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Effects of cognitive load and prosthetic liner on volitional response times to vibrotactile feedback

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Abstract—Artificial tactile feedback systems can improve prosthetic function for people with amputation by substituting for lost proprioception in the missing limb. However, limited data exists to guide the design and application of these systems for mobility and balance scenarios. The purpose of this study was to evaluate the performance of a noninvasive artificial sensory feedback (ASF) system on lower-limb function in the presence of a cognitive load and a liner interface. Reaction times (RTs) and accuracy of leg-movement responses to vibratory stimuli at the thigh were recorded for 12 nondisabled individuals and 3 participants with transfemoral amputation using a custom-built testing apparatus. The results indicate that the addition of a cognitive load increases response times relative to the baseline condition by 0.26 to 0.33 s. The prosthetic liner produced a less pronounced increase in RT of 0.06 to 0.11 s. Participants were able to correctly identify the stimulus location with nearly 100% accuracy. These increased RTs are nontrivial and must be considered in designing ASF systems.

Key words: amputation, biofeedback, cognitive load, lower-limb amputation, mobility, proprioception, sensorimotor responses, sensory feedback, transfemoral, vibration.

INTRODUCTION

Individuals with lower-limb amputation (LLA) commonly use an artificial limb or prosthetic device to at least partially restore function and enable the completion of everyday tasks. However, even state-of-the-art prostheses do not effectively compensate for the sensory deprivation from the missing limb, including information about the loading and position of the residual limb. While some sensory feedback does come through kinetic interactions at the prosthetic socket and the residuum interface, individuals with LLA have to rely on alternate senses, such as sight and hearing, to safely and efficiently perform ordinary mobility tasks [1–2]. Sensory feedback from the lower limbs is integral to maintaining upright gait [3–4], as well as executing complex tasks such as navigating around obstacles [5]. Reduced sensory input is at least partially responsible for the balance and mobility impairments recognized in the amputee population, including abnormal and less efficient gait kinematics [6–8] and increased risk of loss of balance resulting in falls [4]. Therefore, mitigating these impairments and enabling an acceptable level of functional mobility for individuals with LLA is a primary goal in prosthetic rehabilitation. One approach in pursuit of this goal has targeted ways to provide additional artificial sensory information to the user as a substitute for the lost proprioception in the missing limb [9].

Abbreviations: ASF = artificial sensory feedback, BMI = body mass index, LLA = lower-limb amputation, RT = reaction time.

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While most research into artificial sensory feedback (ASF) systems has involved the upper limbs, a number of recent studies have begun to examine the application of such systems to lower limbs to improve balance and mobility. The studies have identified important design requirements for such systems and even demonstrated their potential to improve certain aspects of balance in individuals with LLA [4,9–10]. Specifically, these studies have explored the efficacy of different mechanisms for delivering artificial sensory information, assessing the influence of factors such as stimulus location and stimulus type on both the sensitivity to stimulus and time to voluntary muscle response [11–12]. The time delay between stimulus detection at the mechanoreceptors and voluntary muscle action is especially important for mobility applications, such as gait, where reactions to external stimuli must fall within a small time window to ensure continuous fluid motion. Reaction times (RTs) to signals from ASF systems have previously been explored, but these studies are by no means exhaustive and several important factors remain to be investigated. One particularly interesting factor with relevance to everyday mobility is the presence of cognitive distractions that can alter an individual's ability to accept and utilize sensory information in a timely manner.

Multitasking, or being cognitively occupied, has been shown to impede and distract individuals from achieving simple everyday tasks [13–14]. Williams et al. describes how ambulating with a prosthesis often requires greater cognitive attention in challenging conditions, which may consist of walking on uneven terrain, conversing while walking, or negotiating crowded environments [13]. In contrast to nondisabled adults, who typically manage such conditions with ease, ambulating with a transfemoral prosthesis requires significant cognitive effort because of the loss of proprioception and motor control at the ankle and knee joints, which diminishes the normal motor and balance mechanisms relative to nondisabled individuals [13,15–16]. The literature also describes how people with amputation must rely on visual cues in order to monitor the prosthetic device, which interferes with their ability to perform other tasks and is considered an additional cognitive burden [13,17–19]. For example, persons with LLA are often required to physically look at their prosthesis when descending stairs to prevent themselves from tripping. While cognitive loads clearly play an important role in the mobility of individuals with LLA, the specific effects of these distractions on their ability to utilize sensory information, and consequently the potential performance of ASF systems in everyday situations, remains to be investigated.

A second practical point of interest when considering the application of an ASF system in a prosthesis is the effect of the interface of the prosthesis and the residuum where the stimuli would most likely be applied. Compliant prosthetic liners are commonly prescribed to individuals with LLA to provide cushioning, safety, and comfort to the residual limb within the prosthetic socket [20–21], as well as to prevent ulcerations and other skin conditions [21]. Since these liners are commonly used by people with LLA [22], it is important to understand the influence of these liners on the mechanical transmission of sensory information. Specifically, one would expect that the presence of a liner between the stimulators within the prosthetic socket and receptors in the residual limb would result in potential signal attenuation and therefore a decreased performance of ASF systems.

The goal of this study was to inform the design of an ASF system to compensate for lost mechanoreception in individuals with LLA, as well as potentially other mobility-impaired patient populations. Specifically, we focus on assessing the effects of a cognitive load and prosthetic liner on the ability of a noninvasive ASF system to elicit timely and correct movements of the lower limbs, vis-a-vis the RTs and accuracy of responses associated with the stimulation of the lower-limbs. We hypothesize that adding a cognitive load will increase the time and decrease the accuracy of voluntary responses to stimuli and that adding a prosthetic liner between the skin and stimulator will have a similar effect. Furthermore, we examine how stimulation location and frequency affect these outcome measures for each condition to help guide the design of a potential ASF system to minimize any potential cognitive load or liner-induced detriments on RT and accuracy.

METHODS

Participants

The study involved a convenience sample of 12 nondisabled individuals and 3 individuals with transfemoral amputations. Participants were included if they had no affiliated health issues including neurological disease, diabetes, or peripheral vascular disease and were free from burns, scars, unhealed wounds, blisters, or skin problems on their lower limbs. The nondisabled participants included five females and seven males with a mean age of 27 ± 2 yr, weight of 69.5 ± 17.31 kg, height of 1.72 ± 0.11 m, and body mass index (BMI) of 23.3 ± 1.2. Participant 1 was a 20 yr old male, with a weight of 73 kg, height of 1.78 m, and BMI of 22.9. Participant 2 was a 22 yr old male, with a weight of 86 kg, height of 1.78 m, and BMI of 27.2. Participant 3 was a 35 yr old male, with a weight of 60 kg, height of 1.65 m, and BMI of 22.0. Participants 1 and 2 had congenital amputations, while participant 3 had a trauma-induced amputation. Time since amputations was 18, 20, and 4 yr, respectively, for the three participants. The study was approved by the Holland Bloorview Kids Rehabilitation Hospital Research Ethics Board, and informed written consent was obtained from each participant before commencing.

Equipment

Sensory feedback was provided via a vibratory stimulus using vibrotactile motors [9]. The motors (Model 310–101, Precision Microdrives, Inc; London, United Kingdom) employ a rotating offset mass that when attached to the body produce shear forces in plane with the skin surface and proportional to the speed (frequency) of the motor. Each motor was calibrated to operate at the vibratory frequencies required for the study: 140, 180, and 220 Hz [12]. The Arduino Mega 2560 processor from Sparkfun Inc (Niwot, Colorado) was used with an N-type MOSFET (metal–oxide–semiconductor field-effect transistor) to control the vibrotactile motors.

A custom-built testing frame was fabricated as shown in **Figure 1(a)**, consisting of a rigid platform to support the participants' legs and push buttons (Buddy Button Gator, Bridges Canada Inc; Mississauga, Canada) to record the RTs of response movements. The height of the push buttons was adjustable to the height of the participant's knee, and soft sponges were used to cushion the impact with the push buttons. For participants with amputation, the push buttons were height-adjusted to keep the leg level.

Protocol

The age, height, and weight of all participants were recorded. Since testing was done on the lower limb, all participants wore shorts. Testing was performed on the dominant and amputated legs of the nondisabled and amputee participants, respectively. It should be noted that data presented in this study were collected as a part of another study described in a previous article exploring the effects of vibration frequency and location on RTs and were collected

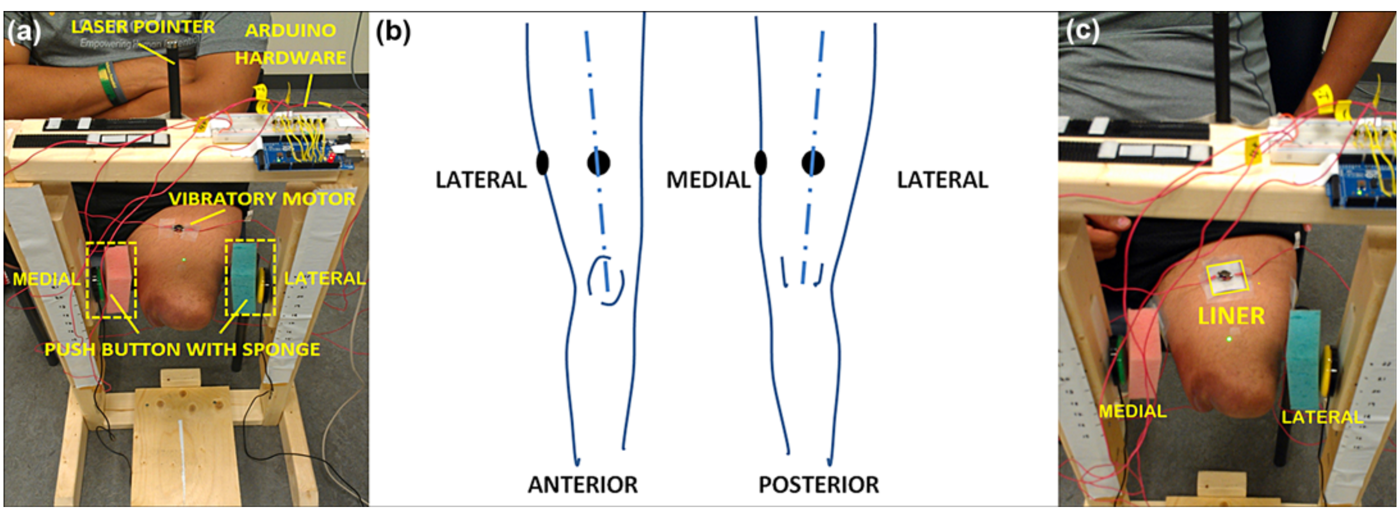

Figure 1.
(a) Custom platform was constructed to support participant's residual limb and house hardware required for experiments, including push buttons and microprocessor. Experiment 1 utilized only lateral button, while Experiment 2 required both medial and lateral. **(b)** Thigh portion of leg (right leg shown) showing motor placement for Experiments 1 and 2. Motors were placed on anterior, posterior, lateral, and medial sides of thigh for Experiment 1 and only lateral and medial sides for Experiment 2. **(c)** Placement of silicone gel liner on individual with transfemoral amputation showing motor mounted on surface of liner. In nondisabled individuals, liner was simply wrapped around surface of thigh.

using similar experimental protocols [12]. The following experiments were performed.

Experiment 1

Four motors were placed around the circumference of the thigh, located approximately halfway down the length of the anterior, posterior, medial, and lateral sides, as seen in **Figure 1(b)**; the arrangements were similar to those used previously by Wentink et al. [11] and Rusaw et al. [4]. The motors were programmed to operate at one of three vibratory frequencies: 140, 180, or 220 Hz. In each trial, one of the three frequencies was applied randomly to one of the four locations, and participants were instructed to push a single button by laterally moving the leg as quickly as possible in response to the vibratory stimulus (lateral button position in **Figure 1(b)**). A laser pointer was used to align and center the limb before each trial. The experiment was conducted under three conditions: (1) baseline control condition, with the motors attached directly to the participant's skin; (2) liner condition, with a 3 mm-thick silicone gel socket liner placed between the motor and the participant's skin (**Figure 1(c)**); and (3) cognitive load condition, with the cognitive task of counting down out loud by seven from a randomly assigned number (with no liner). This cognitive task has previously been demonstrated to effectively preoccupy and distract subjects during postural control [22]. A total of 36 random trials were conducted for each of the three conditions (3 frequencies × 4 locations repeated 3 times).

Experiment 2

The second experiment utilized the same setup as in Experiment 1, but with the anterior and posterior motors removed, so that only the medial and lateral sides of the participant's thigh could be stimulated. Participants were asked to respond as quickly as possible to vibratory stimuli on either the lateral or medial side of the leg by using the same leg to push the button corresponding to the side being excited. For example, if the medial side was stimulated, the participant would move the leg to press the button on the medial side. Two stimulation frequencies, 140 and 220 Hz, were tested at both locations for a total of 40 random trials (2 frequencies × 2 locations repeated 10 times). Response times and selections (medial or lateral button) were recorded for each trial. The experiment was conducted under the same control, cognitive load, and liner conditions as in Experiment 1.

Data Analysis

Data (RTs) were captured using a customized LabVIEW (National Instruments Corp; Austin, Texas) program interfaced with the Arduino and exported as ASCII text files to Microsoft Excel (Microsoft Corp; Redmond, Washington) and restructured to SPSS format. To examine differences among the three tested conditions, a repeated measures analysis of variance was performed in IBM SPSS version 20 (IBM; Armonk, New York) for each of the experiments for the nondisabled participants. Statistical significance was set at an alpha level of 0.05 for all primary analyses and pairwise comparisons, with Bonferroni adjustment used to identify significance in the dependent variable test conditions. Intrasubject variables were defined for the test conditions: baseline control, liner, and cognitive load; motor location: anterior, lateral, posterior, and medial regions; and excitation frequency:140, 180, and 220 Hz. Interactions between the test condition and frequency, as well as the test condition and stimulus location were also examined to further isolate the effects of frequency and location, respectively within condition. The results of Experiment 1 agree with previous research [12] showing that RTs from the 180 Hz stimulation consistently fall between RTs from the 140 Hz and 220 Hz frequencies. Therefore, Experiment 2 included only these latter two frequencies for statistical analysis. Descriptive statistics were used for the amputee data.

RESULTS

For the nondisabled group in Experiment 1, all three test conditions were found to significantly affect RTs ($p = 0.002$, $F = 12.985$, $df = 2$). The mean RT was longest for the cognitive task condition (1.048 ± 0.075 s), followed by the liner (0.853 ± 0.044 s) and baseline (0.712 ± 0.032 s) conditions, as shown in **Figure 2(a)**. Stimulation location, however, did not significantly affect the RTs in either the cognitive load or liner conditions ($p = 0.37$, $F = 1.195$, $df = 3$), as shown in **Figure 3(a)**. Frequency did exhibit a significant overall effect on RTs ($p < 0.001$, $F = 109.893$, $df = 2$), with the 220 Hz stimulation resulting in the shortest RT within each condition (**Figure 3(b)**). The amputee population showed a similar trend as the nondisabled population, with the cognitive condition resulting in longer RTs than the baseline.

For the nondisabled group in Experiment 2, the experimental conditions significantly affected RTs ($p < 0.001$,

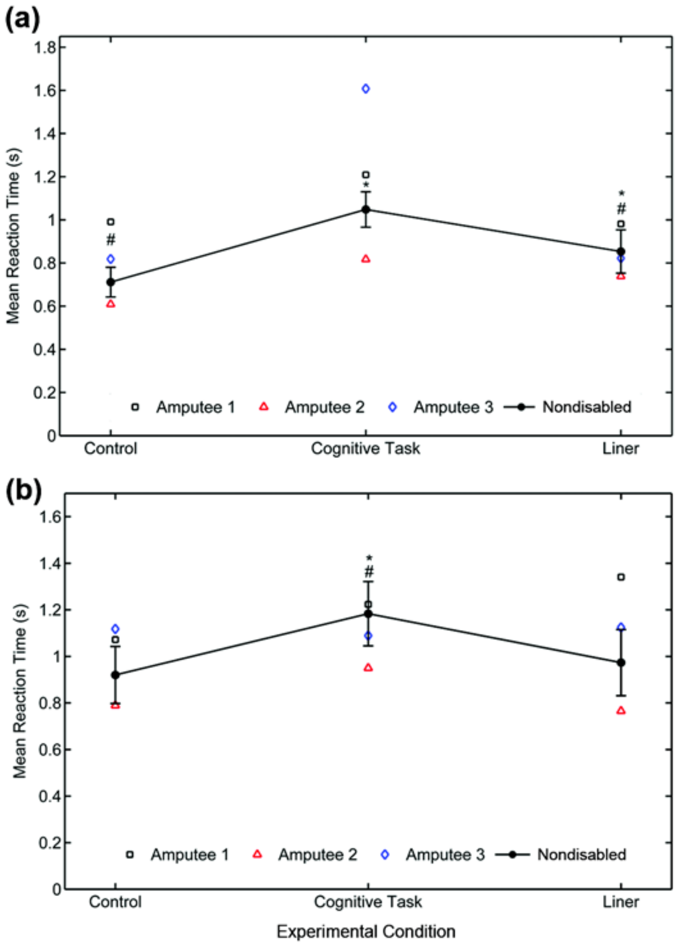

Figure 2.
(a) Overall mean reaction times for each condition of Experiment 1. *Statistically significant differences from Control group at $p < 0.05$. #Statistically significant difference from Cognitive group at $p < 0.05$. Raw data from each participant with amputation are shown as individual datapoints. **(b)** Overall mean reaction times for each condition of Experiment 2. *Statistically significant difference from Control group at $p < 0.05$. #Statistically significant difference from Liner group at $p < 0.05$. Raw data from each amputee are shown as individual datapoints.

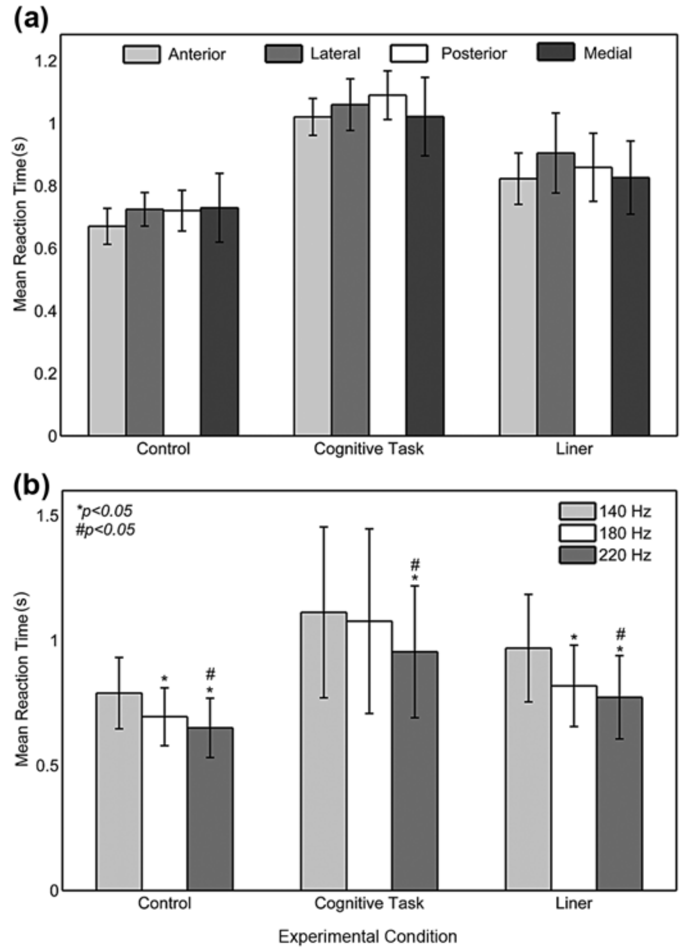

Figure 3.
Mean reaction times for Experiment 1 are shown for each condition, frequency, and location. **(a)** Intracondition effects of location, showing no statistical differences within each condition. **(b)** Intracondition effects of stimulation frequency. *Statistically significant difference from 140 Hz stimulation at $p < 0.05$. #Statistically significant difference from 180 Hz stimulation at $p < 0.05$.

$F = 16.986$, $df = 2$), as shown in **Figure 2(b)**. The cognitive condition (1.813 ± 0.081 s) showed longer RTs compared to the baseline (0.920 ± 0.050 s) ($p = 0.001$). The RTs for liner condition (0.973 ± 0.040 s) were found to be nonsignificant in comparison to the baseline condition ($p = 0.11$), but were significantly shorter compared to the cognitive condition ($p = 0.02$). Location did not have a significant overall effect on RTs ($p = 0.23$, $F = 1.618$, $df = 1$), but the interaction between location and experimental condition ($p = 0.02$, $F = 5.879$, $df = 2$) indicates stimulation location may be preferentially important in some experimental conditions. Specifically, the RTs in the liner condition were longer for medial stimulation than for lateral stimulation. Frequency exhibited a significant main effect on RTs ($p < 0.001$, $F = 30.209$, $df = 1$), but the interaction between frequency and experimental conditions did not show a significant impact on RT ($p = 0.26$, $F = 1.549$, $df = 2$). The latter result suggests frequency affects RT similarly for all three conditions. Participants reacted more quickly to the 220 Hz (0.914 ± 0.043 s) than the 140 Hz frequency

(1.140 ± 0.068 s, $p < 0.001$). Overall, participants averaged 100 percent accuracy in the baseline condition for choosing which side the stimulus originated from. Nondisabled participants averaged 99.38 percent accuracy during the liner condition, and 98.33 percent in the cognitive condition. Analysis showed that neither location nor frequency had any effect on accuracy ($p > 0.15$, $F = 2.295$, $df = 2$). Unlike Experiment 1, the participants with amputation showed a high level of variability in RTs, with no clear trend compared with the nondisabled population. Response accuracies were 100 percent for all three participants with amputation in all three conditions. **Figure 4** shows the results for Experiment 2.

DISCUSSION

The goal of this study was to examine several potential factors affecting the design and performance of an ASF system. Building on previous work, which investigated variables relating to the stimulator type and location [12], the focus of this work was to specifically examine the effects of a cognitive load and the presence of a liner on volitional responses to feedback stimuli. Overall, the findings suggest that both conditions do have an effect on increasing RTs.

Previous work using similar protocols as in this study determined that lower-limb response times to stimuli provided on the leg were upwards of 0.6 s. In this study, as anticipated, both the addition of a liner and cognitive load were found to increase the response times, with the cognitive load having a much more pronounced effect than the liner. Specifically, the cognitive load increased response times on average by about to 0.26 to 0.33 s (29%–47%) from the baseline for the nondisabled group, compared with a 0.06 to 0.11 s (6%–15%) increase with the liner. Although there was much more variability in the data, similar trends were seen for the subjects with amputation. The increase in RTs with accompanying cognitive load agrees with previous investigations, which found that balance and mobility tasks performed under a high cognitive load took significantly longer than the same tasks performed without the distractions [23–24]. In their review of postural control in young adults and attentional demands, Woollacott and Shumway-Cook found no significant difference in the postural sway during a cognitive task. In another dual-task paradigm review, they found that RTs slowed during walking when compared with sitting for young and older adults [24]. Ultimately, they concluded that postural control is attentionally demanding and this demand increases with the complexity of the postural task being performed. Fundamentally, understanding these distraction-related delayed RTs is important in testing the usability and effectiveness of ASF systems, and we therefore recommend that dual-task protocols continue to be employed as part of ongoing and future research.

In addition to cognitive loads, the smaller contribution from the liner is likely due to a subtle attenuating

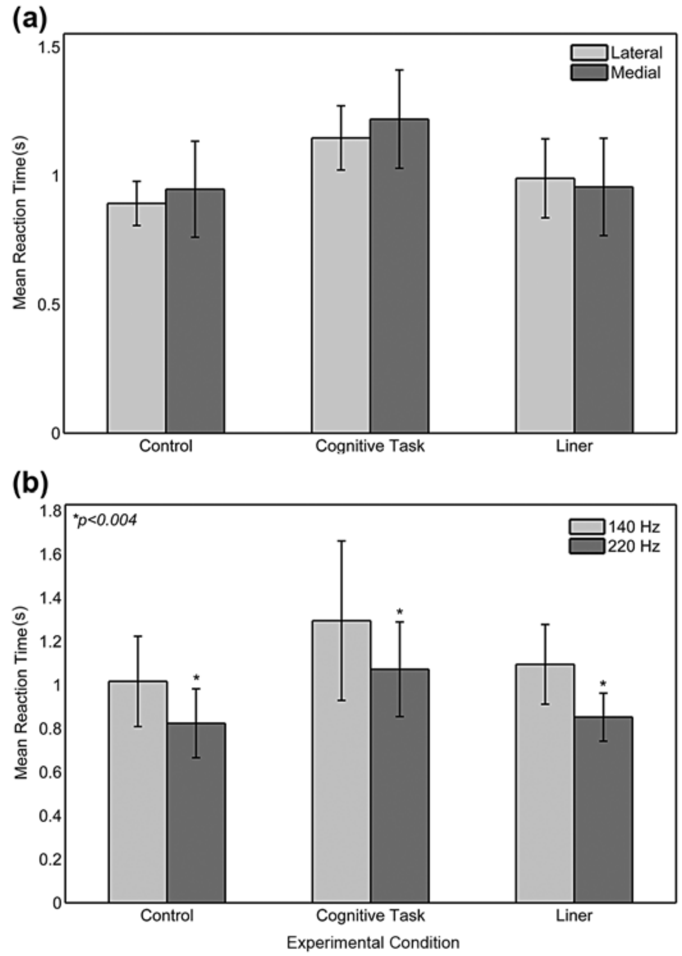

Figure 4.
Mean reaction times for Experiment 2 are shown for each condition, frequency, and location. **(a)** Intracondition effects of location, showing no statistical differences within each conditions. **(b)** Mean reaction times for each frequency and condition, showing shorter reaction times for 220 Hz frequencies within each of three conditions. *Intraconditional significant difference from 140 Hz reaction times ($p < 0.004$).

effect on the vibrations produced by the motor as they pass through the material. This effect is comparable to reducing the vibration amplitude, which has previously been shown to increase response times [12]. It should be noted that the vibration frequency and amplitude generated by motors used in this study are highly coupled, with higher amplitudes accompanying increases in frequency. Therefore, it is unclear whether the significant decrease in response times at 220 Hz relative to 140 Hz is due to the increased amplitude or frequency or a combination of these parameters. Since higher frequencies are attenuated more than lower frequencies as they pass through soft materials, it is possible that lower frequencies with higher amplitudes would be less affected by the addition of a liner. However, the different frequencies affected RTs fairly proportionally for both the liner and control conditions, suggesting that the liner-induced increase in RTs is a result of the damped vibration amplitude rather than frequency-based attenuation.

This work provides insights into important factors to be considered in the development of noninvasive ASF systems for lower-limb applications; however, further work remains to determine how such systems may be best utilized. Given the relatively long RTs that exist with and without additional cognitive loads, incorporating information from these ASF systems in mobility applications requiring real-time decision making may present additional challenges. For example, with a typical gait cycle lasting approximately 1.0 s [25], the RTs presented in this study may be too long to make timely adjustments to the gait cycle based on the sensory feedback. Conversely, postural sway during standing requires adjustments on the order of 0.5 Hz [26], which may be sufficiently slow to incorporate the information from an ASF system. ASF applications to balance and postural stability are particularly interesting because of the combination of volitional and nonvolitional movements required to maintain the body upright. Previous research has demonstrated the potential for ASF systems to improve postural stability in both nondisabled and impaired adults during standing [27–30]; however, few studies have explored the effects of these systems on balance in people with LLA. Similarly, little work has been done to examine how ASF affects balance during dynamic mobility conditions, such as walking, in which both voluntary and involuntary muscle control is required. Future work will focus on evaluating the performance of ASF systems in improving balance in people with LLA under both static and dynamic mobility conditions.

LIMITATIONS

The primary limitation of this study arises from the relatively small sample size, especially of the participants with amputation, who evidenced a large degree of variability in age and BMI, with both congenital and acquired amputations. The specific effects of each of these parameters could not be examined with the small sample size. The static tests performed in these experiments pose an additional limitation. Real-world applications of ASF systems would be used in dynamic environments that produce additional vibrations and noise that could partially mask the stimuli from the motors, potentially further increasing RTs. Recent work by Crea et. al. showed that vibratory feedback could be detected at different phases of the gait cycle, as well as changes to the stimuli pattern, including missing and delayed signals [31]. These results suggest at least some of this dynamic noise can be overcome. However, future studies exploring users' responses to vibratory stimuli in the presence of noise would better guide the application and design of ASF systems. Finally, while care was taken to randomize the combinations of stimulus frequency and location, the trials for each condition (control, liner, cognitive task) were performed sequentially. While this ordering may have introduced some systematic error, randomization of the conditions would require complete detachment, repositioning, and reattachment of the motors between every trial, potentially adding further variability and significantly increasing the testing time required for each participant.

CONCLUSIONS

The efficacy of ASF in improving gait and posture is dependent on the time required for individuals to process feedback information and adjust their gait or posture accordingly. Our results indicate that cognitive loads, and to a lesser degree the stimulator-skin interface, increase this RT and therefore need to be considered in the design of the ASF systems. Future studies should build on this work to develop and evaluate ASF systems under the intended applications conditions, including mobility and balance tasks.

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Analysis and interpretation of data: A. Sharma, M. J. Leineweber, J. Andrysek.
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REFERENCES

1. Edelstein JE, Moroz A. Chapter 10. Transfemoral gait analysis. Lower-limb prosthetics and orthotics: Clinical concepts. Thorofare (NJ): Slack Incorporated; 2011. p. 71–78.
2. Gilbert JA, Maxwell GM, George RT Jr, McElhaney JH. Technical note—auditory feedback of knee angle for amputees. Prosthet Orthot Int. 1982;6(2):103–4. [PMID:7110916]
3. Dozza M, Horak FB, Chiari L. Auditory biofeedback substitutes for loss of sensory information in maintaining stance. Exp Brain Res. 2007;178(1):37–48. [PMID:17021893] http://dx.doi.org/10.1007/s00221-006-0709-y
4. Rusaw D, Hagberg K, Nolan L, Ramstrand N. Can vibratory feedback be used to improve postural stability in persons with transtibial limb loss? J Rehabil Res Dev. 2012; 49(8):1239–54. [PMID:23341316] http://dx.doi.org/10.1682/JRRD.2011.05.0088
5. Arieta AH, Dermitzakis K, Damian D. Sensory feedback for body awareness in prosthetic applications. The Neuromorphic Engineer [Internet]. Institute of Neuromorphic Engineering; 2010. Available from: http://www.ine-news.org/pdf/002880/002880.pdf
6. Göktepe AS, Cakir B, Yilmaz B, Yazicioglu K. Energy expenditure of walking with prostheses: Comparison of three amputation levels. Prosthet Orthot Int. 2010;34(1): 31–36. [PMID:20196687] http://dx.doi.org/10.3109/03093640903433928
7. Miller WC, Speechley M, Deathe AB. Balance confidence among people with lower-limb amputations. Phys Ther. 2002;82(9):856–65. [PMID:12201800]
8. Nolan L, Wit A, Dudziñski K, Lees A, Lake M, Wychowañski M. Adjustments in gait symmetry with walking speed in trans-femoral and trans-tibial amputees. Gait Posture. 2003;17(2):142–51. [PMID:12633775] http://dx.doi.org/10.1016/S0966-6362(02)00066-8
9. Fan RE, Culjat MO, King CH, Franco ML, Boryk R, Bisley JW, Dutson E, Grundfest WS. A haptic feedback system for lower-limb prostheses. IEEE Trans Neural Syst Rehabil Eng. 2008;16(3):270–77. [PMID:18586606] http://dx.doi.org/10.1109/TNSRE.2008.920075
10. Fan RE, Wottawa C, Mulgaonkar A, Boryk RJ, Sander TC, Wyatt MP, Dutson E, Grundfest WS, Culjat MO. Pilot testing of a haptic feedback rehabilitation system on a lower-limb amputee. Proceedings of the International Conference on Complex Medical Engineering; 2009 Apr 9–11; Tempe, AZ. p. 1–4.
11. Wentink EC, Mulder A, Rietman JS, Veltink PH. Vibrotactile stimulation of the upper leg: Effects of location, stimulation method and habituation. Conf Proc IEEE Eng Med Biol Soc. 2011;2011:1668–71. [PMID:22254645] http://dx.doi.org/10.1109/IEMBS.2011.6090480
12. Sharma A, Torres-Moreno R, Zabjek K, Andrysek J. Toward an artificial sensory feedback system for prosthetic mobility rehabilitation: Examination of sensorimotor responses. J Rehabil Res Dev. 2014;51(6):907–17. [PMID:25356723] http://dx.doi.org/10.1682/JRRD.2013.07.0164
13. Williams RM, Turner AP, Orendurff M, Segal AD, Klute GK, Pecoraro J, Czerniecki J. Does having a computerized prosthetic knee influence cognitive performance during amputee walking? Arch Phys Med Rehabil. 2006;87(7): 989–94. [PMID:16813788] http://dx.doi.org/10.1016/j.apmr.2006.03.006
14. Pinzur MS, Gold J, Schwartz D, Gross N. Energy demands for walking in dysvascular amputees as related to the level of amputation. Orthopedics. 1992;15(9):1033–36, discussion 1036–37. [PMID:1437862]
15. Dault MC, Geurts AC, Mulder TW, Duysens J. Postural control and cognitive task performance in healthy participants while balancing on different support-surface configurations. Gait Posture. 2001;14(3):248–55. [PMID:11600328] http://dx.doi.org/10.1016/S0966-6362(01)00130-8
16. Lajoie Y, Barbeau H, Hamelin M. Attentional requirements of walking in spinal cord injured patients compared to normal subjects. Spinal Cord. 1999;37(4):245–50. [PMID:10338343] http://dx.doi.org/10.1038/sj.sc.3100810
17. Pellecchia GL. Postural sway increases with attentional demands of concurrent cognitive task. Gait Posture. 2003;

18(1):29–34. [PMID:12855298] http://dx.doi.org/10.1016/S0966-6362(02)00138-8
18. Fernie GR, Holliday PJ. Postural sway in amputees and normal subjects. J Bone Joint Surg Am. 1978;60(7):895–98. [PMID:701337]
19. Heller BW, Datta D, Howitt J. A pilot study comparing the cognitive demand of walking for transfemoral amputees using the Intelligent Prosthesis with that using conventionally damped knees. Clin Rehabil. 2000;14(5):518–22. [PMID:11043877] http://dx.doi.org/10.1191/0269215500cr345oa
20. Boutwell E, Stine R, Hansen A, Tucker K, Gard S. Effect of prosthetic gel liner thickness on gait biomechanics and pressure distribution within the transtibial socket. J Rehabil Res Dev. 2012;49(2):227–40. [PMID:22773525] http://dx.doi.org/10.1682/JRRD.2010.06.0121
21. Klute GK, Glaister BC, Berge JS. Prosthetic liners for lower limb amputees: A review of the literature. Prosthet Orthot Int. 2010;34(2):146–53. [PMID:20384553] http://dx.doi.org/10.3109/03093641003645528
22. Andersson G, Hagman J, Talianzadeh R, Svedberg A, Larsen HC. Effect of cognitive load on postural control. Brain Res Bull. 2002;58(1):135–39. [PMID:12121823] http://dx.doi.org/10.1016/S0361-9230(02)00770-0
23. Weast RA, Neiman NG. The effect of cognitive load and meaning on selective attention. In: Ohlsson S, Catrambone R, editors. COGSCI 2010: Cognition in Flux—Proceedings of the 32nd Annual Meeting of the Cognitive Science Society; 2010 Aug 11–14; Portland, OR. Austin (TX): Cognitive Science Society; 2010. p. 1477–82.
24. Woollacott M, Shumway-Cook A. Attention and the control of posture and gait: A review of an emerging area of research. Gait Posture. 2002;16(1):1–14. [PMID:12127181] http://dx.doi.org/10.1016/S0966-6362(01)00156-4
25. Levine D, Richards J, Whittle MW. Whittle's gait analysis. 5th ed. Edinburg (UK): Churchill Livingstone Elsevier; 2012.
26. Freitas SM, Wieczorek SA, Marchetti PH, Duarte M. Age-related changes in human postural control of prolonged standing. Gait Posture. 2005;22(4):322–30. [PMID:16274914] http://dx.doi.org/10.1016/j.gaitpost.2004.11.001
27. Goodworth AD, Wall C 3rd, Peterka RJ. Influence of feedback parameters on performance of a vibrotactile balance prosthesis. IEEE Trans Neural Syst Rehabil Eng. 2009;17(4):397–408. [PMID:19497820] http://dx.doi.org/10.1109/TNSRE.2009.2023309
28. Goodworth AD, Wall C, Peterka RJ. A balance control model predicts how vestibular loss subjects benefit from a vibrotactile balance prosthesis. Conf Proc IEEE Eng Med Biol Soc. 2011;2011:1306–9. [PMID:22254556] http://dx.doi.org/10.1109/IEMBS.2011.6090307
29. Wall C. Vestibular prostheses: Engineering and biomedical issues. ASHA Lead. 2008;13(9):14–17. http://dx.doi.org/10.1044/leader.FTR2.13092007.14
30. Priplata AA, Niemi JB, Harry JD, Lipsitz LA, Collins JJ. Vibrating insoles and balance control in elderly people. Lancet. 2003;362(9390):1123–24. [PMID:14550702] http://dx.doi.org/10.1016/S0140-6736(03)14470-4
31. Crea S, Cipriani C, Donati M, Carrozza MC, Vitiello N. Providing time-discrete gait information by wearable feedback apparatus for lower-limb amputees: Usability and functional validation. IEEE Trans Neural Syst Rehabil Eng. 2015;23(2):250–57. [PMID:25373108] http://dx.doi.org/10.1109/TNSRE.2014.2365548

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