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Research

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Improving Postural Stability among Amputees by Tactile Sensory Substitution

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ABSTRACT

Background

For lower-limb amputees, wearing a prosthetic limb helps restore their motor abilities for daily activities. However, the prosthesis's potential benefits are hindered by limited somatosensory feedback from the affected limb and its prosthesis. Previous studies have examined various sensory substitution systems to alleviate this problem; the prominent approach is to convert foot-ground interaction to tactile stimulations. However, positive outcomes for improving amputees' postural stability are still rare. We hypothesize that the intuitive design of tactile signals based on psychophysics shall enhance the feasibility and utility of real-time sensory substitution for lower-limb amputees.

Methods

We designed a wearable device consisting of four pressure sensors and two vibrators and tested it among the unilateral transtibial amputees (n=7) and the able-bodied (n=8). The real-time measurements of foot pressure were fused into a single representation of foot-ground interaction force, which was encoded by varying vibration intensity of the two vibrators attached to the participants' forearm. The layout of vibrators was spatially congruent with the foot force sensors' placement; the vibration intensity followed a logarithmic function of the force representation, in keeping with principles of tactile psychophysics. The participants

were tested with a classical postural stability task in which visual disturbances perturbed their quiet standing.

Results

With a brief familiarization of the system, the participants exhibited better posture stability against visual disturbances when switching on sensory substitution than without. The body sway was substantially reduced, as shown in head movements and excursions of the center of pressure. The improvement was present for both amputees and able-bodied controls and was particularly pronounced in more challenging conditions with larger visual disturbances.

Conclusions

Substituting otherwise-missing foot pressure feedback with vibrotactile signals can improve postural stability for lower-limb amputees. The intuitive design of the mapping between the foot-ground interaction force and the tactile signals is essential for the user to utilize the surrogated tactile signals for postural control, especially for situations that their postural control is challenged.

KEYWORDS: sensory substitution, amputees, postural control, intelligent prosthesis

Introduction

For amputees, wearing a prosthetic limb can help restore their motor functions and improve life quality. For fluent and adaptive motor performance, the nervous system employs a close sensorimotor loop where efferent motor outputs are continuously coupled with afferent sensory feedback (1). The development of typical lower-limb prosthetics, even those robotic prosthetics with actuation, focuses on the efferent control, i.e., controllability and usability of the prosthetic limb without providing the missing sensory feedback caused by amputation (2). Studies on intelligent lower-limb prosthesis have made impressive progress in adaptive control of the knee and ankle joints for walking (3-7) and even used electromyography of residual limb muscles to adjust the force or torque of prosthetic joints (8-10). Essentially, these studies aimed to realize fluent control of the robotic prosthetics with efficiency and precision. However, supplying suitable afferent feedback for lower-limb prosthesis users is still understudied.

The lower-limb amputee lacks direct foot contact with the ground and the feedback from foot mechanoreceptors, critical for balance control (11). With a broken sensorimotor loop, amputees often show poor balance and gait function with fear of falling and a high prevalence of falls (12, 13). When an amputee wears a prosthesis, the residue limb of the amputee physically interacts with the prosthetic sockets and provides limited haptic feedback that indirectly reflects foot-ground interaction. Augmenting this essential feedback for prosthesis wearers has the potential to close the sensorimotor control loop and subsequently improve their gait control and postural stability (14, 15).

Sensory substitution is to encode the missing sensory information and route it to the nervous system via an alternative, intact sensory channels. For example, auditory and haptic feedback has been used to surrogate visual feedback for the blinded to explore the surroundings (16). For upper-limb amputees, sensory substitution has been shown to provide effective sensory feedback for controlling robotic arms (17). Previous researchers have also explored the coding of movement-related information via visual, auditory, or tactile channels for lower-limb amputees. For example, Zambarbieri, Schmid (18) used a pressure-sensing insole to estimate the center of pressure (CoP) underneath the foot and visually present the estimate to the participant. This method is apparently impractical since the processing of the surrogated visual information is cognitively demanding and thus limits the benefit of sensory substitution for gait and postural control, which are typically controlled with minimal cognitive load. Other researchers have also used auditory feedback to deliver gait balance information and demonstrated a positive effect on gait asymmetry (19, 20). However, the auditory solutions are also impractical given their high demands on cognitive resources and their surroundings' quietness. Thus, it is understandable that most researchers have turned to tactile sensory substitution for prosthetic control. The tactile feedback is typically delivered by electrotactile stimulation (21, 22) or vibrotactile stimulation (23-27), the latter being the more favorable one for amputees since it is more comfortable to wear (28).

However, the potential benefits of tactile sensory substitution for lower-limb amputees have not been firmly established. Fan, Culjat (24) developed a tactile device consisting of four pneumatically controlled balloon actuators that pressed against the residue thigh of the

amputated leg with a force magnitude linearly scaled by the pressure measurements from the insole of the prosthesis. They found that, based on the data from a single transtibial amputee, the intensity and the order of pressing forces applied by the balloon actuators could be estimated with decent accuracy (24, 25). However, they did not assess the efficacy of the system in any motor task with prosthesis use. Furthermore, the large size of the balloon actuators might prevent its wide use in the amputee population. Plauché, Villarreal (29) and Crea, Cipriani (23) used similar instrumented insoles but applied electrotactile vibrations on the thigh to inform the amputee about the phase transitions of gait. However, these studies only tested the device on non-disabled participants to show its feasibility and efficacy. The only study that actually examined the postural balance in amputees with tactile sensory substitution returned mixed or little beneficial results (26). This study again placed four vibrators on the thigh to applied tactile stimuli contingent on the measurement of four plantar pressure sensors placed in the insole. The vibration intensity changed in proportion to the amount of plantar pressure. Three separate tasks were used to assess its effect on postural balance, including quiet standing, reaching to a visual target with a cursor representation of CoP, and continuously tracking an oscillatory target with the CoP cursor. Among dozens of performance variables, only the reaction time of the CoP reaching task showed improvement with sensory substitution among amputees. In fact, the mediolateral range of CoP movements, negatively correlated with postural stability during quiet standing, increased with sensory substitution. In sum, previous researches on lower-limb amputees either did not examine the

effect of tactile sensory substitution on balance performance or failed to provide a convincing beneficial effect.

These findings appear discouraging for the application of tactile sensory substitution in lower-limb amputees. However, recent studies have shown that foot-ground contact feedback delivered by directly stimulating the afferent nerves in the residuum of transtibial amputees can improve their postural stability and gait (30, 31). Furthermore, extra tactile feedback also has been shown to improve postural control among vestibular patients (32-34) and patients with Parkinson's disease (34). We thus hypothesize that proper design of the vibrotactile system can enhance standing balance among lower-limb amputees. Previous approaches can be improved in at least two technical aspects. First, the spatial correspondence between the foot's missing sensation and the surrogate tactile signal shall be intuitive to the prosthesis user. For instance, most studies measured plantar pressure at four locations underneath the foot and mapped it onto vibrotactile stimulations applied at four locations on the thigh (23, 26, 29). However, the spatial layout of the vibrators was around the thigh. It is conceivable that the motor system needs considerable training before incorporating the spatially-incongruent tactile information into the sensorimotor control loop. However, none of those above studies provided any dedicated training session. The solution is either giving participants extensive training with the device, or making the vibrotactile stimulus more intuitive to learn, or both. Second, previous studies typically encoded tactile stimulation as a linear function of the magnitude of plantar pressure. However, human tactile perception is a nonlinear function of stimulus amplitude, i.e., perceptual discrimination of changes

deteriorates with stimulus intensity (35, 36). Thus, a high-intensity tactile stimulus is less informative. Currently, this nonlinearity in tactile perception has not been taken into consideration to enhance the efficacy of sensory substitution. One of our previous studies also confirmed that amputees have more difficulty distinguishing the intensities of tactile stimuli than locating them on the skin (37).

In the present study, we designed an intuitive tactile stimulation system to provide real-time feedback on plantar pressure. We tested its efficacy in improving postural stability among amputees and the non-disabled. We measured plantar pressure at four insole locations and mapped it nonlinearly to tactile intensity. Critically, to make the learning of sensory substitution easy and intuitive, our system only encodes CoP excursions in the anteroposterior direction, a more critical direction of instability among amputees than other directions (38). Thus, we only needed to use two vibrators and aligned them on the forearm's longitudinal axis, which corresponds to the anteroposterior body sway. Previous studies usually used quiet standing or treadmill walking when comparing postural control ability between lower-limb amputees and non-disabled participants (38-41). We similarly used quiet standing but examined its stability under visual disturbance with the classical moving-room paradigm (42). We found that after minimal training with the system, both amputees and the non-disabled improved their postural stability under visual disturbances. The improvement was particularly large when the visual disturbances are more challenging.

Methods

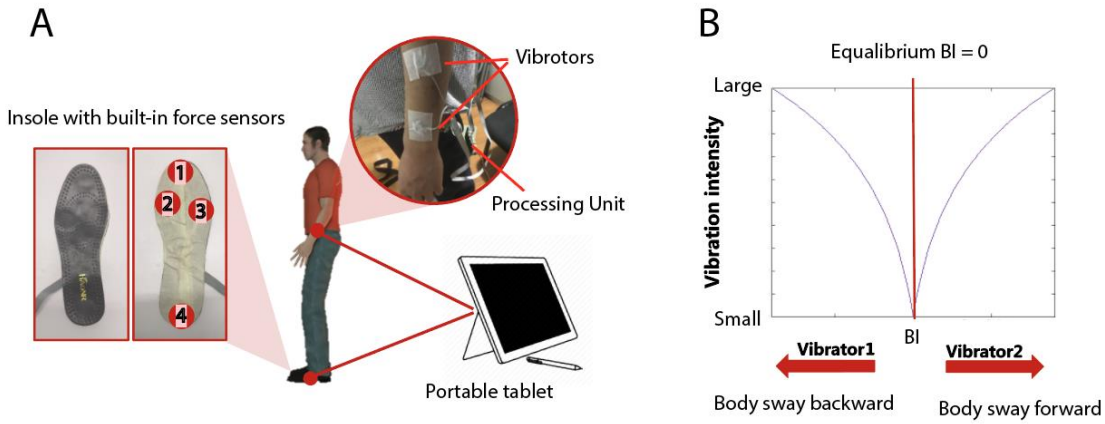
The hardware of the sensory substitution system

This study designed a sensory substitution device consisting of four electropiezo force sensors (*FlexiForce* A401, Tekscan, Inc.) and two miniaturized vibrators. We instrumented an insole with the force sensors at four critical locations, including the areas under the calcaneus tuberosity, the fourth metatarsal, the first metatarsal, and the hallux (Figure 1A). One of our previous researches has found that the force readings from these four locations can capture most of the data variance in plantar pressure during walking (8). Since the feet size varied among participants, we customized the shape of the insole for each individual participant. The sensor was circular with a diameter of 2.54 cm and a thickness of merely 0.208 mm. The response time of the sensor was less than 5 μs with a sampling rate of 100 Hz.

The vibrotactile feedback was delivered by the two circular vibrators, which were 12 mm in diameter, 3.4 mm in height, and 1.7 g in mass. They were placed along the long axis of the forearm of the affected side for amputees (Figure 1A). For the control participants, both the instrumented insole and the vibrators were placed on the body's left side. The two vibrators were separated by 10 cm, which was distant enough to prevent possible perceptual ambiguity across the simulated locations. The vibration amplitude and frequency were coupled together for the miniaturized vibrators. Thus, we only adjusted their vibration intensity by a pulse width modulation (PWM). The vibration intensity was modulated by the duty cycle of the PWM signal. Both the force sensors and the vibrators were connected to a tablet computer (Microsoft Surface 4) via an RS232 serial interface with a customized driver circuit. A customized Matlab application was used for real-time signal processing (Mathworks, version

186 2013a). The plantar pressure signals drove the vibrators in real-time with a nonlinear mapping

187 function (see below).



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189 **Figure 1**: Illustration of the sensory substitution system. (A) The insole is instrumented with

190 four thin electropiezo force sensors whose measurements are routed to a tablet for real-time

191 data processing. The measured force determines the vibration intensity of the two tactile

192 vibrators attached along the forearm's longitudinal axis on the same side of the insole. When

193 participants walk or stand still, the vibration provides real-time feedback of the balance

194 performance from the measured foot. (B) The vibration intensity follows a logarithmic

195 relationship with the balance index (BI), which is determined by the changing force loading

196 caused by body sway. Forward or backward body lean would lead to one vibrator working,

197 respectively.

198

199 **The mapping between the plantar pressure signals and the vibrotactile stimulation**

200 The readings from the force sensors were converted into an index signifying the body weight

201 shifts in the anterior-posterior direction. We named this index as balance index (*BI*). It was

calculated as the ratio between the average force of the three force sensors in the forefoot (marked as 1 to 3 in Figure 1A) and the force of the 4th force sensor placed under the hallux:

$$BI = \frac{(F1 + F2 + F3)/3}{F4}$$

Where F1, F2, F3, and F4 are the readings from the four force sensors, respectively. Thus, the changes in the amplitude of *BI* denote the postural sway in the anteroposterior direction. If the body leans forward, the signal strength of the force sensors in the forefoot will increase while the signal strength of the 4th force sensor under the heel would decrease, increasing *BI*. Conversely, a backward body sway would lead to a decrease in *BI*. We estimated the average *BI* for the neutral posture when each participant was asked to stand still without any disturbance in so-called baseline trials (see below). This average *BI* was defined as an equilibrium point (EP), and typically the *BI* would oscillate around each subject's EP. The *BI* changes around the EP would be transformed into vibrotactile stimuli delivered to the forearm.

To reduce the ambiguity of vibrotactile signals, we only activated one vibrator at a time: when the *BI* was larger than the EP, the vibrator placed in the front would vibrate to signal a forward lean, and vice versa. The intensity of vibration for each vibrator was determined by the absolute difference in *BI* between the current state and the equilibrium state at EP:

$$Intensity = \frac{\log(BI - BI_{EP})}{\log BI_{max}}$$

Where BI_{EP} is the average *BI* estimated at EP when no visual perturbation was applied, and BI_{max} is the maximum *BI* in the forward or the backward direction estimated from the trials

222 when the participants first encountered visual perturbation on day 1 (sensory substitution was
223 off; see below). The relation between the vibration intensity and the *BI* followed a logarithmic
224 function (Figure 1B). When the *BI* slightly oscillated around the equilibrium point as
225 participants maintained a relatively neutral position, the vibrotactile feedback was weak. As
226 the *BI* deviated more from EP, the intensity would increase, approaching the maximum
227 vibration intensity specified by the maximum *BI* estimated in the baseline trials. Thus small
228 body sways would be more perceivable with the logarithmic transformation than a simple
229 linear function. Correspondingly, for large body sways, the tactile stimulation is not as strong
230 as with a linear function. We "sacrifice" the range of large signals in our tactile coding since
231 large body sways are readily perceivable by other sensory modality such as vision and
232 proprioception. Furthermore, studies of human psychophysics indicated that tactile perceptual
233 discrimination deteriorates with stimulus intensity (35, 36), suggesting that large tactile
234 signals are less informative. Thus, our sensory substitution's encoding scheme is to highlight
235 the feedback of small body sways but discount that of large body sways in keeping with
236 psychophysics principles.

237 **Participants**

238 We recruited seven transtibial amputees as the test group (including six males and one female
239 with an average age of 40.86 ± 9.40 years old) and eight non-disabled participants as the
240 control group (including six males and two females with an average age of 23.13 ± 1.69
241 years). The amputation time for amputee participants ranges from 8 to 26 years (15.29 ± 5.99
242 years). Amputation was on the left side for six amputees and on the right for one amputee. All

243 participants recruited in this study had no neuromotor disease or severe cardiovascular and
244 cerebrovascular diseases. All of them provided informed and written consent before the
245 experiment and were paid for their participation. The Institutional Review Board of Peking
246 University approved all procedures.

247 **Experiment**

248 The whole experiment was split into two parts and completed in two successive days. On day
249 1, all participants finished four blocks of 36 trials wearing the sensory substitution system
250 without turning on the vibration. However, their plantar pressure data were collected during
251 quiet standing. These trials also serve as baseline trials for computing BI_{EP} and BI_{max}. In this
252 way, we took the individual difference of body weight and foot conditions into consideration
253 for designing individualized vibrotactile stimulation for each participant. These parameters
254 were adopted in the sensory substitution system for subsequent tests of posture stability under
255 visual disturbances.

256 After baseline trials, postural stability was evaluated with the moving-room paradigm. The
257 experiment was conducted in a dark room while the participant maintained a quiet standing
258 posture 50 cm in front of a back-projection screen (Figure 2). The visual stimuli to provide
259 postural disturbance were projected onto the vertically-installed translucent screen by a
260 projector (InFocus, model IN104). The viewing area was 102 cm long and 68 cm high,
261 centered in between two eyes. Throughout the experiment, the participants wore a pair of
262 goggles limiting the field of view to approximately 120° wide and 60° high. Thus, the screen
263 edge was not visible to the participant, preventing it from being served as a visual reference

for stabilizing posture. We tracked participants' head movements throughout the experiment by an infrared motion capture system (OptiTrack, V120: Trio, Natural Point Inc.). A reflective marker was fixed on the goggle side and approximately centered in the measurement volume of the motion capture cameras. As participants stood on a plantar pressure mat (RsScan Inc., Model footscan), their center of pressure (CoP) movements were simultaneously measured along with their head movement. The sampling frequency was set at 60 Hz for both measurements. The stimuli presentation was generated by using Psychtoolbox package in Matlab, and data acquisition was controlled by a single customized Matlab program (Mathworks, version 2013a). We also used customized Matlab codes for data analysis.

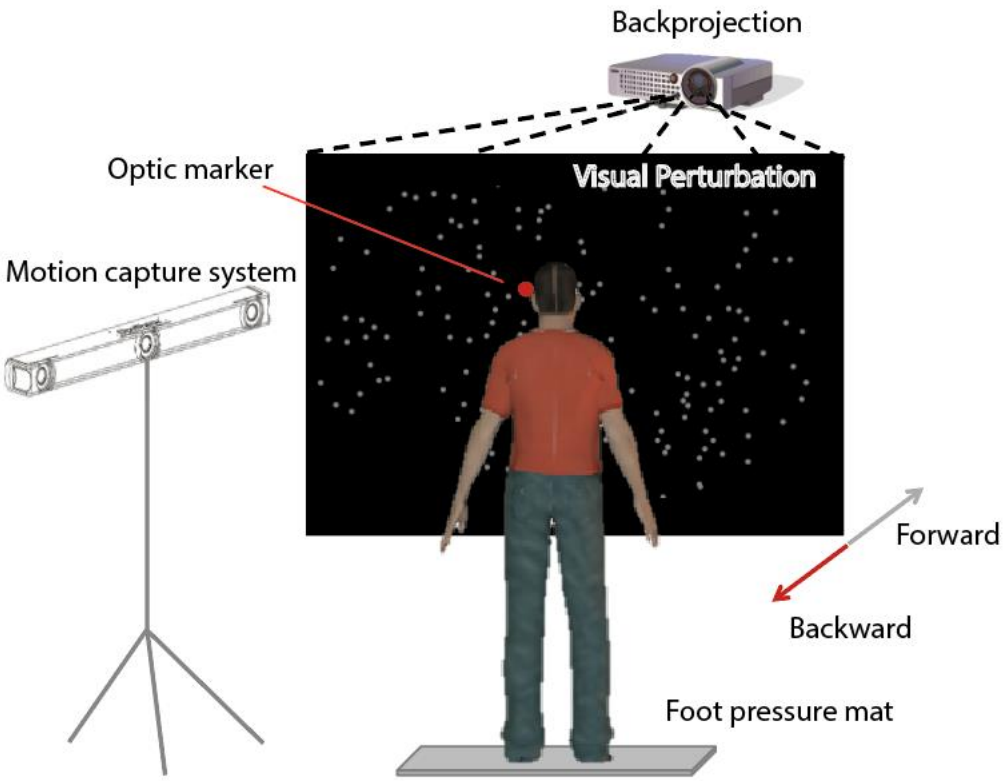

Figure 2: Illustration of the experimental setup for the postural stability test. The participant stands on a plantar pressure mat, facing a large projection screen. The field of view is limited to the screen by asking the participant to wear a pair of goggles. The head motion is simultaneously tracked by a motion tracking system with a marker placed at the eye level. The screen displays a cloud of random dots with simulated motion in the depth direction to perturb the standing posture in the anteroposterior direction.

For the postural stability test, we adopted the classical moving-room paradigm where the visual oscillatory disturbance was continuously presented to the participant (42, 43). The stimulus consisted of 200 randomly generated dots, each with a size of 0.57 deg in diameter. The dots were randomly distributed in an annulus between 10 deg and 45 deg visual eccentricity (44). No stimulus was presented in the central foveal region to avoid aliasing effects (42). Effectively, the dots simulated a space with depth before participants' eyes. During the experiment, the depth of the visual scene oscillated in the anteroposterior direction. This was achieved by changing the size of the dots and the distance between the dots according to visual perspective. The anteroposterior movement of the visual stimulus was sinusoidal with a certain frequency and amplitude. As the body sway was modulated by both the frequency and amplitude of the oscillation, we used three frequencies (0.1/0.3/0.5 Hz) and three amplitudes (2/4/8 cm) to cover the parameter range typically reported in the literature. This resulted in a total of nine stimulus conditions.

Both the amputee group and the control group were examined for their postural stability with and without sensory substitution. Each participant went through all the nine stimulus conditions, four trials each condition. The total 72 trials were arranged as eight trial blocks, four blocks with sensory substitution and the other four without. Each block thus consisted of 9 trials, one trial for each of the nine stimulus conditions. Trials were randomly ordered within each block. Each trial lasted 140s, and the first 20s were left out of subsequent analysis since the large but transient postural sway at the beginning of a trial was a reflexive response to the abruptly-induced visual disturbance (44). For testing their postural stability, participants were instructed to fixate at the center of the display, which was left free of moving dots with a 10° eccentricity. As the visual scene moved in the anteroposterior direction, the participant's CoP was also displaced in the same direction, accompanied by *BI* index changes and the corresponding intensity changes of the tactile stimulation when the sensory substitution system was on (Figure 3A). They were also encouraged to stand in a relaxed manner during stimuli presentation. To prevent fatigue, we administrated a rest of 2 to 3 minutes between trials and a mandatory rest of 5 minutes between blocks.

Given that the whole experiment lasted about 7 hours, we divided the whole experiment into two days with four blocks of trials on each day. Participants needed to complete a total of 36 trials in 4 blocks without sensory substitution on day 1 to establish their baseline postural stability before sensory substitution. They then completed another 4 blocks of 36 trials on day 2 to examine the effect of sensory substitution. Note, as previous studies have not shown any

habituation of visual disturbance in the moving room paradigm, we did not counterbalance the conditions between days.

Data analysis

We analyzed the CoP or head movements while the participant was visually disturbed to evaluate their postural stability (see Figure 3B for an exemplary trial). For each trial, we computed the range of CoP and head movement in the A-P direction as a measure for postural stability against the visual disturbance. The range measure specifically quantifies the maximum body sway induced by the visual disturbance. We standardized the range of head movement by dividing it with the height of each participant to minimize the effect of individual differences in body height. We used Fourier transforms to analyze the CoP and head movement in the anteroposterior direction and obtained the signal power over the frequency range between 0 and 2Hz. As the frequencies of body sway and of visual disturbance were way below 2Hz, this power measurement specifically quantifies the average body sway over time. Thus, both range and power measurements quantify the postural stability with their larger values corresponding to less stability. For each measure, we conducted a 4-way mixed-design ANOVA with 3 (stimuli frequency) x 3 (stimuli amplitude) x 2 (sensory substitution on vs. off) x 2 (amputee vs. control group). As we observed the stability improvement across all the conditions, we also computed the performance improvement in the four performance variables by taking their difference between with and without sensory substitution. Then, the improvements were submitted to 3-way mixed-design ANOVA with 3 (stimuli frequency) x 3 (stimuli amplitude) x 2 (amputee vs. control group) to

examine whether the sensory substitution effect differed between conditions and groups. A greenhouse-Geisser correction was used when the data did not meet the sphericity assumption of ANOVA. We set the significance level at $\alpha = 0.05$.

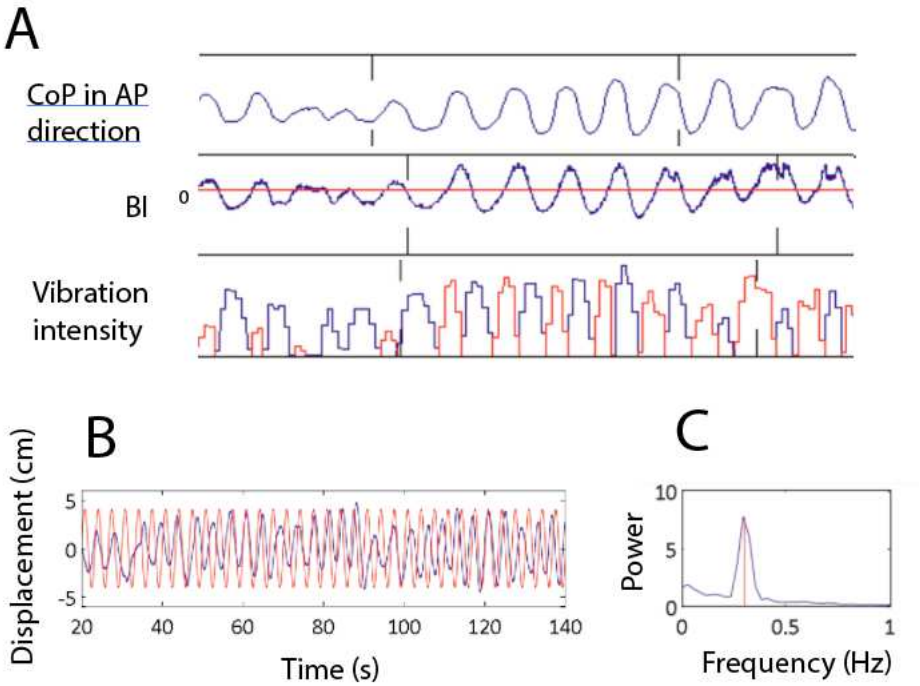

Figure 3: Exemplary data from the moving room paradigm. A) An exemplary trial segment to show how the sensory substitution system works. The participant is perturbed by the oscillatory visual stimuli, resulting in large CoP displacement in the anteroposterior (AP) direction. Our system computed the *BI* index in real-time and changed the vibration intensity of the two vibrators (shown in blue and red, respectively) placed on the forearm of the participant. B) An exemplary trial with head displacement (blue) and visual stimulus displacement (red) in the anteroposterior direction. The formal data collection begins at the 20th second. C) The power spectrum of head movement data of the same trial. The frequency

of the visual stimulus here is 0.3Hz. The integral of the power over the frequency range between 0 and 2Hz was used for evaluating postural stability.

Results

We found that the visual disturbance modulated coP displacement and head movement, and the body sway was reduced when the sensory substitution system was on for both groups of participants. These effects can be readily shown as reduced CoP displacement (Figure 4 & 5) and head movement (Figure 6 & 7). The four-way ANOVA on the range of CoP displacement revealed no main effect of group ($F_{(1, 13)} = 0.12$, $p = 0.74$, *partial* $\eta^2 = 0.009$), but a significant main effect of sensory substitution ($F_{(1, 13)} = 19.47$, $p = 0.001$, *partial* $\eta^2 = 0.60$). Across groups, the CoP range before applying sensory substitution was larger than after (5.22 ± 0.77 cm v.s. 3.38 ± 0.47 cm, mean ± std. error, same below). In fact, the improvement in CoP stability was significant in all the nine stimulus conditions (3 stimulus frequency × 3 amplitude) after Bonferroni correction (all *p*s<0.01). The main effect of stimulus frequency was not significant ($F_{(2, 26)} = 1.48$, $p = 0.26$, *partial* $\eta^2 = 0.10$) but the main effect of the stimulus amplitude was ($F_{(2, 26)} = 12.62$, $p < 0.001$, *partial* $\eta^2 = 0.49$). For interaction effect, only the interaction between stimulus frequency and sensory substitution reached significance ($F_{(2, 26)} = 5.85$, $p = 0.008$, *partial* $\eta^2 = 0.31$). This interaction suggested that the benefit brought by sensory substitution was larger in the conditions with a higher frequency, which was more perturbing than lower frequency conditions.

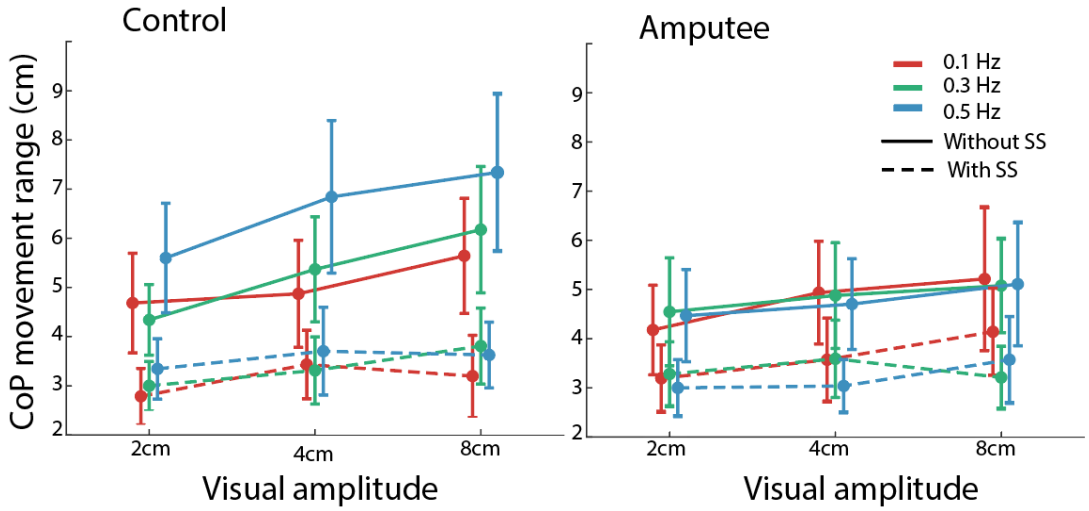

Figure 4: The range of CoP displacement in the anteroposterior direction plotted as a function of stimulus amplitude and frequency. The conditions with and without sensory substitution (SS) are shown in separate lines. The able-bodied control group and the amputee group are shown in the left and right panels, respectively.

Power spectrum analysis of CoP displacement showed similar effects as the CoP range (Figure 5). The power of CoP displacement was submitted to the same four-way ANOVA. The main effect of sensory substitution was significant ($F_{(1, 13)} = 9.197$, $p = 0.010$, *partial* $\eta^2 = 0.41$), indicating that turning on the sensory substitution system reduced the COP excursion in response to the visual disturbance. The main effects of group and stimulus frequency were not significant ($F_{(1, 13)} = 1.40$, $p = 0.26$, *partial* $\eta^2 = 0.10$ for group; $F_{(1.19, 15.47)} = 2.00$, $p = 0.177$, *partial* $\eta^2 = 0.13$ for stimulus frequency). The main effect of stimulus amplitude was marginally significant ($F_{(1.04, 13.51)} = 4.24$, $p = 0.058$, *partial* $\eta^2 = 0.25$). None of the interaction effects was significant except the interaction between sensory substitution and

stimulus amplitude ($F_{(1.06, 13.74)} = 4.59$, $p = 0.049$, *partial* $\eta^2 = 0.26$). The interaction, again, indicates that the benefit of sensory substitution was more pronounced in the conditions with larger visual amplitudes than with lower amplitudes. Overall, power spectrum analysis revealed reduced body sway for both groups of participants when the sensory substitution system was on.

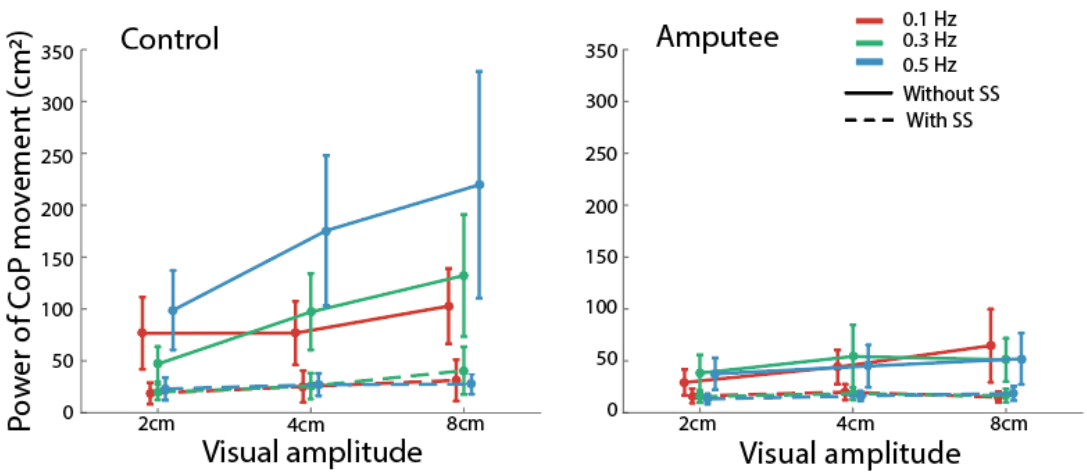

Figure 5: The power of CoP displacement plotted as a function of stimulus amplitude and frequency. The conditions with and without sensory substitution (SS) are shown in separate lines. The able-bodied control group and the amputee group are shown in the left and right panels, respectively.

We further examined how the improvement of CoP sway by sensory substitution varied across stimulus conditions. The reduction in CoP range by sensory substitution was computed for each condition and then submitted to a 2 (group) × 3 (stimulus frequency) × 3 (stimulus amplitude) mixed-design ANOVA. Either the main effect of group or stimulus amplitude was significant ($F_{(1, 13)} = 1.17$, $p = 0.298$, *partial* $\eta^2 = 0.08$ for group; $F_{(2, 26)} = 2.42$, $p = 0.109$,

partial η^2 = 0.16 for stimulus amplitude). However, the main effect of stimulus frequency was significant ($F_{(2, 26)}$ = 5.85, p = 0.008, *partial* η^2 = 0.31). None of the interactions was significant. The same ANOVA on the power of CoP displacement yielded similar results: no group difference or interaction was detected. However, the main effect of stimulus amplitude but not of stimulus frequency reached significance ($F_{(1.06, 13.74)}$ = 4.59, p = 0.049, *partial* η^2 = 0.26 for stimulus amplitude; $F_{(1.15, 14.91)}$ = 1.67, p = 0.219, *partial* η^2 = 0.11 for stimulus frequency). These results suggest that both groups of participants benefited more from sensory substitution in those more challenging conditions with larger stimulus frequency or amplitude.

While the CoP displacement reflects the overall body weight shifts during standing, the head movement directly reflects the body sway at the eye level. We found that head movements also showed a similar benefit of sensory substitution (Figure 6). For the head movement range, the main effect of group was not significant ($F_{(1, 13)}$ = 0.05, p = 0.820, *partial* η^2 = 0.004). The average head movement range was comparable between the amputee group (4.20 ± 0.75 cm) and the control group (4.44 ± 0.70 cm). Importantly, the main effect of sensory substitution was significant ($F_{(1, 13)}$ = 11.98, p = 0.004, *partial* η^2 = 0.48). Across groups, the head movement range decreased from 4.94 ± 0.66 cm to 3.71 ± 0.40 cm when the sensory substitution was used. Again, we found that the improvement brought by sensory substitution was significant in all nine stimulus conditions (all ps<0.05 after Bonferroni correction) except in the condition with 0.3Hz stimulus frequency and 2cm amplitude (marginally significant with p = 0.098). The main effect of stimulus frequency was not significant ($F_{(2, 26)}$ = 0.75, p =

0.484, *partial* η^2 = 0.05) but the main effect of stimulus amplitude was ($F_{(2, 26)}$ = 7.96, p = 0.002, *partial* η^2 = 0.38). Thus, stimulus amplitude, but not stimulus frequency, modulated the head motion, a similar result as the CoP range. All the interactions failed to reach significance except the interaction between sensory substitution and stimulus amplitude ($F_{(2, 26)}$ = 3.74, p = 0.037, *partial* η^2 = 0.22), again suggesting that the benefit of sensory substitution was more pronounced with larger visual disturbances.

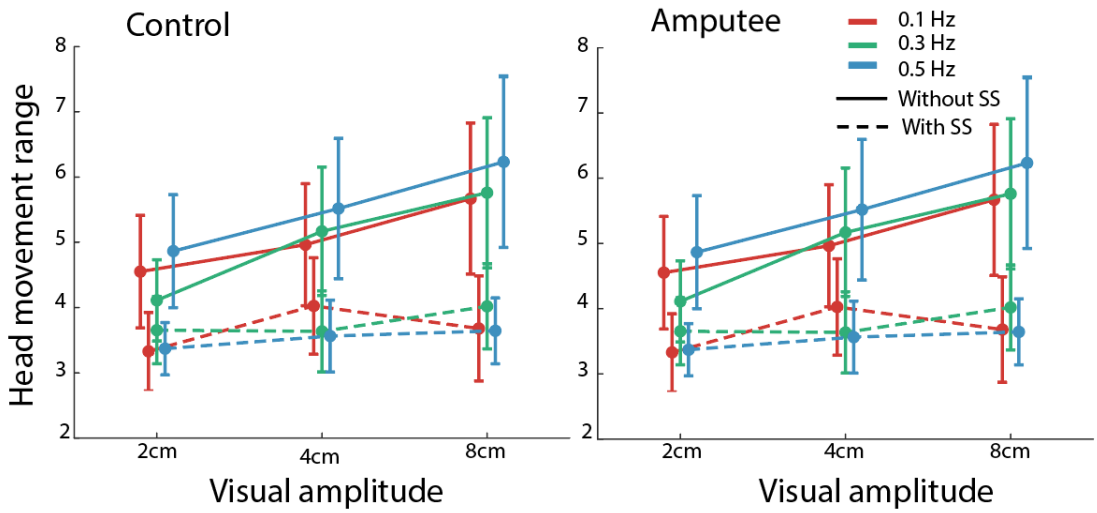

Figure 6: The head movement range in the anteroposterior direction plotted as a function of stimulus amplitude and frequency. The conditions with and without sensory substitution (SS) are shown in separate lines. The able-bodied control group and the amputee group are shown in the left and right panels, respectively. Note the head movement range is unit-less as it is normalized by dividing the participant's body height.

Power spectrum analysis of head movement revealed a similar pattern as the range of head movement (Figure 7). The power of head movement was submitted to the same four-way

ANOVA. The main effect of group was not significant ($F_{(1, 13)} = 1.40$, $p = 0.259$, *partial* $\eta^2 = 0.10$). The main effect of sensory substitution was significant ($F_{(1, 13)} = 9.20$, $p = 0.010$, *partial* $\eta^2 = 0.41$), and the power decreased when the sensory substitution system was on. Main effect of stimulus amplitude was marginally significant ($F_{(1.04, 13.51)} = 4.24$, $p = 0.058$, *partial* $\eta^2 = 0.25$), but not for stimulus frequency ($F_{(1.19, 15.47)} = 2.00$, $p = 0.177$, *partial* $\eta^2 = 0.13$). None of the interactions was significant except the interaction between sensory substitution and stimulus amplitude ($F_{(1.17, 15.22)} = 5.53$, $p = 0.028$, *partial* $\eta^2 = 0.30$). Thus, while the larger stimulus amplitude tended to cause larger body sway, the same sensory substitution effect was also larger with larger stimulus amplitudes.

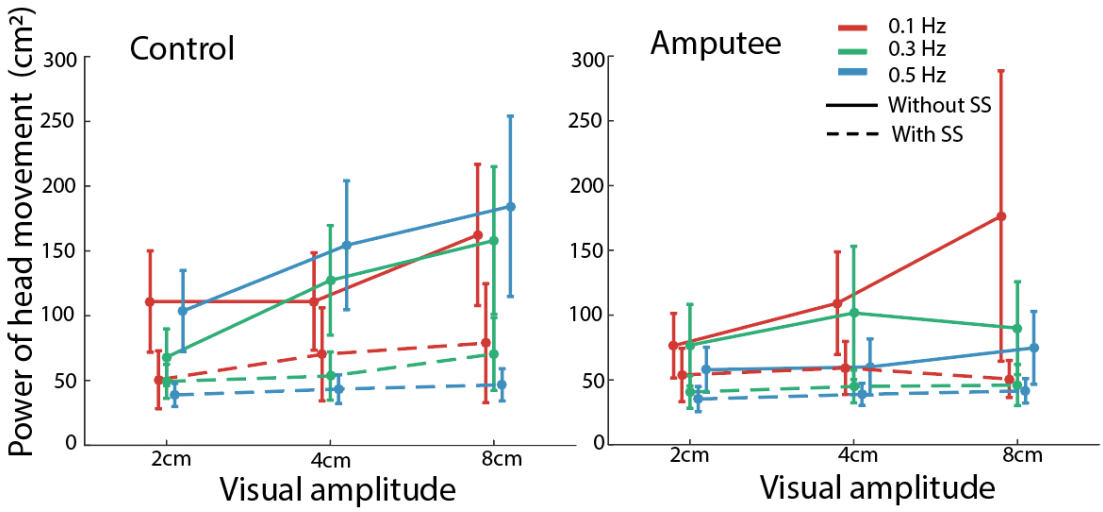

Figure 7: The power of head movement plotted as a function of stimulus amplitude and frequency. The conditions with and without sensory substitution (SS) are shown in separate lines. The able-bodied control group and the amputee group are shown in the left and right panels, respectively.

We then examined how the improvement in head stabilization brought by sensory substitution differed between stimulus conditions. The reduction of head movement range was submitted to the 2 (group) × 3 (stimulus frequency) × 3 (stimulus amplitude) mixed-design ANOVA. Either main effect of group or stimulus frequency was significant ($F_{(1, 13)}$ =0.72, p = 0.413, *partial* η^2 = 0.05 and $F_{(2, 26)}$ = 0.86, p = 0.436, *partial* η^2 = 0.06, respectively). None of the interactions was significant. The reduction in the power of the head movement was also examined by the same three-way ANOVA. Again, no main effect of group or interaction was detected. The main effect of stimulus amplitude but not stimulus frequency was significant ($F_{(1.17, 15.22)}$ =5.53, p = 0.028, *partial* η^2 = 0.30 and $F_{(1.26, 16.44)}$ = 0.22, p = 0.698, *partial* η^2 = 0.017, respectively). Thus, similar to the CoP displacement, we found that head movements tend to be stabilized more by sensory substitution in those more challenging stimulus conditions.

Discussion

This study aims to investigate whether lower-limb amputees can improve their postural stability with real-time vibrotactile feedback to surrogate their missing foot plantar pressure information. We designed an intuitive coding scheme for vibrotactile feedback, which was spatially congruent with body weight shifts and in keeping with tactile psychophysics principles. We assessed the standing stability of both lower-amputees and non-disabled control participants in the classical moving-room paradigm. Analyses of CoP displacement and head movement indicated that both groups improved their balance control when the

sensory substitution was applied across various visual conditions. We also found that the balance improvement brought by sensory substitution was more pronounced for more challenging conditions with larger visual disturbance. Thus, our findings suggest that closing the broken sensorimotor loop of lower-limb amputees by using real-time sensory substitution can help improve postural control and, potentially, other actions that involve ground-foot interactions.

Postural control is under the simultaneous influence of multiple sensory modalities, including visual, vestibular, proprioceptive, and tactile modalities. For maintaining postural stability during standing, the nervous system adjusts the relative contributions of sensory inputs from different channels during the multisensory integration process according to sensory precision of individual channels (45-47). In the moving-room paradigm, the visual scene oscillates and biases the estimated standing posture, resulting in postural sway (44). Sensory inputs from other modalities, including the augmented tactile feedback applied on body parts other than the foot, can negate the visual disturbance. For example, researchers have shown that the light touch of a fingertip on a stable surface can provide subtle tactile feedback for stabilizing posture during quiet standing and standing under visual interference (48, 49). Vuillerme, Chenu (50) used a 6×6 electrotactile matrix on the tongue to provide feedback of CoP changes for the non-disabled participants and improve their postural stability in a condition where neck proprioceptive and vestibular inputs were compromised with an unnatural posture. Thus, extra tactile inputs can indeed improve postural stability among non-disabled participants. Our study went a step further to show that lower-limb amputees could improve

their postural stability against visual disturbances with vibrotactile information contingent on the plantar pressure changes. Presumably, this stabilizing effect follows the same sensory integration principles that have been repeatedly reported in different paradigms (51).

Previous studies using similar vibrotactile feedback to substitute foot pressure have failed to show consistent benefit in postural stability (e.g., 26). We postulate that differences in the tactile coding scheme and the postural test are responsible for the discrepancy. The intuitiveness and comfort of augmented tactile feedback presented to the human wearer were not systematically investigated until recently (52). Our approach paid particular attention to make the tactile feedback intuitive. First, only bodyweight shifts, as measured by plantar pressure underneath the foot, were encoded. This is in contrast to the one-to-one signal mapping between a pressure sensor and a tactor in previous studies (e.g., 26). One-to-one mapping is technically straightforward, but it would pose a challenge for the wearer to understand tactile signals' meaning. Second, our system encodes the body sway in the anteroposterior direction, the prominent direction of instability during quiet standing, with the two tactile stimulators aligned with the forearm's longitudinal axis. Thus, the tactile feedback is spatially congruent with the visual disturbance and postural sway. Third, we limited the two stimulators to work one at a time and used a logarithmic transfer function to use better the perceptual range of tactile stimuli (35, 36). These signal designs help resolve the so-called neutral zone problem when people receive little tactile feedback around a neutral posture (26). These design aspects appeared to help participants, especially amputee participants who have

512 not received direct foot contact pressure information for long, quickly learn to use surrogate
513 sensory feedback to improve their postural control.

514 It is noteworthy that the benefit of our sensory substitution system manifested itself without
515 extensive training. Our participants familiarized themselves with the system on day one over
516 36 trials. Previous studies on sensory substitution typically required several weeks of practice
517 time (53, 54). We postulate that the intuitive encoding scheme of the vibrotactile feedback
518 facilitated this fast adoption of sensory substitution.

519 We used three stimulus frequencies (0.1/0.3/0.5 Hz) and three amplitudes (2/4/8 cm) to
520 perturb the participant visually in our postural control task. We found that the amplitude of
521 visual stimulus predominantly affected postural stability, as shown by different independent
522 measures. When the amplitude of visual motion increased, the body sway increased, as shown
523 in head movements and overall CoP displacement, consistent with previous research (55).
524 The oscillation frequency of visual disturbance showed an inconsistent effect on body sway.
525 For example, control participants tend to increase their power of CoP displacement and head
526 movement with increasing stimulus frequency, but amputee participants showed an opposite
527 tendency (Figure 5 and 7). When visual stimuli moved with a lower frequency (e.g., 0.1 Hz),
528 the body swayed periodically in synchrony with the driving visual stimuli. When visual
529 stimuli moved with a high frequency (e.g., 0.5 Hz), it became hard for the body sway to keep
530 up with the stimuli, resulting in a smaller power (42, 56). This saturation effect appears to be
531 more evident for amputees than for non-disabled participants.

We also computed the performance difference before and after sensory substitution to compare the effect size of sensory substitution across conditions. Three out of the four measures (i.e., the power of CoP displacement, the range and the power of head movement) showed a larger effect size in conditions with larger visual-stimuli amplitudes. The range of CoP displacement, the last measure, did not increase with visual amplitude, but it did increase with visual frequency. Thus, the sensory substitution system benefited both groups of participants more when they were faced with more challenging visual disturbance.

We found that sensory substitution stabilized the head and CoP with similar effect sizes. For the CoP range, the effect size of sensory substitution was 0.60 in partial η^2, which is equivalent to a 35.3% reduction after sensory substitution. In comparison, for the head movement range, the effect size was 0.48 with a 24.8% reduction. The same pattern was found in the power analysis, where both CoP displacement and head movement yielded an effect size of 0.41 with a reduction of 22.6%. Previous research has established that humans prioritize stabilizing the head with visual feedback when the overall posture changes (57). Furthermore, if we assume that the standing body resembles an inverted pendulum as in typical postural models (58), the head movement should decrease more when the CoP decreases. Thus, theoretically, we shall expect a more significant stabilizing effect of sensory substitution for the head than for the CoP. The lack of difference between the head and the CoP, or even a slightly more significant effect for the CoP, does not fit the theoretical prediction. We postulate that this might be attributed to the specificity of surrogated sensory information delivered by our sensory substitution system: the vibrotactile feedback reflects

553 plantar pressure changes directly related to CoP excursion, not to head movement. Thus,
554 when the nervous system integrates this surrogate sensory information, it readily responds to
555 CoP displacement induced by visual disturbances. Therefore, our findings appear to suggest
556 that sensory substitution exerts its influence on motor control in a stimulus-specific way, at
557 least for the situation investigated here where sensory substitution is adopted for a short
558 period of time. Future studies could test this hypothesis by comparing the responses to
559 substituted stimuli that encoded different body motion signals, e.g., head motion instead of
560 CoP displacement.

561 Interestingly, no group difference of postural stability between amputees and the control
562 reached significance for all the performance measures investigated. We expected that
563 amputees would be perturbed more by the visual disturbances since previous studies have
564 shown that amputees are more dependent on visual inputs (39-41). However, we recognize
565 that these studies used paradigms that reduced visual sensory feedback for the participants.
566 Understandably, it was harder for amputees than the non-disabled to accommodate visual
567 deprivation due to the loss in somatosensory feedback associated with amputation. In the
568 present study, however, we used a visual perturbation paradigm rather than visual deprivation.
569 According to multisensory integration theory in postural control (45-47), both amputees and
570 the non-disabled could adjust the weights of different sensory channels when sensory inputs
571 (i.e., visual input) became inaccurate. Furthermore, previous studies reported worse standing
572 balance among amputees typically used short trials, e.g., 20 s per trial (59). Our experiment
573 instead used as long as 140 s per trial; thus, both groups had ample time to adjust their

weights of different sensory channels and adapt to the visual stimuli. The other factor is that most of our participants have worn artificial limbs for more than ten years. After prolonged use of prosthesis, their performance in simple motor tasks such as quiet standing become indistinguishable from that of the non-disabled. In sum, the lack of group difference thus suggests that lower-limb amputees can effectively accommodate continuous visual disturbances.

The development of robotic artificial limbs has been made dramatic progress in fusing signals from various sensors for sensing the environment and the internal state of the prosthesis, but the research focus is more on intelligent control of prostheses (60). It is equally essential to route real-time sensory feedback for the agent, i.e., the human controller, to reduce the fear of falling, enhance the sense of embodiment of the prosthesis, and better motor control. This sensory augmentation for the agent can be achieved by invasive methods such as electrical peripheral nerve stimulation of the sciatic nerve (61) or noninvasive methods such as sensory substitution. As we pointed out in the introduction, substituting the missing feedback of foot-ground interaction is probably most important for lower-limb amputees. Still, the previous endeavors have been hampered by high demands of cognitive loads, unintuitive design, and inconsistent behavioral benefits. Our study has shown that these shortcomings of noninvasive sensory substitution can be overcome. It paves the way for us to integrate this method with robotic lower limbs. As most actuated lower-limb prostheses still lack afferent feedback to the user, it would be interesting to examine the outcome when our sensory substitution system integrates with these systems to achieve better human-centered close-loop control.

Furthermore, even though our postural tests showed the feasibility and effectiveness of tactile sensory substitution in stabilizing people's standing posture with minimal training, they were limited to transtibial amputees in the laboratory environment with a classical experimental task. Future endeavors should be directed to testing the system among transfemoral amputees and via dynamic balancing tasks, such as walking on different surfaces. We expect the need for specific modifications of the signal encoding scheme for diverse movement scenarios.

Conclusions

Using vibrotactile stimulation to substitute the missing plantar pressure information for transtibial amputee leads to improvements in postural stability during visually-perturbed quiet standing. Both amputees and able-bodied can benefit from sensory substitution, especially when large visual perturbations challenge their posture. Future development for sensory substitution shall consider making surrogated sensory inputs spatially congruent with the to-be-simulated sensory inputs and following psychophysical principles.

List of abbreviations

CoP: center of pressure; BI: balance index; SS: sensory substitution; ANOVA: analysis of variance

Declarations

- **Ethics approval and consent to participate**

The study was designed following the Declaration of Helsinki. The study protocol (NSFC2018-06-02) was approved by the Ethical Committee of Peking University (Beijing, China). All participants gave written informed consent.

- **Consent for publication**

Not applicable

- **Availability of data and materials**

The datasets used during the current study are available from the corresponding author on reasonable request.

- **Competing interests**

None of the authors have any competing interests to report.

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Authors' contributions

K.W. and Q.W. conceived and designed the experiment. L.C., Y.F., and B.C. implemented the experimental setup. L.C. and K.W. performed the data analysis. L.C and K.W. prepared figures; L.C. and K.W. wrote the manuscript. Data collection was performed by L.C., Y.F., and B.C.

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Figures

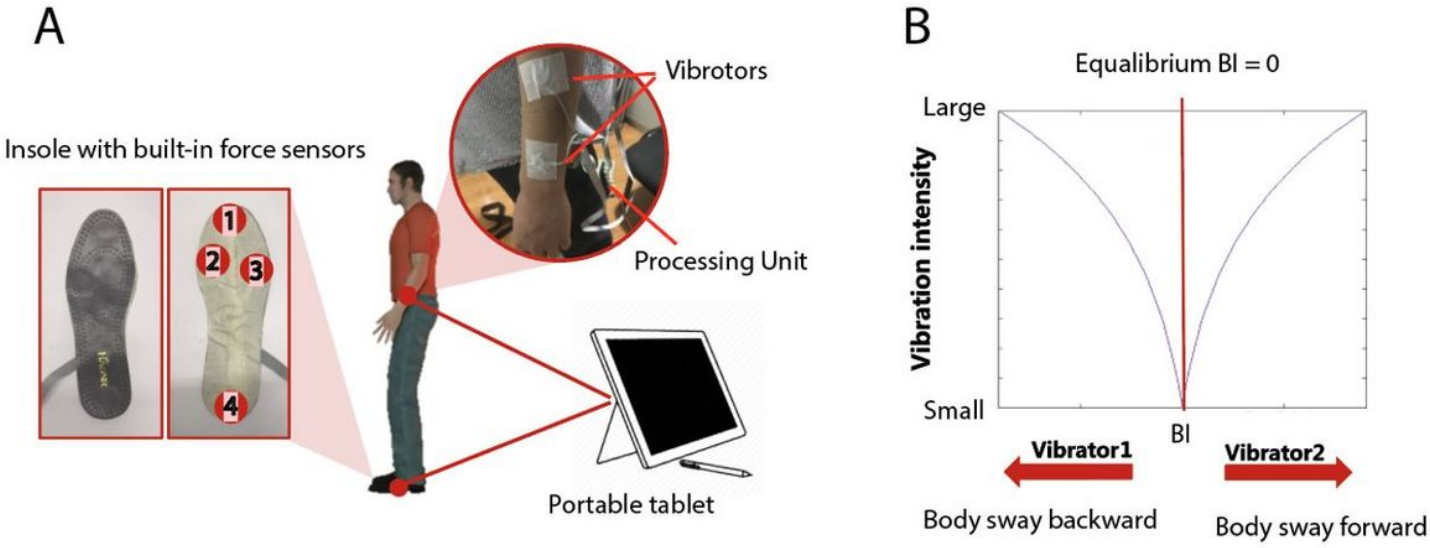

Figure 1

Illustration of the sensory substitution system. (A) The insole is instrumented with four thin electropiezo force sensors whose measurements are routed to a tablet for real-time data processing. The measured force determines the vibration intensity of the two tactile vibrators attached along the forearm's longitudinal axis on the same side of the insole. When participants walk or stand still, the vibration provides real-time feedback of the balance performance from the measured foot. (B) The vibration intensity follows a logarithmic relationship with the balance index (BI), which is determined by the changing force loading caused by body sway. Forward or backward body lean would lead to one vibrator working, respectively.

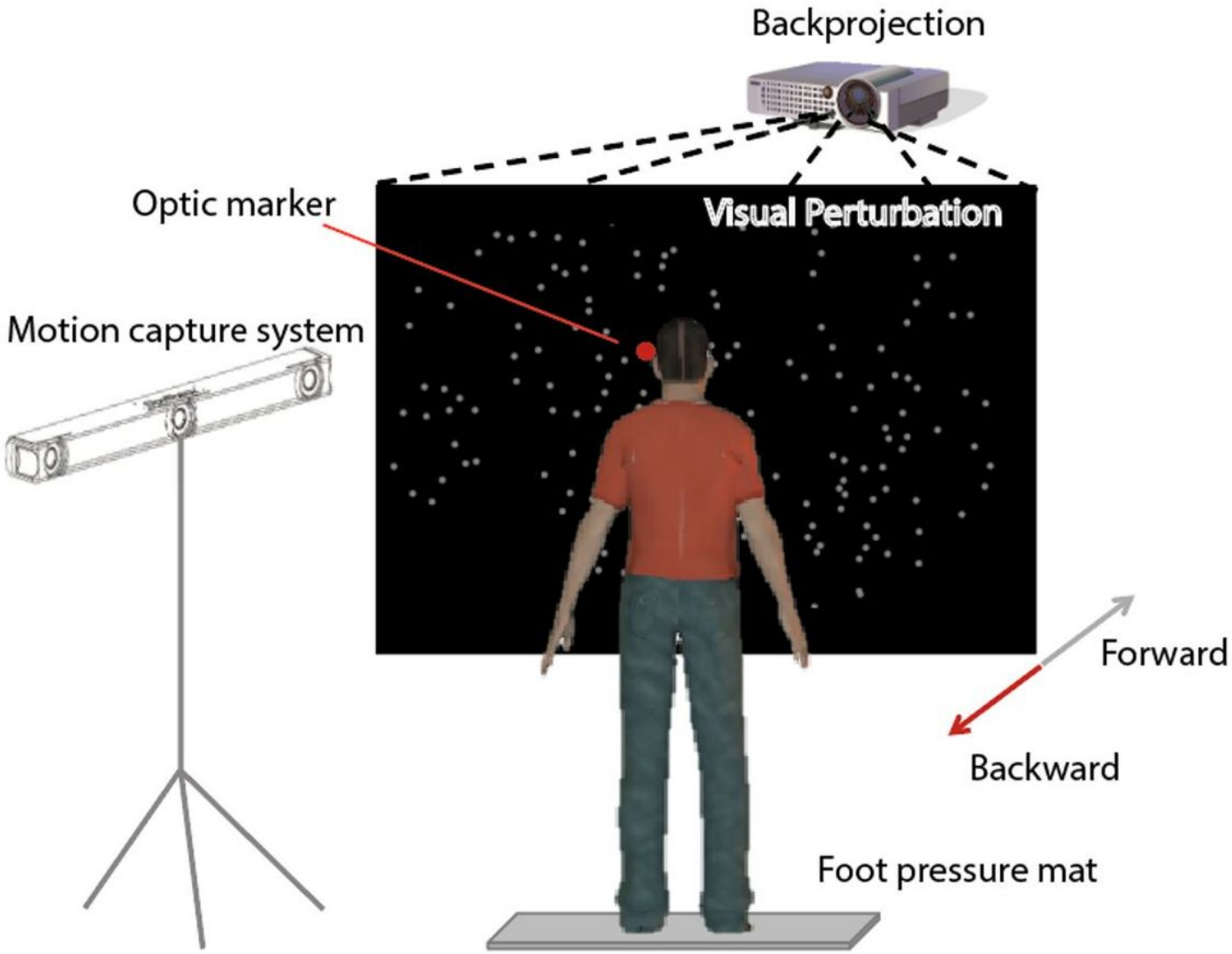

Figure 2

Illustration of the experimental setup for the postural stability test. The participant stands on a plantar pressure mat, facing a large projection screen. The field of view is limited to the screen by asking the participant to wear a pair of goggles. The head motion is simultaneously tracked by a motion tracking system with a marker placed at the eye level. The screen displays a cloud of random dots with simulated motion in the depth direction to perturb the standing posture in the anteroposterior direction.

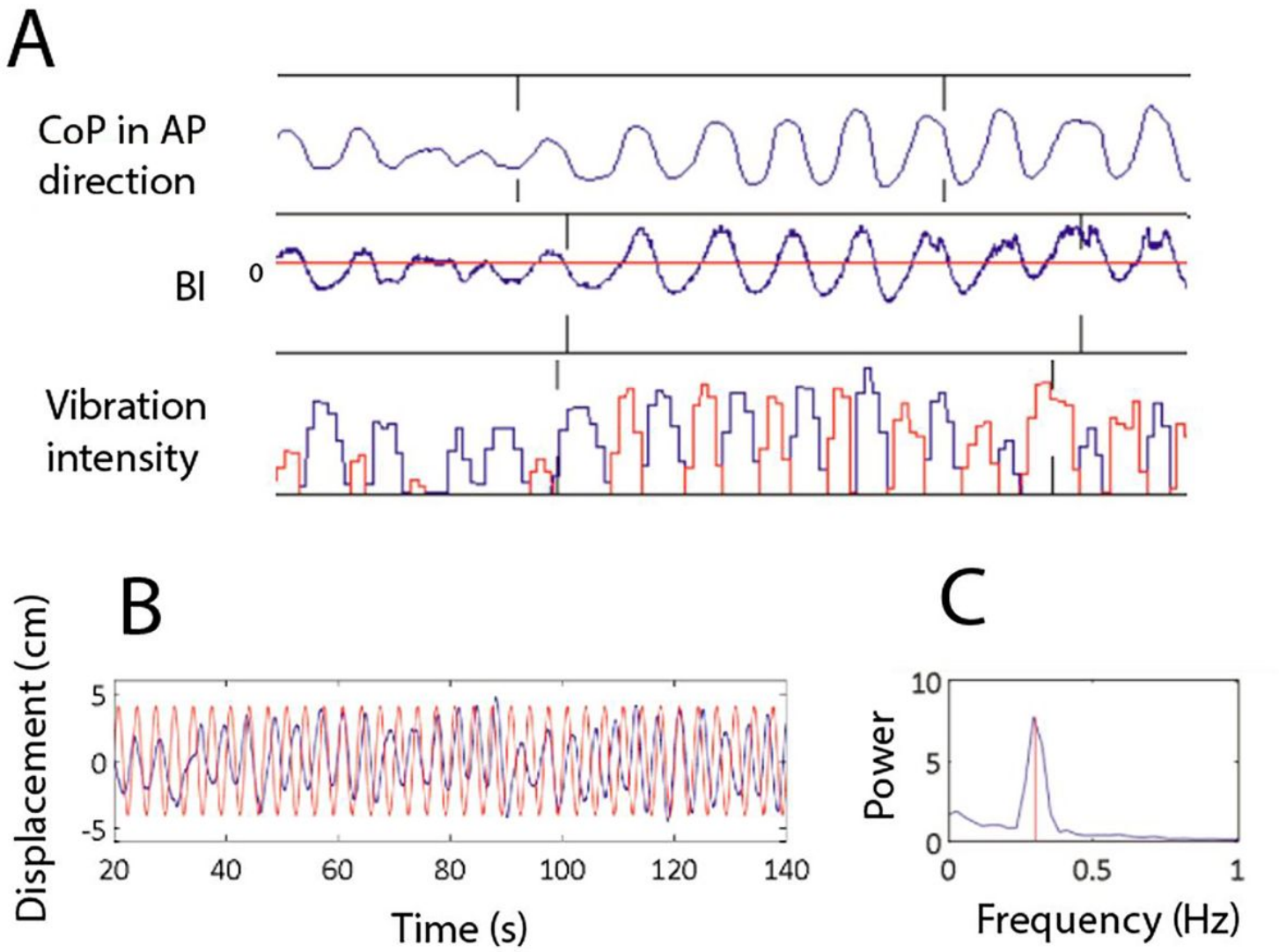

Figure 3

Exemplary data from the moving room paradigm. A) An exemplary trial segment to show how the sensory substitution system works. The participant is perturbed by the oscillatory visual stimuli, resulting in large CoP displacement in the anteroposterior (AP) direction. Our system computed the BI index in real-time and changed the vibration intensity of the two vibrators (shown in blue and red, respectively) placed on the forearm of the participant. B) An exemplary trial with head displacement (blue) and visual stimulus displacement (red) in the anteroposterior direction. The formal data collection begins at the 20th second. C) The power spectrum of head movement data of the same trial. The frequency of the visual stimulus here is 0.3Hz. The integral of the power over the frequency range between 0 and 2Hz was used for evaluating postural stability.

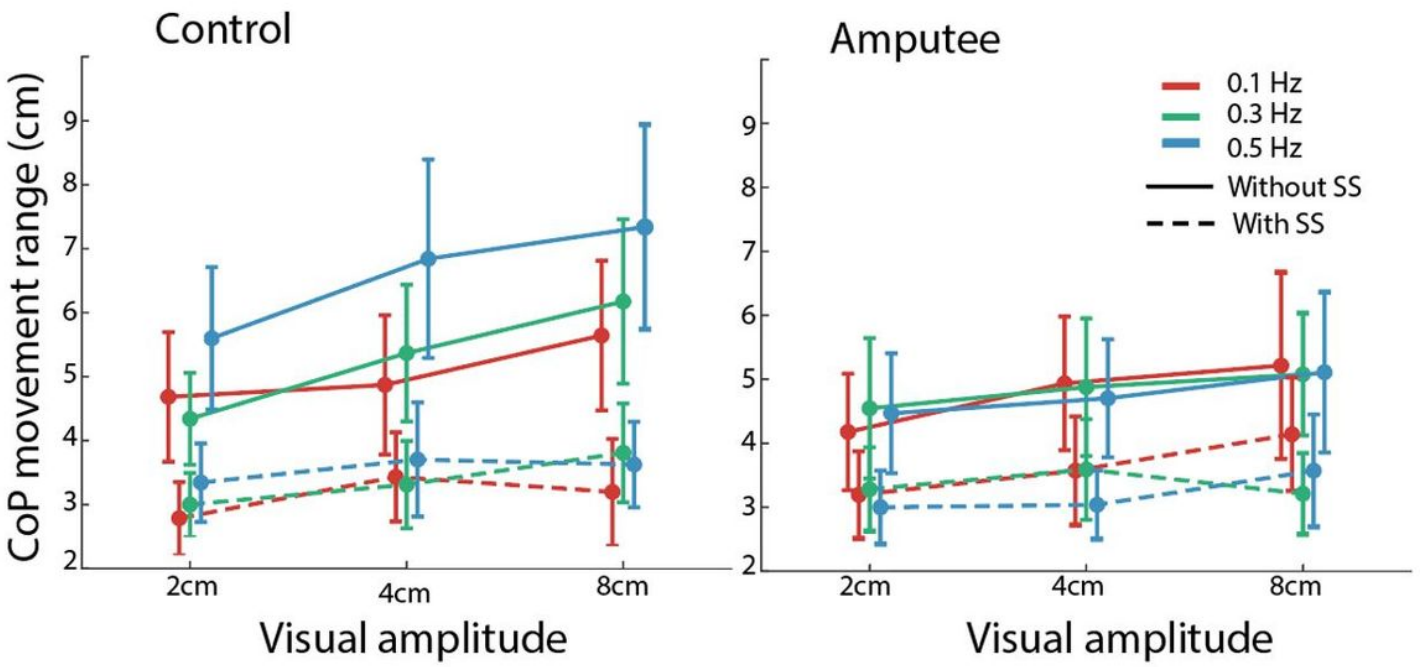

Figure 4

The range of CoP displacement in the anteroposterior direction plotted as a function of stimulus amplitude and frequency. The conditions with and without sensory substitution (SS) are shown in separate lines. The able-bodied control group and the amputee group are shown in the left and right panels, respectively.

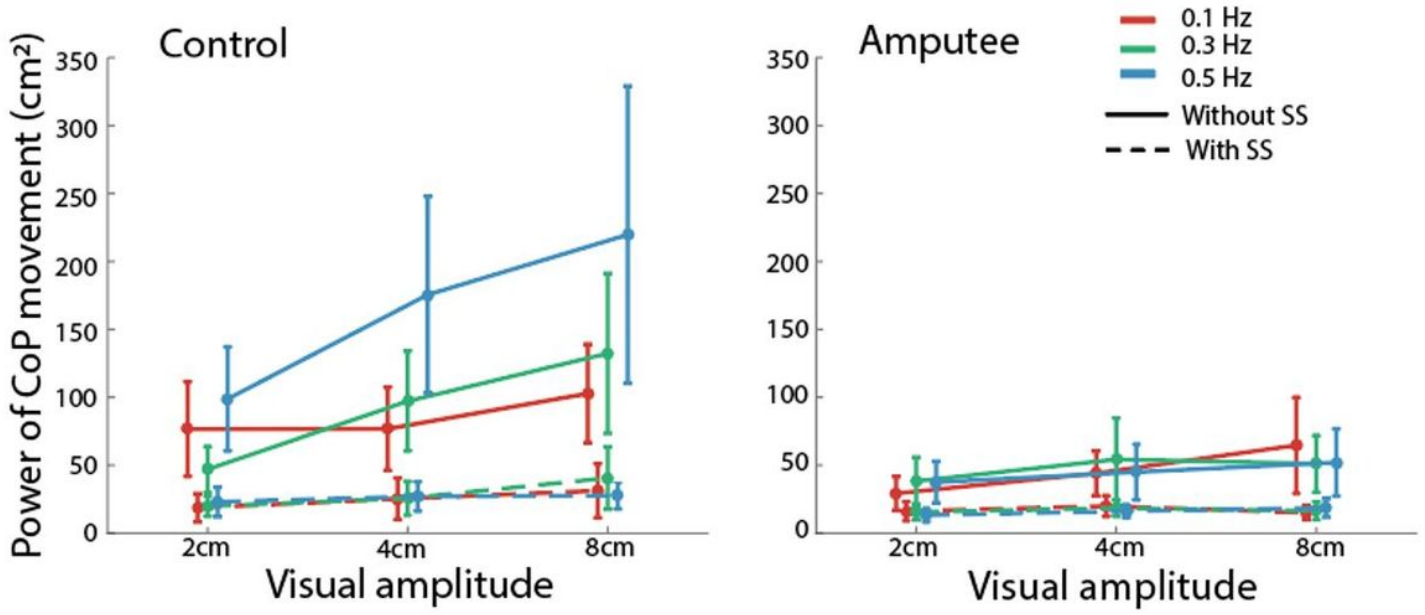

Figure 5

The power of CoP displacement plotted as a function of stimulus amplitude and frequency. The conditions with and without sensory substitution (SS) are shown in separate lines. The able-bodied control group and the amputee group are shown in the left and right panels, respectively.

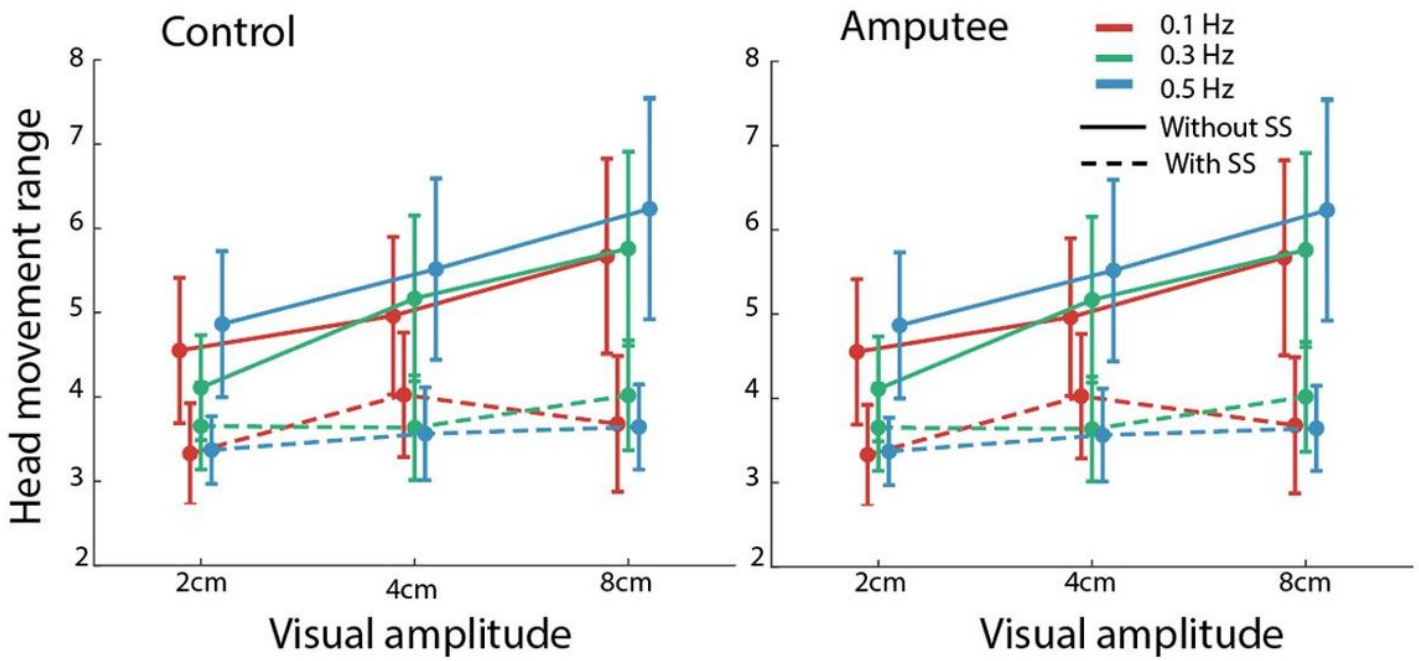

Figure 6

The head movement range in the anteroposterior direction plotted as a function of stimulus amplitude and frequency. The conditions with and without sensory substitution (SS) are shown in separate lines. The able-bodied control group and the amputee group are shown in the left and right panels, respectively. Note the head movement range is unit-less as it is normalized by dividing the participant's body height.

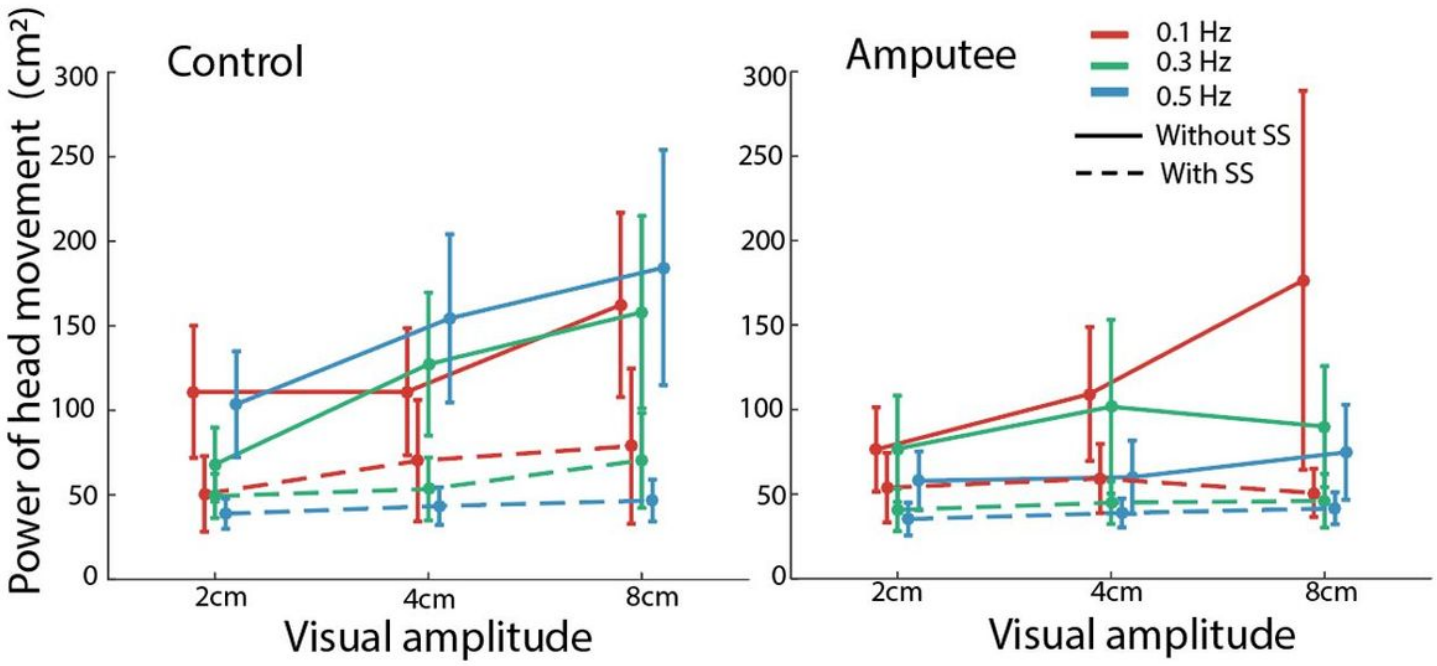

Figure 7

The power of head movement plotted as a function of stimulus amplitude and frequency. The conditions with and without sensory substitution (SS) are shown in separate lines. The able-bodied control group and the amputee group are shown in the left and right panels, respectively.