hallpike Maneuver	Negative signs/symptoms
Dynamic Visual Acuity (mean line difference)	0
Vestibular/Ocular Motor Screening	Negative signs/symptoms
Sensory Integration	
Clinical Test of Sensory Interaction and Balance	6/6 conditions

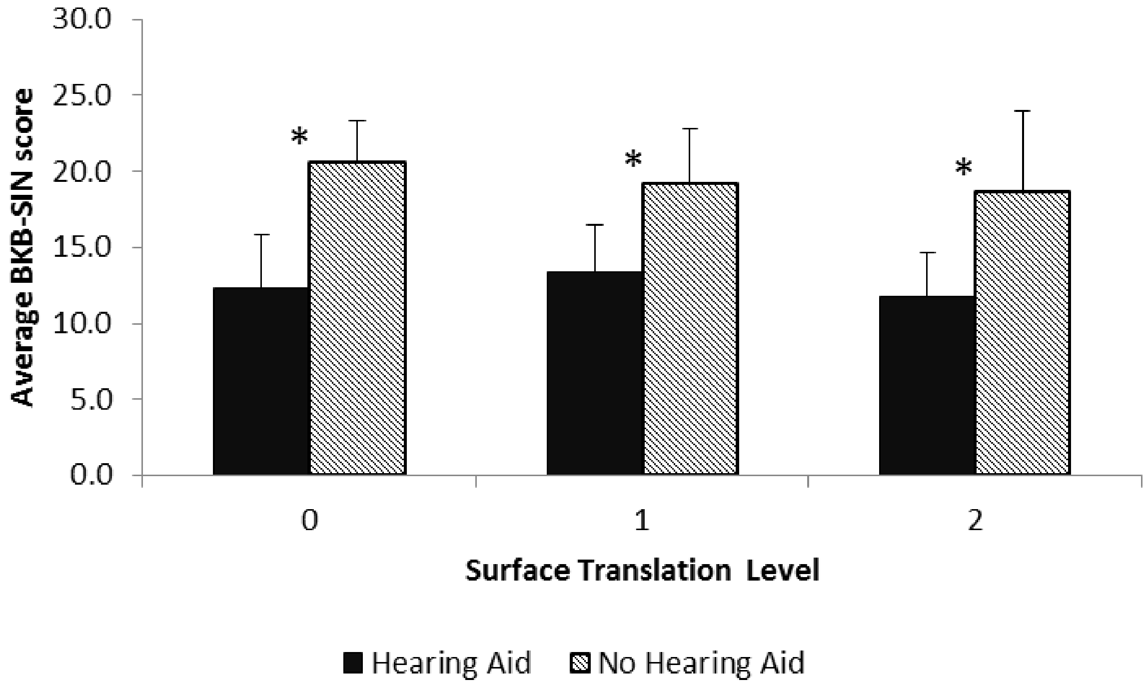

Figure 1. Group averages + stdv for BKB-SIN scores under the Hearing Aid condition (dark bars) and No Hearing Aid condition (hashed bars) at the three levels of surface translations perturbation: Level 0 = no surface translation 0m/s^2, resulting in single task of repeating back; Level 1 = surface translation at 2m/s^2, and Level 2 = surface translation at 5m/s^2. Combinations of Hearing Conditions and Levels 1 and 2 resulted in dual-task conditions of repeating back sentences while maintaining standing balance in response to surface translation perturbations. The higher the score on BKB-SIN, the lower the performance on the auditory task of repeating back sentences.

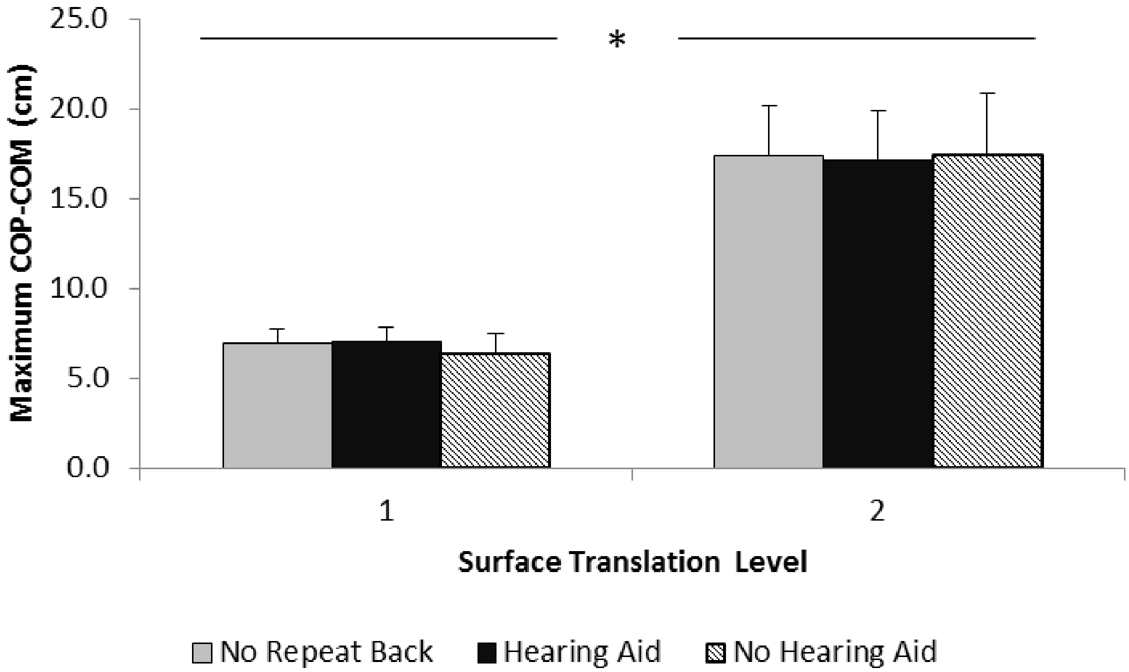

Figure 2. Group averages + stdv of Maximum COP-COM difference (cm) during the first compensatory step in response to surface translations at Level 1 = 2m/s^2 (left bars) and at Level 2 = 5m/s^2 (right bars), under the no repeat back condition (grey bars), hearing aid condition (black bars) and no hearing aid condition (hashed bars).

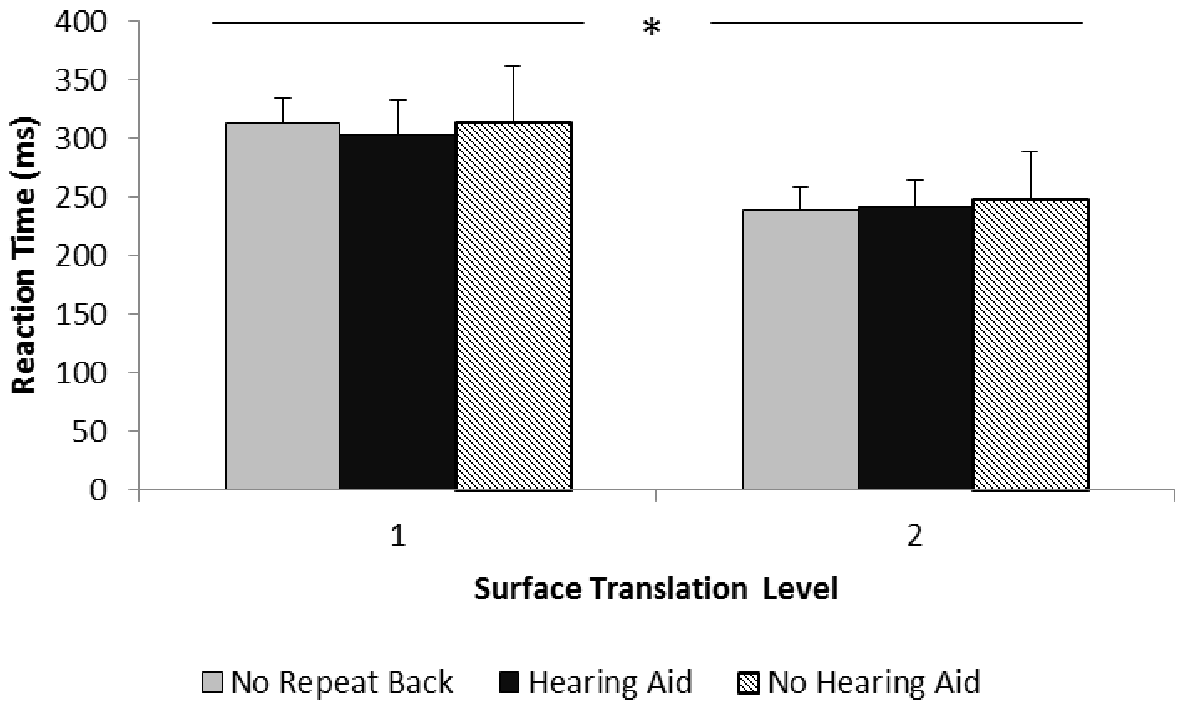

Figure 3. Group averages + stdv of Reaction Time (ms) during the first compensatory step in response to surface translations at Level 1 = $2m/s^2$ (left bars) and at Level 2 = $5m/s^2$ (right bars), under the no repeat back condition (grey bars), hearing aid condition (black bars) and no hearing aid condition (hashed bars).

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CHAPTER 5

HEARING LOSS CONTRIBUTES TO BALANCE DIFFICULTIES IN BOTH YOUNGER AND OLDER ADULTS

INTRODUCTION

Falls are a common problem with adults 65 years and older. One-third of older adults fall annually, costing the United States government approximately 34 billion dollars to cover direct medical expenses for procedures and hospitalization [1, 2]. Falls are not only financially costly; falls also burden families taking care of the older adult, stress the constantly shrinking budget for Medicare, decrease the quality of life for the older adult, and may even lead to death of the older adult [3, 4].

Another common problem plaguing older adults is hearing loss. Age-related hearing loss affects greater than 60% of people aged 70-79 and 80% of those 80 and older [5]. In the USA, hearing loss has expanded at a rate of 160% of the total population growth and continues to grow due to an aging population [6, 7]. Evidence now suggests older adults should address hearing loss because untreated hearing loss may have consequences such as depression, cognitive impairment, and even dementia [8-10].

Moreover, recent evidence has also linked hearing loss to balance deficits in older adults through fall-risk associated assessments, such as slower walking speed and poor Romberg scores [11, 12]. These balance deficits increase when noise is present during balance testing [13]. Although the mentioned evidence highlights the need to identify older adults with hearing loss who are at risk for falling, to our knowledge no study has investigated the impact of hearing loss on ability to regain balance following an unexpected loss of balance. Number of steps is an observable clinical outcome measure that can be used when administering reactive balance tests, such as the Nudge Test, to identify an older adult faller [14, 15]. An increased number of recovery steps after an unexpected loss of balance are associated with an increased risk for falling [16-18].

We aimed to answer two questions: 1) does hearing loss negatively affect the ability to regain balance as reflected by an increased number of steps needed after a perturbation, and 2) do hearing aids reverse this effect and improve balance control, reflected by a decrease in number of steps needed to regain balance.

We hypothesize older adults with hearing loss will take a greater number of steps during an unexpected loss of balance, compared to young adults with normal hearing and older adults with normal hearing.

METHODS

Twenty-five young adults, 33 healthy older adults with normal hearing, and 22 older adults with bilateral hearing loss were verbally informed about the research study and voluntarily agreed to participate through Institutional Review Board (IRB)-approved informed consent. All participants were phone screened prior to enrollment to ensure no visual, vestibular, somatosensory, auditory, health conditions or balance impairments existed that would restrict ability or confound results of the study. Participants were excluded if they had a history of motion sickness/dizziness, or were currently taking medications that affect balance.

Participants underwent cognitive and sensory screening to ensure no undiagnosed cognitive or sensory impairments were present. Five young adults, 13 older adults with normal hearing, and three older adults with hearing loss were excluded due to either undiagnosed cognitive or sensory impairments, or withdrew from the study; resulting in a final count of 20 young adults, 20 older adults with normal hearing, and 19 older adults with hearing loss who participated.

Participants underwent dual-task auditory and balance testing while standing on an instrumented dual-belt treadmill. Participants were required to stand and maintain their balance with unexpected surface translations while simultaneously listening and repeating back sentences at 60 dBA. Dual-task auditory-balance sentences were randomized to control for a learning effect [19-21]. Participants were required to listen and repeat sentences from the standardized audiology outcome measure, the Bamford-Kowal-Bench Speech-In-Noise (BKB-SIN) test [22]. These are simple sentences like "the football player lost a shoe." Each sentence has a specific speech-to-noise ratio, and the scoring on the BKB-SIN outcome measure indicates the required ratio of speech-in-noise for a participant to be able to correctly repeat back 50% of the sentences. The higher the BKB-SIN score, the lower the performance on the auditory test. There were three auditory conditions: 1) no audio sound, no repeat back resulting in the single task of maintaining balance; 2) normal hearing condition in which the BKB-SIN audio files were played, and participants with a diagnosis of hearing loss wore their hearing aids; and 3) hearing loss condition in which the audio files were manipulated to simulate hearing loss for the young and old adults without a hearing loss diagnosis, and participants with a hearing loss diagnosis performed the task without their hearing aids. Participants with hearing loss received the audio input through the speakers, which was delivered directly to the ear via hearing aids. In order to standardize audio input directly to the ear, participants with normal hearing received the audio input to the ear through Bose® QuietComfort 35 wireless headphones.

Backward surface translations were delivered through the treadmill dual-belt system causing the participant to experience a forward loss of balance, while he or she simultaneously listened and repeated the sentence. Three balance conditions were delivered: “0” at $0m/s^2$ and no backward surface translations, resulting in the single task of listening and repeating back the sentence; “1” backward surface translations at acceleration of $2m/s^2$; and “2” backward surface translations at acceleration of $5m/s^2$. The surface translations induced a loss of balance requiring the participants to take 1 or more compensatory steps forward to maintain their balance. An overhead harness system equipped to support up to 181 kg was in place to prevent participants from hitting the ground if a fall would occur.

Combinations of three auditory and three balance conditions were provided randomly and each participant completed 8 trials per combination of auditory-balance conditions.

A 12 camera Motion Analysis System collected kinematic data from 54 reflective markers placed on anatomical landmarks of the body. The V-gait treadmill system by Motek Medical containing 2 separate force plates mounted underneath each belt was used to deliver surface translations and record force data (Figure 1).

The primary outcome measures were number of steps and BKB-SIN scores. Number of steps was recorded using visual observation with Cortex Motion Analysis to verify for any uncertainties. Only 1 fall into the harness occurred during data collection; therefore, steps leading to the fall were counted. The BKB-SIN was scored by a single grader, who wore headphones connected via Bluetooth to a microphone worn by the participant. Participants were encouraged to guess every time they heard babble.

The number of steps and BKB-SIN scores across the 8 trials per combination of auditory-balance conditions were averaged by person, resulting in an average outcome score per combination. For each outcome, repeated measures ANCOVA models were run in Stata 13.1 using group, time, auditory condition, and balance condition as predictors. All independent variables were coded as categorical with the first group entered as the referent group, and time, auditory, and balance conditions designated as repeated measures variables. The between-participants error terms was designated as "ID|time", ID was designated the variable representing the lowest unit in the between-participants error term, and time was designated as the group variable for computing the pooled covariance matrix.

RESULTS

Baseline characteristics for the sample of 59 participants by group are presented in Table 1, which includes: Young adults with normal hearing (YANH), Older adults with normal hearing (OANH) and Older adults with hearing loss (OAHL).

The results suggest that group, auditory condition and balance condition (perturbation level) are significantly related to both outcome measures. There were significant difference in the BKB-SIN score between groups, auditory, and balance conditions (all $p<0.0001$) (Figure 2). There was a significant difference in number of steps between groups, ($p=0.0001$), auditory conditions ($p= 0.0301$), and balance conditions ($p<0.0001$) (Figure 3). In addition, the perturbation level has a greater impact on steps, and auditory condition has a greater impact on BKB-SIN. Time was not significant for number of steps ($p= 0.1828$), but was significant for BKB-SIN ($p=0.0001$), meaning that repeated trials lead to different performances.

DISCUSSION

The results of this study state there are significant differences in BKB-SIN scores and number of steps between young and older adults with normal hearing, and older adults with hearing loss. These results suggest older adults with hearing loss have poorer reactive balance compared to young and older adults with normal hearing. In older adults with normal hearing, simulated hearing loss negatively affects the ability to regain balance as reflected by an increased number of steps needed after a perturbation. However, the balance performance, as measured by the number of steps required to regain balance while wearing hearings aids, may not have significantly improved enough to prevent a fall. This suggests that while, hearing aids are beneficial for speech recognition, their impact in reversing the negative effect and improve balance control is not as easily measured or understood.

These results coincide with the mixed literature regarding hearing loss and balance difficulty among older adults, as well as whether hearing aids improve balance for older adults with hearing loss [23]. Older adults with hearing loss have been shown to have increased sway compared to older adults with normal hearing, and hearing aids have been shown to improve static balance and balance outcome measures such as the Berg Balance Scale (BBS) [12, 24, 25]. Older adults with hearing loss have also been shown to have no difference in performance on physical tasks and outcome measures, such as the Timed-Up-And-Go (TUG), and hearing aids did not improve physical function [26, 27].

One limitation to the study was the size of the treadmill and the harness system. All individuals were limited in the number and direction of steps able to be taken compared to a setting where participants are able to move more freely [28]. Another limitation is the variably of

BKB-SIN scores, particularly among older adults with hearing loss. Some older adults with hearing loss scored close to older adults with normal hearing on the BKB-SIN, while others experienced the floor effect with the BKB-SIN – with higher scores indicating worse performance – and could only attend to a small handful of sentences. The BKB-SIN was designed and is usually administered in a sitting position in a sound-proof booth. The test may have a floor or ceiling effect that has yet to be examined while participants are standing and interacting in a 'real-world' setting [29]. The older adults with hearing loss experiencing the floor effect on the BKB-SIN may not have demonstrated a true listening-auditory dual-task based on their hearing ability and these results could actually mask this population at risk for falling, especially in noisy environments [30]. Lastly, many older adults with hearing loss read lips, but the role of vision on speech perception while performing a balance test was not able to be assessed based on the nature of the BKB-SIN [31, 32].

It is currently unknown how and why older adults with hearing loss fall more often compared to older adults with normal hearing [33]. More research needs to be performed in order to determine reasons behind why older adults with hearing loss fall more often in order to create proper assessment and treatment strategies for older adults with hearing loss who are at risk for falling [34].

CONCLUSION

Older adults with hearing loss appear to require an increased number of steps to regain balance and may be at a greater risk for falling compared to older adults with normal hearing. Number of steps may be an appropriate balance outcome measure to assess fall risk for older adults with hearing.

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Table 1. Baseline characteristics of young adults with normal hearing, older adults with normal hearing, and older adults with hearing loss.

Baseline Characteristics	**YANH**	**OANH**	**OAHL**
Number of Participants (n)	20	20	19
Age (yrs) (mean ± SD)	27.2 ± 3.0	68.7 ± 4.3	73.2 ± 9.1
Height (cm) (mean ± SD)	170.4 ± 8.8	163.6 ± 8.2	169.4 ± 9.0
Weight (kg) (mean ± SD)	74.1 ± 11.5	70.5 ± 19.0	74.0 ± 16.6
Gender (%)			
Male	55%	25%	55%
Female	45%	75%	45%
Race (%)			
White	65%	95%	95%
Asian	25%	5%	0%
Black	10%	0%	5%
Initial HL diagnosis (yrs)			52.3 ± 22.5

YANH = Young Adults with Normal Hearing; OANH = Older Adults with Normal Hearing; OAHL = Older Adults with Hearing Loss

Figure 1. An example image of the research study performed in the laboratory. The participant is standing on a dual-belt treadmill and wearing 54 reflective markers that are being captured by 12 surrounding cameras.

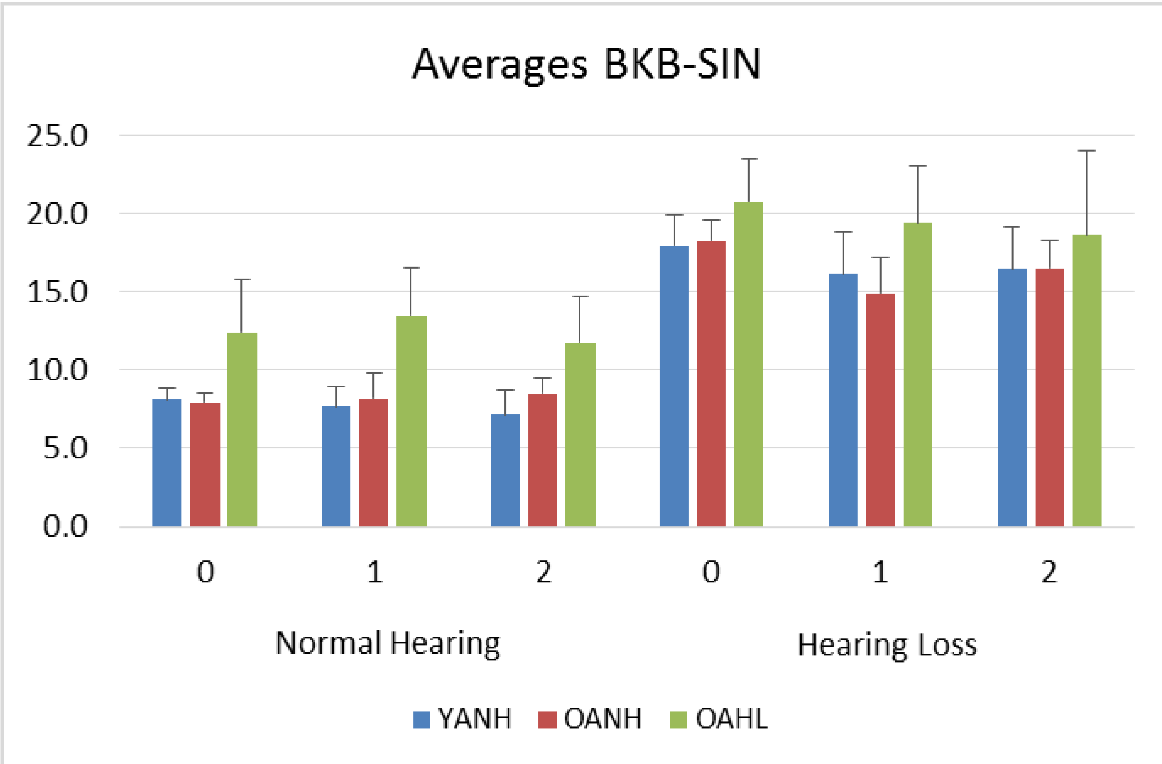

Figure 2. Older adults with hearing loss have significantly higher average BKB-SIN scores, with a higher score indicating worse performance, compared to young adults and older adults with normal hearing during Level 0, 1, and 2 surface translations. All adults perform significantly worse under the hearing loss condition. YANH = Young Adults with Normal Hearing; OANH = Older Adults with Normal Hearing; OAHL = Older Adults with Hearing Loss. Normal Hearing = Normal Hearing/Hearing Aid condition; Hearing Loss = Simulated Hearing Loss/No Hearing Aid condition.

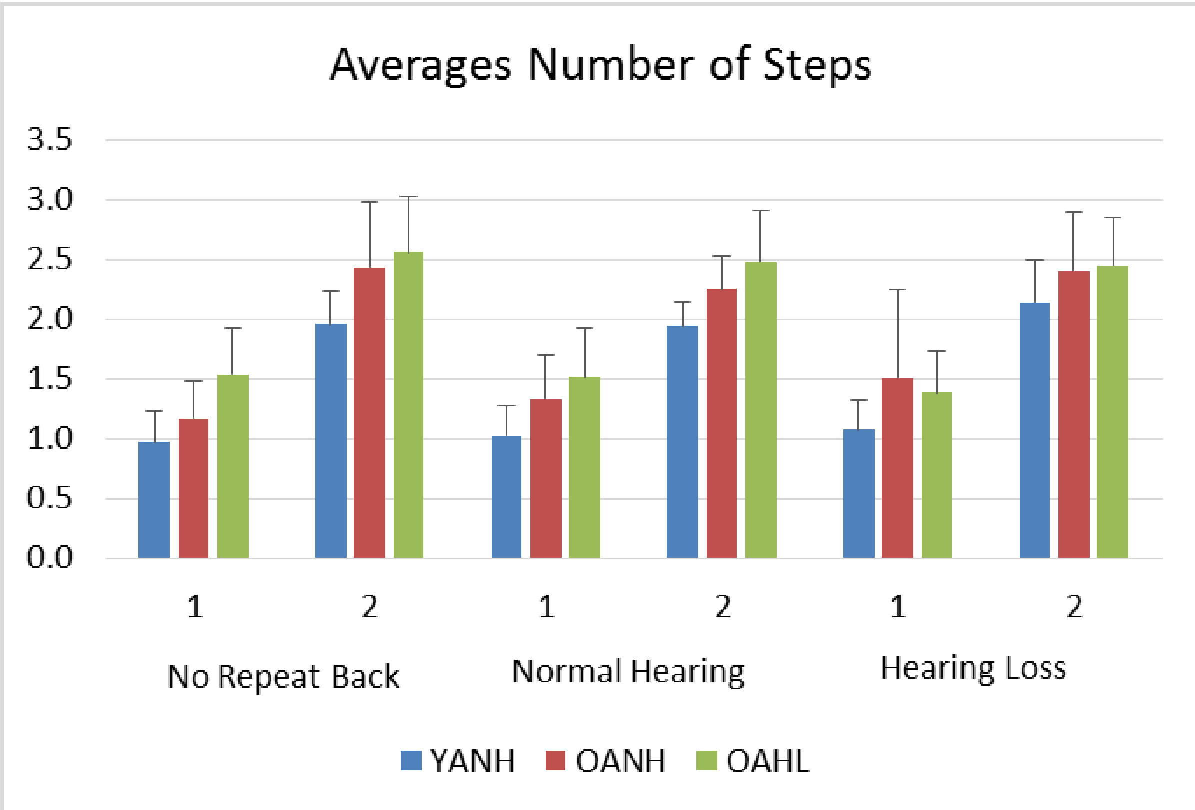

Figure 3. Older adults with hearing loss take a significantly greater number of steps on average compared to young and older adults with normal hearing during no repeat back, normal hearing, and hearing loss conditions, significantly increasing as perturbation level increases from 1 to 2. Number of steps significantly changes across all groups as challenge of task increases from single-task, no repeat back to dual-task condition under normal hearing to dual-task condition under hearing loss. YANH = Young Adults with Normal Hearing; OANH = Older Adults with Normal Hearing; OAHL = Older Adults with Hearing Loss. Normal Hearing = Normal Hearing/Hearing Aid condition; Hearing Loss = Simulated Hearing Loss/No Hearing Aid condition.

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CHAPTER 6

DISCUSSION

We investigated the contribution of auditory input to reactive balance control by simulating hearing loss in healthy young and older adults with normal hearing, and testing older adults with hearing loss with and without their hearing aids. The subjects completed single- and dual- tasks consisting of repeating back sentences from the standardized audiology test Bamford-Kowal-Bench Speech-In-Noise (BKB-SIN) and maintaining standing balance in response to surface translation perturbations. These experiments were innovative in several ways.

A. Innovation

1. *Defining the Relationship between Auditory Inputs and Balance Control*

In order to maintain balance, the brain must effectively integrate inputs from visual, vestibular, proprioceptive, and auditory systems. Of these, the contribution of auditory inputs to balance has received the least attention (Manchester, Woollacott, Zederbauer-Hylton, & Marin, 1989). Hearing loss, prevalent in older population, deprives the system of relevant acoustic information (Kanegaonkar, Amin, & Clarke, 2012). When auditory inputs are reduced or conflicting, the cognitive resources allocated to effortful listening may be increased or the interpretation of movement in the environment may be inaccurate leading to maladaptive balance responses (Bruce et al., 2017).

Most of the findings linking hearing loss and falls came from epidemiological analyses, formalized self-assessment inventories, or correlational analyses comparing different baseline measures of hearing and balance (Lopez et al., 2011; Sihvonen, Era, & Helenius, 2004; Wojszel

& Bień, 2004). While these approaches are certainly informative, they do not provide a clear understanding of cause-and-effect relationship or an explanation for how and why this relationship exists.

This research was innovative, in our opinion, because it mechanistically studied the link between hearing loss and reactive balance deficits using a hypothesis-driven approach to determine if a cognitive mechanism partially or fully explains the link between hearing loss and balance deficits. It also investigated if hearing aids were an appropriate intervention to improve reactive balance for older adults with hearing loss.

2. *Ecological valid testing protocols integrating standardized audiology and balance tests in realistic real-life virtual environments (VE).*

One of the barriers to understanding the impact of hearing loss on balance has been the disconnect between findings obtained in artificial, yet highly-controlled laboratory settings (e.g. hearing tests performed sitting in soundproof booths with little requirement for balance control), and the impact of these results in the real world. While auditory laboratory hearing tests focus on things such as sound localization, audibility and speech intelligibility, perhaps real-world outcome measures such as balance and walking while conducting a conversation are even more meaningful (Wilson, McArdle, & Smith, 2007). This research was innovative because we used measures of balance to assess the impact of hearing loss, and the effect of hearing aids on outcomes that go beyond speech intelligibility in noise. We used a V-gait CAREN system (Computer Assisted Rehabilitation Environment Network). The CAREN software integrates real-time data streams from motion analysis and combines it with virtual reality in a way that the subject is completely immersed in the VE and controls it. We have integrated standardized audiology tests in the VE and systematically manipulated characteristics of the auditory stimuli, as well as the physical and cognitive challenges and measured the effects on balance. Specifically, we measured the noise level in the laboratory and calibrated the sound level in wave files delivering the standardized sentences in the BKB-SIN, a standardized auditory test. This innovative research study now provides a new avenue for healthcare professionals to test hearing loss outside of a sound booth in safe, life-like environments, and to utilize this new testing environment to drive hearing aid development. Furthermore, our research is innovative because it is the first of its kind to simulate hearing moderate hearing loss according to levels documented in the literature (Cruickshanks, Wiley, et al., 1998). Although other studies have attempted to simulate hearing loss, these studies have been performed by audiologist to understand how hearing loss affects individuals and to improve the quality of hearing aids (Adams & Moore, 2009; Hornsby, Johnson, & Picou, 2011; McPherson, McMahon, Wilson, & Copland, 2012). Our research study is the first of its kind to determine the effect of simulated hearing loss on balance control, as well as observe the effects of audiological testing while the individual is standing and performing tasks outside of a soundproof booth.

3. Validation of the COP-COM outcome measure for compensatory stepping reactions

Our research also validated a novel balance control outcome measure, Center of Pressure – Center of Mass (COP-COM). Most of the traditional biomechanical research assesses COP and COM individually (Winter, 1995; Winter, Prince, Frank, Powell, & Zabjek, 1996). Although assessing COP and COM individually provides valuable insight into an individual's ability to control their balance, studying the relationship of COP and COM together provides a more accurate clinical picture of an individual's ability to control his or her balance (Jančová, 2008; Lafond, Duarte, & Prince, 2004). Maximum COP-COM distance during static balance and voluntary step initiation have been previously reported (Corriveau, Hébert, Prince, & Raîche, 2001; Papa, Foreman, & Dibble, 2015). Our current studies have provided a methodology platform for measuring and analyzing the maximum COP-COM distance during the first compensatory stepping reaction in response to balance perturbations. This is not a trivial task given that the surface translation movement producing the perturbation displaces both COP and COM variables. Determining baseline values, thresholds, and windows of data analysis for COP-COM has advanced our knowledge regarding reactive balance and opened the door for researchers and clinicians to better understand, assess, and provide interventions for reactive balance.

B. <u>Results</u>

Overall results of these studies advance our knowledge in several domains.

1. <u>Acute sudden simulated hearing loss did not change reactive balance in young healthy adults.</u>

The young adults performed worse on the BKB-SIN test under the simulated hearing loss condition compared to the normal hearing condition, confirming that hearing loss was successfully simulated. However, the hypothesis that simulated hearing loss will negatively affect balance control in healthy young adults was not confirmed. Young adults responded as expected to the loss of balance induced by increasing levels of perturbations, decreasing their reaction time (initiating stepping response quicker) and increasing the maximum COP-COM distance (allowing larger separation of COP and COM predictive of larger step). These results suggest that healthy, young adults with normal hearing have the cognitive and physical capacity to overcome potential balance difficulty under all conditions, including performing dual tasks with acute, simulated hearing loss.

2. <u>Simulated hearing loss did not change the execution of compensatory step but decreased ability to modulate initiation of compensatory steps as a function of challenge difficulty in older adults.</u>

The hypothesis that simulated hearing loss will negatively affect reactive balance in older adults with normal hearing was only partially confirmed. Young adults and older adults with normal hearing had no significant group differences in BKB-SIN scores, maximum COP-COM distance, and reaction time. However, the ability to shorten reaction time (initiate stepping response faster) as the balance challenge increased was not as robust in older adults compared to young adults. This may explain the increased number of steps taken by older adults to regain balance.

3. <u>Older adults with hearing loss have poorer reactive balance compared to older adults with normal hearing, and require a greater number of steps to regain balance.</u>

When number of steps was assessed between all 3 groups, number of steps was a significant indicator of worse balance in older adults with hearing loss taking the greatest number of steps, followed by older adults with normal hearing, and then by young adults with normal hearing. These results suggest number of steps during a loss of balance may be an appropriate clinical outcome measure to assess balance ability for older adults hearing loss.

Upon further analysis using a three-way ANOVA, some differences in balance outcome measures between older adults with normal hearing and older adults with hearing loss were identified. There was a significant main effect of perturbation level on maximum COP-COM distance ($p<0.001$). It should be noted that although auditory conditions did not reach significance, the shortest COP-COM distances were recorded in older adults with hearing loss (Figure 1). A shorter the COP-COM distance leads to a smaller the compensatory step, which may lead to a greater potential for requiring additional steps to regain balance. For reaction time, there were significant main effects of group ($p<0.001$) and perturbation level ($p<0.001$), as well as a significant interaction between auditory condition and group ($p=0.0183$) (Figure 2). These results suggest that compared to older adults with normal hearing, older adults with hearing loss have worse reactive balance (longer reaction times in initiating a stepping response) both in single task and under dual-task condition when not wearing their hearing aids.

4. <u>Hearing aids did not improve reactive balance in older adults with hearing loss.</u>

Although the simulation of hearing loss was achieved using published values for moderate hearing loss, comparing the results of BKB-SIN across all three groups of participants shows that older adults with diagnosed hearing loss have significantly poorer ability to recognize speech in noise even when wearing their hearing aids (Cruickshanks, Klein, et al., 1998) (Chapter 5, Figure 2). This may explain in part why no significant changes in balance responses were noted between conditions of hearing aid and no hearing aid.

Unfortunately, when the balance responses of older adults with hearing loss are analyzed with and without their hearing aid in absence of comparison to other groups; the significance of auditory condition on balance scores disappears, suggesting that hearing aids may not improve reactive balance for older adults with hearing loss ($p>0.05$). It should be noted the average age of the older adults with hearing loss was 5 years older compared to older adults with normal hearing

(68.7 ± 4.3 vs. 73.2 ± 9.1), but we do not expect these significant differences in balance to be due to age because age-related declines in balance are typically documented every 10 years versus every 5 years (Steffen, Hacker, & Mollinger, 2002).

C. Limitations

Several limitations existed during this research study. One limitation was the inability to control for cognitive changes associated with age-related hearing loss in the young and older adults with sudden, acute simulated hearing loss. Older adults with hearing loss have been seen to have non-global brain changes and possible re-mapping in the brain regions associated with hearing loss, executive function, and memory (Jayakody, Friedland, Eielboom, Martins, & Sohrabi, 2017; Mudar & Husain, 2016). Furthermore, our study only provided audio for older adults with hearing loss to rely on, compared to audiovisual. Older adults with hearing loss appear to rely more heavily on the visual centers of the brain for providing sensory inputs to control balance, particularly when they have experienced hearing loss for an extended period of time (Puschmann & Thiel, 2017). Providing audiovisual for individuals with hearing loss in a dual-task condition may change the results for older adults with hearing loss in the future, particularly those who rely on lip-reading.

Another limitation of our study is recruiting hearing aid users in the community, versus an audiology clinic. The exclusion criteria for hearing loss subjects regarding hearing abilities and hearing aid use could only be done through self-report; therefore, we cannot determine the true extent of hearing aid use or whether these adults had recently had their hearing aids refitted within the past 12 months.

Variability in performance on the BKB-SIN test was another limitation of the study. Variability of BKB-SIN results may be due to each participant's working memory ability (Rudner, Lunner, Behrens, Thorén, & Rönnberg, 2012). Although all participants included in Visit 2 were determined to have appropriate working memory abilities and clinical analysis showed no within-group differences existed, the ability to use working memory and divide attention during a dual-task situation was not assessed prior to dual-task auditory and balance testing. Individuals with higher working memory abilities have been shown to perform better at listening tasks compared to individuals with lower working memory abilities (Ng, Rudner, Lunner, Pedersen, & Ronnberg, 2013). Therefore, increased variability due to differing working memory abilities during listening tasks may have created some non-significant results that inadvertently conceal the true impact of listening tasks with both sudden, acute simulated hearing loss and sensorineural hearing loss.

D. Emerging New Evidence and Future Directions

New literature is emerging in the biomechanical field regarding the relationship between auditory input and postural control since this research protocol and proposal was initially approved. The majority of research regarding hearing loss and balance control has been performed on static standing balance tasks (Rumalla, Karim, & Hullar, 2015). Although research observing standing balance is useful for a greater understanding of how hearing loss may impact control of balance, limited research has been performed to observe reactive balance. One study observing older adults with age-related hearing loss performed a dual-task that included a loss of balance requiring hip or ankle strategy and a listening cognitive task; the results of the study suggest older adults with age-related hearing loss prioritize the loss of balance task over the listening task (Bruce et al., 2017).

Although our laboratory and other laboratories have performed research observing loss of balance during static standing, minimal research has been performed on how older adults with hearing loss maintain balance during dynamic balance tasks, such as walking, and whether hearing aids improve balance control (Bruce et al., 2017). The results of these studies have used varying outcome measures, such as head pitch trunk rotation, gait speed, or gait variability; regardless, the results suggest hearing loss may negatively impact walking ability, and hearing aids may improve walking ability (Kamil et al., 2016; Lau, Pichora-Fuller, Li, Singh, & Campos, 2016; Wollesen et al., 2018). While these studies are useful, no studies to our knowledge have observed reactive balance of older adults with hearing loss during walking (ie. slips or trips) in combination with cognitive tasks and/or background noise, or whether hearing aids improve balance recovery during a loss of balance requiring compensatory steps during walking. Future research observing how older adults with hearing loss regain balance, particularly while simultaneously performing cognitive or listening tasks in noise, may shed light in how older adults with hearing loss lose their balance and whether hearing aids improve balance in a 'real-world' setting.

Recent literature regarding hearing loss and balance control has also focused less on determining a mechanistic cause, but rather on the notion of sensory reweighting. Typically, when the visual, vestibular, or somatosensory system encounter inaccurate or unexpected inputs (ie. uneven surface, dim room, spinning ride), the entire sensory system reweights the sensory inputs to neglect the inaccurate input in order to maintain control of balance (Allison, Kiemel, & Jeka, 2006). The effect of auditory input has recently been assessed through sensory reweighting clinical outcome measures observing COP sway (Viljanen et al., 2009; Vitkovic, Le, Lee, & Clark, 2016). Young or middle-aged adults with normal hearing have been found to have increased COP sway while performing the modified Clinical Test of Sensory Integration and Balance (mCTSIB) while wearing ear protectors to deafen sound or while standing in a sound-proof room (Kanegaonkar et al., 2012; Maheu, Sharp, Landry, & Champoux, 2017), which suggests some form of sensory reweighting may be occurring for individuals with normal hearing; however, hearing does not have as great of an impact as vision or somatosensory (Vitkovic et al., 2016). The evidence of sensory reweighting for individuals with hearing loss suggests older adults with hearing loss cannot utilize auditory cues the way individuals with normal hearing can, as shown by COP sway that does not change based on the hearing condition; but the evidence is mixed whether hearing aids improve overall COP sway by providing auditory cues in noisy environments (Negahban & Nassadj, 2017; Vitkovic et al., 2016).

Further supporting the notion of sensory reweighting is functional Magnetic Resonance Imaging (fMRI) evidence of an increased baseline and task-dependent cross-talk between the auditory center of the brain and the visual center of the brain. These results suggest the brain may perform remapping to accommodate for gradual loss of auditory signal with sensorineual hearing loss (Puschmann & Thiel, 2017). Additional changes found in brain imaging studies, not associated with sensory reweighting, suggests decreased activation of brain regions associated with memory, learning, and executive function (Jayakody et al., 2017; Mudar & Husain, 2016). Therefore, the brain may remap away from cognitive regions of the brain and toward additional sensory areas of the brain to compensate for the loss of auditory signaling.

Currently, the evidence is unclear as to how individuals with hearing loss use auditory cues and how much audiovisual perception occurs during balance tasks, particularly during loss of balance (Kanegaonkar et al., 2012; Puschmann & Thiel, 2017). Further research should be performed determining the effect of sensory input on sensory reweighting on individuals with hearing loss.

Recent evidence has also arisen on potential holistic treatment interventions for individuals with hearing loss and potential cognitive and/or balance deficits. Individuals with hearing loss could experience a patient-centered auditory rehabilitation program, which would likely include: 1) Amplification (ie. hearing aids); 2) Auditory training (ie. improving hearing speech-in-noise), 3) Cognitive training (ie. various working memory tests), and 4) Social engagement (ie. socializing while using hearing aid) (Mudar & Husain, 2016). A randomized control trail is currently underway to determine the effects of exercise in combination with auditory rehabilitation; however, the results have not been published yet to our knowledge (Lambert et al., 2017). These interventions may not only improve quality of life, but may reduce cognitive and functional decline associated with age-related hearing loss.

Although treatment interventions are being created to address older adults with hearing loss and cognitive or balance deficits, minimal research with varying outcome measures is being performed to our knowledge to determine appropriate assessment strategies for older adults with hearing loss and balance deficits (Agmon, Lavie, & Doumas, 2017). The majority of studies assessing static balance in older adults with hearing loss have used assessments, such as the mCTSIB or the Romberg (Negahban & Nassadj, 2017; Rumalla et al., 2015). The majority of the studies utilizing dynamic balance in older adults with hearing loss have using varying walking parameters, such as gait speed or the Dynamic Gait Index (DGI) (Bruce et al., 2017; Kamil et al., 2016; Lau et al., 2016; Li, Simonsick, Ferrucci, & Lin, 2013; Weaver, Shayman, & Hullar, 2017). It should be noted, however, no clinical assessment tool has yielded consistent, significant results to our knowledge that could be used as a ‘Gold Standard’ to identify older adults with hearing loss who have balance deficits, or to identify if a hearing aid improves balance deficits. Further research should be performed to determine whether an outcome measure can be used or created as a ‘Gold Standard’ that will consistently identify older adults with hearing loss and balance deficits.

E. Conclusion

We investigated how auditory input affects balance control by testing twenty healthy young adults with normal hearing, twenty older adults with normal hearing, and twenty older adults with hearing loss and hearing aids. The normal hearing subjects performed testing under randomized normal hearing and simulated hearing loss conditions, while older adults with hearing loss performed testing under hearing aid and no hearing aid conditions. The subjects completed single- and dual- tasks consisting of a standardized audiology test from a standardized audiological test, the Bamford-Kowal-Bench Speech-In-Noise (BKB-SIN) and maintaining standing balance in response to surface translation perturbations. Participants performed an auditory task of repeating back sentences from the BKB-SIN either played through wireless noise-cancelling headphones for the normal hearing subjects or played through the surround-sound speakers for the hearing loss subjects. Simulated hearing loss was achieved using Adobe Audition software and a FFT logarithmic curve in which sound volume and frequencies of standardized sentences were manipulated according to age-related moderate hearing loss documented in literature (Cruickshanks, Wiley, et al., 1998). Backward surface translation perturbations inducing a forward loss of balance were synchronized with the auditory task and presented randomly at three surface translation perturbation levels: Level 0 = $0 m/s^2$, Level 1 =

$2 m/s^2$, and Level 2 = $5 m/s^2$. Primary outcome measures included: maximum Center of Pressure – Center of Mass (COP-COM) distance in response to perturbation during the first compensatory step, reaction time for initiating the first compensatory step, number of steps after loss of balance, and performance on the BKB-SIN. Repeated measures ANOVA, three-way ANOVA, or an ANCOVA were conducted for each dependent variable with respect to perturbation level and auditory condition. Results show reaction time decreases, maximum COP-COM distance increases, and number of steps increases as perturbation level increases across all groups. BKB-SIN scores and reaction time were significantly worse under the simulated hearing loss condition, but no significant differences existed between the normal hearing groups. Older adults with hearing loss had significantly worse BKB-SIN scores, significantly worse reaction time, and significantly greater average number of steps. Hearing aids significantly improved BKB-SIN scores, but not balance scores. Hearing loss negatively affects reactive balance, particularly while simultaneously performing auditory tasks. Hearing aids may or may not improve balance control during loss of balance. Older adults with hearing loss have an increased risk of falling compared to individuals with normal hearing due to non- age-related cognitive and neurodegenerative decline associated with hearing loss (Mudar & Husain, 2016). Further research needs to be performed to determine how any why older adults with hearing loss are at a higher fall risk, whether dual-task scenarios increase fall risk for an older adult with hearing loss, and whether hearing aids improve balance control.

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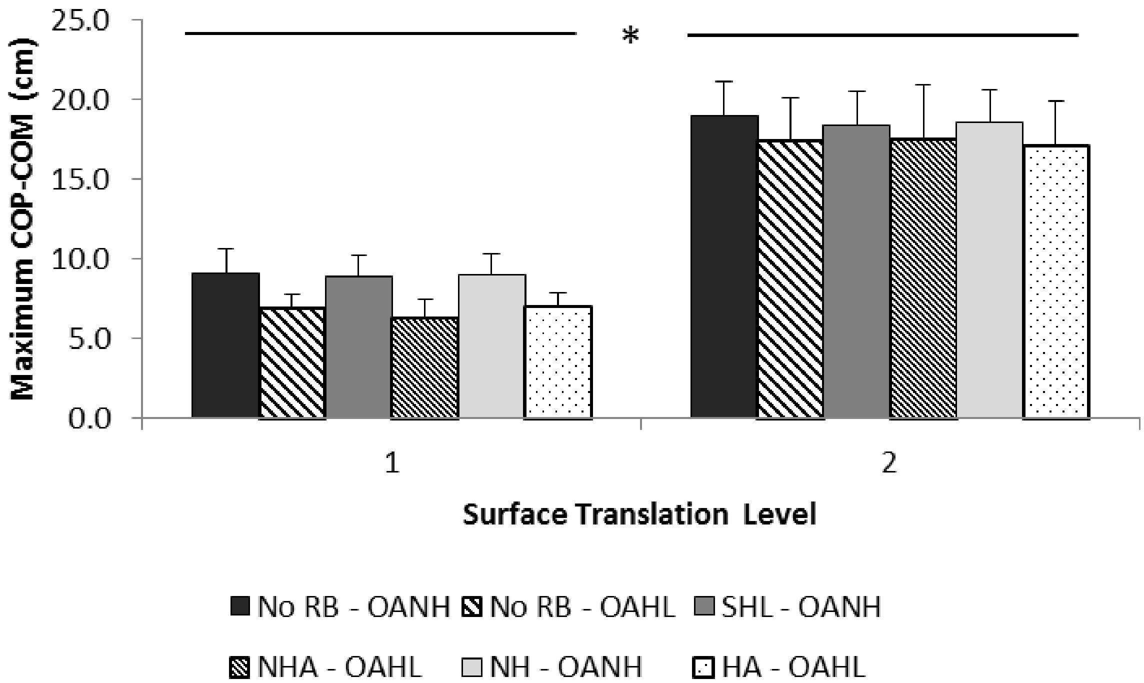

Figure 1. Group averages + stdv of Maximum COP-COM difference (cm) during the first compensatory step in response to surface translations at Level 1 = $2m/s^2$ (left bars) and at Level 2 = $5m/s^2$ (right bars), under the no repeat back condition (black/large hashed bars), normal hearing condition or hearing aid condition (dark grey/medium hashed bars) and simulated hearing loss condition or no hearing aid condition (light grey/small dotted bars). No RB = No Repeat Back/Audio condition, NH = Normal Hearing condition, SHL = Simulated Hearing Loss condition, HA = Hearing Aid condition, NHA = No Hearing Aid condition, OANH = Older Adult Normal Hearing, OAHL = Older Adult Hearing Loss.

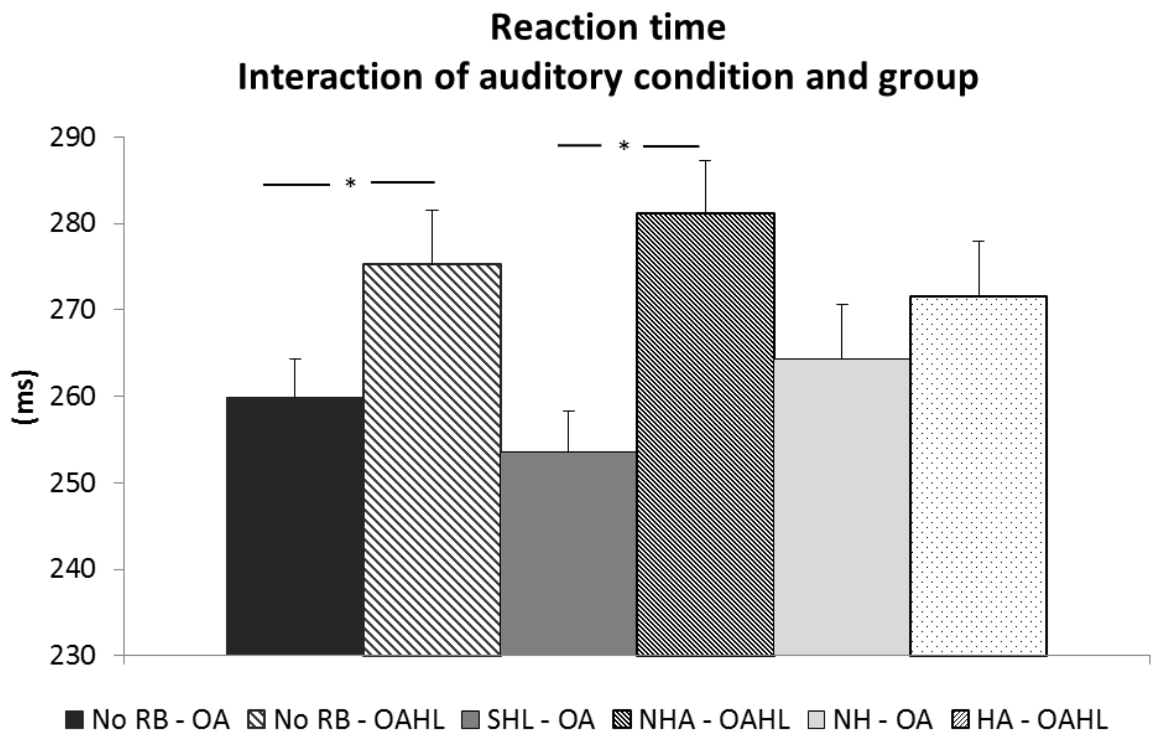

Figure 2. Group averages + stdv of Reaction Time (ms) during the first compensatory step in response to all surface translations, under the no repeat back condition (black/large hashed bars), hearing loss condition either simulated or no hearing aid condition (dark grey and small hashed bars), and normal hearing condition or hearing aid condition (light grey and small dotted bars). No RB = No Repeat Back/Audio condition, NH = Normal Hearing condition, SHL = Simulated Hearing Loss condition, HA = Hearing Aid condition, NHA = No Hearing Aid condition, OA = Older Adult Normal Hearing, OAHL = Older Adult Hearing Loss.

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