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Effects of White Noise Achilles Tendon Vibration on Quiet Standing and Active Postural Positioning

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Abstract

Applying white noise vibration to the ankle tendons has previously been used to improve passive movement detection and alter postural control, likely by enhancing proprioceptive feedback. The aim of the present study was to determine if similar methods focused on the ankle plantarflexors affect the performance of both quiet standing and an active postural positioning task, in which participants may be more reliant on proprioceptive feedback from actively contracting muscles. Twenty young, healthy participants performed quiet standing trials and active postural positioning trials designed to encourage reliance on plantarflexor proprioception. Performance under normal conditions with no vibration was compared to performance with 8 levels of vibration amplitude applied to the bilateral Achilles tendons. Vibration amplitude was set either as a percentage of sensory threshold ($n = 10$) or by root-mean-square (RMS) amplitude ($n = 10$). No vibration amplitude had a significant effect on quiet standing. In contrast, accuracy of the active postural positioning task was significantly ($P = .001$) improved by vibration with an RMS amplitude of 30 μm . Setting vibration amplitude based on sensory threshold did not significantly affect postural positioning accuracy. The present results demonstrate that appropriate amplitude tendon vibration may hold promise for enhancing the use of proprioceptive feedback during functional active movement.

Keywords

biomechanics; force plate; motor control; neurophysiology

Kinesthesia, the sense of body movement, is widely acknowledged to play an important role in functional mobility. Both proprioceptive receptors (eg, muscle spindles) and cutaneous receptors (eg, Ruffini corpuscles) can provide sensory feedback related to body segment position or velocity, thus contributing to kinesthesia.^{1–3} Deficits in the peripheral sensory sources contributing to kinesthesia are often observed among clinical populations (eg, stroke, peripheral neuropathy),^{4,5} potentially preventing the integration of multisensory feedback into an accurate perception of the body's mechanical state.⁶ Such somatosensory feedback clearly plays an important role in the control of standing posture,⁷ with ankle

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proprioception accuracy notably linked to anteroposterior postural sway among stroke survivors.⁸ Therefore, the enhancement of ankle proprioceptive feedback may hold promise for improving postural control.

One possible method to enhance sensory feedback involves stochastic resonance, a phenomenon by which low-amplitude noise increases the likelihood that weak signals will exceed a given threshold, improving detection of signal fluctuations.^{9,10} Such sensation-based methods have most commonly been applied using vibration or electrical current to enhance tactile sensation derived from cutaneous receptors.^{11–13} For example, noise applied to the foot soles has been used to reduce postural sway^{14–16} and gait variability,^{16–18} indicative of improved balance.

While less well studied than tactile enhancement, stochastic resonance can also be used to enhance muscle proprioceptive feedback. Early human¹⁹ and animal²⁰ experiments found that white noise tendon vibration increases the stretch-sensitivity of muscle spindles embedded in the vibrated musculotendon unit. Such vibration of ankle tendons can enhance the conscious detection of passive motion²¹ and decrease low-frequency sway during quiet standing.²²

A key factor in stochastic resonance is noise amplitude, which must be of sufficient magnitude to increase the likelihood of stimulus detection, but not so large that the underlying signal is masked.¹⁰ This amplitude dependence is often described as a “U-shaped curve,”²³ with varied noise levels able to either improve or impair performance. Many experiments intended to enhance tactile feedback from cutaneous sensors have set the noise amplitude just below the tactile sensory threshold^{14,17,24} with generally positive results. Other such studies have reported optimal effects with amplitudes ranging from 33–100% of sensory threshold.^{13,25,26} However, the present work seeks to enhance proprioceptive feedback related to changes in ankle angle rather than feedback related to tactile sensation. Therefore, setting the noise amplitude based on the conscious detection of vibration may be inappropriate. Instead, recent work focused on enhancing feedback from muscle spindles has set the vibration strength based not on a conscious detection threshold, but instead on root-mean-square (RMS) amplitude, with optimal reported effects at an amplitude of 30 μm .^{21,22}

The previous work investigating the use of stochastic resonance to enhance proprioception is promising, but an ultimate goal of this approach is to improve functional movement in clinical populations. In many functional tasks, individuals rely on processing proprioceptive feedback from more strongly contracting muscles.^{27,28} For example, plantarflexor muscle activity is increased during the type of active postural positioning that accompanies a forward reach.^{29,30} The plantarflexors are activated even more strongly during walking, but humans still appear to use plantarflexor proprioceptive feedback to economically adapt their gait pattern.³¹ It is presently unclear if stochastic resonant tendon vibration will have beneficial effects on proprioception during these more functional tasks, as tendon vibration tends to have reduced effects with increased mechanical demand.³² Additionally, it is unclear if the amplitude of vibration intended to enhance proprioceptive feedback should be chosen based on tactile sensory threshold (as with cutaneous sensors^{13,14,17,24–26}) or based

on RMS amplitude (as with muscle spindles^{21,22}). The present work experimentally addresses these gaps in the literature.

The purpose of this study was to determine whether white noise vibration of appropriate amplitude can consistently improve the performance of 2 distinct postural tasks in neurologically intact controls. We delivered vibration to the skin directly over the Achilles tendon, potentially influencing sensory feedback related to either skin stretch or length changes in the plantarflexor musculotendon complex. We tested whether white noise vibration improved the performance of quiet standing and active postural positioning, with a focus on anteroposterior sway. While both tasks involved postural control, differences in mechanical demand may influence the effects of enhanced sensory feedback. In half of the participants, we set vibration amplitude based on tactile sensory threshold (following previous methods for enhancing cutaneous feedback). In the other half of participants, we set vibration amplitude by RMS amplitude (following previous methods for enhancing proprioceptive feedback). We hypothesized that white noise vibration, either with amplitudes just below the sensory threshold or with an RMS value around 30 μm , would reduce postural sway during quiet standing, and reduce errors in active postural positioning.

Methods

Twenty young (25 ± 3 y; mean $\pm$ SD), right-leg dominant participants were randomly enrolled in either Group A ($n = 10$) or Group B ($n = 10$). All participants provided informed consent using a form approved by the Medical University of South Carolina Institutional Review Board and consistent with the Declaration of Helsinki.

In some experimental trials, vibration was applied using small mechanical devices (C-2 tactors; Engineering Acoustics Inc, Casselberry, FL, USA). In preliminary work, we quantified the relationship between applied white noise voltage signals and mechanical vibration output using a laser Doppler vibrometer (Optical Measurement Systems, Laguna Hills, CA, USA). We found that the tactors were relatively unresponsive to frequencies below ~ 30 Hz and exhibited a large amplitude peak at their resonant frequency of ~ 230 Hz. To achieve a relatively flat power spectrum (as required for true white noise), we applied a white noise voltage signal bandpass filtered between 30–100 Hz (Figure 1). Given the weak response of C-2 tactors to low-frequency signals, our applied vibration was likely quite similar to that previously used (C-2 tactors powered by a white-noise voltage signal low-pass filtered at 100 Hz) to enhance foot cutaneous sensation¹⁴ and proprioceptive feedback.²¹ The relationship between voltage magnitude and vibration RMS amplitude was highly linear ($R^2 > 0.99$) across the applied voltage range.

Several aspects of our protocol were consistent across Groups A and B. Participants stood on a force-plate (Bertec, Columbus, OH, USA) with their feet parallel and as close together as possible without touching, arms crossed, and wearing headphones to prevent auditory detection of vibration. Participants wore standardized socks and no shoes to avoid potential sway differences due to footwear. The tactors were strapped over the bilateral Achilles tendons approximately 4 cm above the ankle joint. For both Groups A and B, we first identified the sensory threshold for both legs as participants stood in the posture described

above with their eyes closed. We used a “method of limits” technique³³ in which 1-second periods of white noise vibration were delivered over the Achilles tendon, separated by 3–5 seconds. Participants were instructed to verbally indicate when they felt the vibration. The vibration RMS amplitude began at 75 μm (sensed by all participants) and was decreased in increments of 1.5 μm until the participant did not sense the vibration. Vibration amplitude was then increased (1.5 μm increments) until the participant sensed the vibration. This serial decreasing and increasing process was repeated 5 times, with the sensory threshold defined as the geometric mean of the endpoints of each series.³³

The sole difference between Groups A and B was how vibration amplitude was set during experimental trials. For Group A, vibration amplitude was set to 1 of 9 levels, based on the sensory threshold: No Vibration; 20%, 40%, 60%, 80%, 90%, 100%, 150%, and 200% of the sensory threshold (ST). These values were chosen in order to both span the wide amplitude range previously reported as optimal,^{13,25,26} and to test the effects of superthreshold noise thought to typically harm performance. For Group B, vibration amplitude was set to 1 of 9 levels of tactor RMS displacement: No Vibration, 5 μm , 10 μm , 15 μm , 20 μm , 30 μm , 40 μm , 50 μm , and 60 μm . These values were based on previous work^{21,22} that tested a relatively sparse set of vibration amplitudes (20, 30, 100, and 280 μm) and found beneficial effects only at the 2 smallest amplitudes. Through more systematic inclusion of low amplitudes, we sought to maximize our chances of identifying an amplitude that optimally improved postural function. Additionally, prior work has observed continuous Ia afferent firing in response to 80-Hz tendon vibration with a peak-to-peak amplitude of 100 μm (corresponding to RMS displacement of $\sim 35 \mu\text{m}$).³⁴ Therefore, our vibration amplitudes likely spanned the muscle spindle recruitment threshold.

Following sensory threshold identification, participants performed a series of 27 quiet standing trials while standing on a force plate (Bertec, Columbus, OH) that measured ground reaction forces and moments at a sampling rate of 1000 Hz. During these 30-second trials, participants were instructed to stand as still as possible in the posture described above with their eyes closed. Between trials, participants were given at least a 30-second rest break in which they could open their eyes and sway naturally. The 27 trials were organized in 3 blocks of 9 trials, 1 trial for each of the 9 vibration levels assigned for this experiment. Trial order within each block was randomized.

After the quiet standing trials, participants completed a series of 27 postural positioning trials designed to increase reliance on the sensed sagittal plane ankle angle. Participants again stood on the force plate, while a custom written LabView program was used to calculate center of pressure (CoP) location in near real-time. A monitor able to display CoP location in graphical form was placed in front of participants (eye level; 0.5 m anterior). Participants were instructed to sway forward and backward using only their ankle joints, keeping their hips and spine in a neutral position. We first established the limits of anterior and posterior CoP motion (CoP_{ANT} and CoP_{POST}) by instructing participants to sway as far as possible forward and backward with their feet remaining fully on the ground (monitored by an experimenter). For subsequent trials, we used a constant posterior target located at 20% of the distance from CoP_{POST} to CoP_{ANT}. We used anterior targets that varied from trial to trial, located at 40%, 60%, or 80% of this distance (Figure 2A).

The sequence of events for postural positioning trials is illustrated in Figure 2B. First, the CoP location and posterior target were displayed, and participants moved their CoP within the target. Once the anteroposterior CoP location remained within 5 mm of the target's center for 3 seconds, the target relocated to an anterior target location. All anterior target locations appeared the same on the monitor, so participants did not have visual feedback of the prescribed sway distance. Participants were instructed to move their CoP within the new anterior target, and remember this posture. After 3 seconds within 5 mm of this anterior target, the target moved back to its posterior position. Participants again swayed to match their CoP location to this target. Once the anteroposterior CoP location remained within 5 mm of the target's center for 3 seconds, the real-time CoP location disappeared and the target moved to its anterior location. Participants were instructed to sway to this location (without visual feedback of their CoP), and press a hand-held button when they perceived they were in the correct location. By quantifying positioning accuracy in this forward-leaning posture, we sought to maximize the likelihood that participants would rely on feedback from the active, lengthened plantarflexors. All participants were given at least 2 practice trials prior to data collection to ensure they understood the task. Laser lines marked the boundaries of the feet, so participants could move around during the rest periods before returning to their baseline posture. The 27 postural positioning trials included 3 trials at the 9 vibration levels in randomized order, reducing the risk of the results being influenced by a training effect. For each vibration level, 1 trial was performed to each of the anterior sway targets.

During quiet standing trials, our analyses focused on anteroposterior CoP motion, which is strongly influenced by plantarflexor proprioceptive perturbations.³⁵ We did not include measures of mediolateral sway, which would likely be more influenced by vibration of the medial or lateral ankle tendons.³⁵ Our primary measure of sway was the mean absolute value of the CoP anteroposterior velocity. Secondarily, we calculated the spectral power density of the anteroposterior CoP location signal, a metric used previously to quantify postural sway.²² Briefly, we calculated the signal power across all frequencies up to the Nyquist frequency, and then calculated the average power within the frequency range of primary interest (0.05–0.5 Hz). This frequency range contains the vast majority of the total spectral power for anteroposterior sway, and can be influenced by white noise ankle tendon vibration.²²

For postural positioning trials, we quantified the absolute value of the error in anteroposterior CoP location when the push-button was pressed (see Figure 2B). It should be noted that quantifying error only at this instant could introduce error due to ongoing sway. However, the typical magnitude of this sway is only 5 mm (standard deviation in anteroposterior sway),³⁶ substantially smaller than the average targeted CoP displacements (25, 52, and 79 mm). During this portion of the trial, visual feedback of CoP location was not provided, so participants were forced to rely on other sources of feedback, likely including those providing information related to ankle angle (eg, plantarflexor proprioception; posterior ankle skin stretch).

To quantify the effects of white noise Achilles tendon vibration of discrete amplitudes, we used paired *t*-tests to compare performance during No Vibration trials with performance at

each vibration amplitude level. We thus performed 8 paired *t*-tests (corresponding to the 8 vibration amplitudes in each experiment) for each metric. We used Bonferroni corrections to account for multiple comparisons, resulting in a required alpha value of .0063 for statistical significance. Statistical analyses of Groups A and B were performed separately. Estimated effect sizes of interest are provided as Cohen's *d*.

Results

Participants in Groups A and B were similar in terms of their basic demographics, anthropometric characteristics, and average sensory thresholds (Table 1). Based on the group average sensory threshold in Group A, the corresponding vibration RMS amplitudes set across trials ranged from approximately 4–40 μm , similar to the amplitudes applied in Group B. However, these amplitudes varied across participants, based on individual sensory thresholds.

During quiet standing, white noise Achilles tendon vibration with a wide range of amplitudes had no clear effects on CoP absolute velocity. The expected U-shaped pattern was not observed when amplitude was set in terms of either sensory threshold (Group A; Figure 3A) or RMS displacement (Group B; Figure 3B). While not significant by our relatively conservative criteria, there was a trend ($P = .03$; Cohen's $d = -0.82$) for sway to be reduced by 200% ST amplitude vibration in Group A. Similarly, in Group B there was an insignificant trend ($P = .02$; Cohen's $d = -0.92$) for reduced sway with 40 μm amplitude vibration.

Tendon vibration also had no clear effects on slow postural sway, as quantified by mean spectral power density in the 0.05–0.5 Hz frequency band. Again, a U-shaped pattern was not apparent for either Group A (Figure 3C) or B (Figure 3D). None of the individual comparisons indicated a statistically significant effect of vibration, although there was an insignificant trend ($P = .02$; Cohen's $d = -0.91$) for a reduction in spectral power with a vibration amplitude of 60% S.T. in Group A. In Group B, no trends for improved sway performance were observed for any vibration RMS amplitude ($P > .12$ for all comparisons).

The effects of vibration during active postural positioning were more apparent when amplitude was set in terms of RMS displacement. In Group A, there were no clear effects of any of the vibration amplitudes on accuracy ($P > .13$ for all comparisons) (Figure 4A). In Group B however, these average error values across participants took the form of the expected U-shape curve across vibration amplitudes (Figure 4B). Vibration with an amplitude of 30 μm significantly ($P = .001$; Cohen's $d = -1.43$) reduced absolute error, as all 10 participants exhibited improved accuracy. None of the other vibration amplitudes caused clear improvements in positioning accuracy ($P > .05$ for all comparisons).

While not part of our original statistical plan, we performed 2 additional analyses to further investigate the observed improvement in active postural positioning in Group B. Specifically, we quantified the average directional error (ie, mean signed error) for each vibration amplitude, providing insight into whether participants consistently overshoot or undershot the target. We also quantified the variable error for each amplitude, calculated as the standard

deviation in directional error across individual trials. Directional error was not significantly influenced by any vibration amplitude ($P > .10$ for all comparisons) (Figure 5A). In contrast, variable error was significantly reduced with 30 μm vibration ($P = .01$; Cohen's $d = -1.12$), with a trend for reduced variable error with 20 μm vibration ($P = .01$; Cohen's $d = -1.07$) (Figure 5B).

Discussion

This study investigated whether white noise Achilles tendon vibration improved the performance of 2 standing posture tasks. As hypothesized, 30 μm amplitude vibration consistently improved active positioning accuracy during a task designed to rely on sensed ankle angle. However, positioning accuracy was not improved by vibration with amplitudes based on sensory threshold. Also contradicting our hypotheses, no tested vibration amplitude consistently reduced sway during a quiet standing task.

We found no convincing evidence that appropriate amplitude vibration reduced quiet postural sway (quantified using CoP velocity), whether amplitude was set based on sensory threshold or RMS amplitude. While we observed insignificant trends for reduced sway with amplitudes of 200% sensory threshold and 40 μm RMS (~200% of the average sensory threshold), future studies with either larger sample sizes or fewer tested amplitudes would be required to support this result. Our generally negative result matches the recent finding of Borel and Ribot-Ciscar,²² in which ankle tendon vibration did not significantly influence a traditional measure of quiet postural sway. While commonly used, such traditional measures may not accurately reflect beneficial changes in postural control.³⁷

A more complex metric of quiet sway also did not reveal clear effects of tendon vibration. No vibration amplitude significantly influenced the mean spectral power density of anteroposterior CoP motion in the 0.05–0.5 Hz frequency range. Our negative result contradicts the recent finding of Borel and Ribot-Ciscar²²; while this prior work reported a significant beneficial effect of 30 μm vibration, we found no such improvement ($P = .77$; Cohen's $d = -0.26$). One possible explanation for this discrepancy is that we applied vibration only over the Achilles tendon, instead of both the plantarflexor and dorsiflexor tendons.²² During quiet standing, the relatively silent dorsiflexors may be a more useful source of proprioceptive feedback than the actively contracting plantarflexors, as dorsiflexor muscle spindle activity may more accurately reflect ankle position changes without the complicating influence of muscle contractions.³⁸ Therefore, enhanced dorsiflexor proprioceptive feedback may cause greater improvements than enhanced plantarflexor proprioception alone. However, the previous finding that sub-threshold white noise electrical stimulation can reduce postural sway when delivered over the dorsiflexors alone, plantarflexors alone, or dorsiflexors and plantarflexors combined³⁹ would seem to contradict this explanation. Alternatively, the observed discrepancy may be due to differences between our applied vibration bandwidth (30–100 Hz) and that used previously (100–300 Hz),²² as cutaneous receptors can be quite responsive to vibration in this higher range.⁴⁰ Finally, our instructions to “stand as still as possible” may have caused participants to intentionally reduce their baseline sway levels in comparison to simply “standing straight”,²² limiting the possible beneficial effects of vibration.

Unlike during quiet standing, appropriate amplitude vibration had clear beneficial effects on the performance of an active postural positioning task. Accuracy was influenced by vibration RMS amplitude, as evidenced by the expected U-shaped curve and significant improvement with 30 μm vibration, the same amplitude found to improve motion detection²¹ and quiet standing.²² The observed errors were less variable when appropriate amplitude vibration was applied, suggesting that improvements may be attributed to more reliable interpretation of the available sensory feedback. This improvement was consistently present (10/10 participants) despite the increased complexity of relating sensory feedback to task performance during active, dynamic movement.^{22,41} The relatively large effects in comparison to during quiet standing may be due to the more explicit performance goal,⁴² or altered plantarflexor fusimotor activity caused by increasing attention paid to the task.^{36,43}

The clearer effects of tendon vibration during active postural positioning than during quiet standing may initially appear to contradict the previously reported ineffectiveness of mechanical noise under challenging postural conditions.²² However, we designed our postural positioning task explicitly to encourage reliance on sensed ankle angle; accuracy was measured during anterior leaning, when plantarflexor length would be most closely related to CoP location. In contrast, the previous methods of challenging postural control required participants to stand on foam or on a moving platform.²² Both of these methods likely reduced the usefulness of ankle proprioceptive feedback, as humans' use of ankle proprioception is downweighted when standing on an unstable surface.^{44–46} The present results are consistent with enhanced proprioceptive feedback having the greatest effect on posture for contexts in which this source of sensory information is relied upon.

While setting vibration strength by RMS amplitude had the expected effects on postural positioning, this was not the case when vibration strength was based on sensory threshold. Essentially, methods previously used to enhance proprioceptive feedback^{21,22} were effective, while methods previously used to enhance cutaneous feedback^{13,14,17,24–26} were not. This finding is consistent with the possibility that tactile feedback from the skin around the ankle is less useful than proprioceptive feedback from the plantarflexors in controlling anteroposterior sway. However, our results should be interpreted cautiously, as we did not directly monitor muscle spindle or cutaneous feedback using microneurographic methods.²¹ Our findings may instead be due to the use of a vibration frequency bandwidth that preferentially excites muscle spindles, although white noise vibration in a similar frequency range (below 100 Hz) has previously been reported as effective in enhancing cutaneous feedback.¹⁴ Alternatively, it is possible that our reliance on self-report psychophysical methods for identifying the cutaneous threshold may have introduced problematic variability into the selection of vibration amplitudes. Indeed, the vibration amplitudes based on tactile thresholds often differed between the left and right legs, which may have made accurate positioning more difficult.

Beyond our inability to definitively attribute our results to either proprioceptive or cutaneous enhancement, several other limitations of the present work influence our interpretations. Our focus on CoP prevents insight into joint-level control that may be possible with kinematic measures. For example, providing visual feedback of ankle angle instead of CoP would reduce the risk that participants were using a hip strategy as a major component of their

anteroposterior sway. Additionally, the present study focused on healthy, young controls, and thus cannot provide direct insight into the effects of white noise vibration in older adults or individuals with neurological injuries.

Future work will extend these results to the investigation of white noise tendon vibration during more demanding functional tasks, and the investigation of potential benefits for clinical populations. We have previously proposed that humans use proprioceptive feedback to identify metabolically economical movement patterns,⁴⁷ and demonstrated that disruption of plantarflexor proprioception slows adaptation toward an economical gait pattern.³¹ Future experiments will use the present methods to enhance plantarflexor proprioception, testing whether such methods can increase the rate of gait adaptation and reduce metabolic cost. We also plan to extend our investigations into clinical populations with reduced proprioceptive function due to peripheral deficits (eg, diabetic neuropathy⁵) or central deficits (eg, stroke⁴). Rehabilitation methods able to enhance the use of residual proprioceptive feedback during active movement may have high value for restoring function.

In conclusion, the present experiments have demonstrated that Achilles tendon vibration of appropriate amplitude can improve the accuracy of an active postural positioning task. Importantly, the postural task was designed to rely on proprioceptive feedback from the active plantarflexors, demonstrating that the beneficial effects of tendon vibration can be retained with increased muscular demands. The potential improvement of residual proprioceptive feedback during active movement may have high value for restoring functional mobility among clinical populations.

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Figure 1 —.
The vast majority of the mechanical vibration power was within the 30–100 Hz frequency range (shaded area), as illustrated using a power spectrum.

Figure 2 —.

We designed an active postural positioning task to encourage reliance on plantarflexor proprioception. Targets for CoP location (A) were evenly spaced between each participant's limits of achievable anterior and posterior CoP motion (CoP_{ANT} and CoP_{POST}) using an ankle strategy. Participants sequentially swayed to a series of posterior and anterior targets, schematically illustrated in panel (B) as the visual monitor display (top row) and the underlying anterior CoP motion (bottom row). For the first 3 targets (i, ii, and iii), real-time visual feedback of CoP location was available to participants. For the final anterior target (iv) this feedback was removed, and we calculated the postural positioning error between the target and actual anteroposterior CoP locations.

Figure 3 —.

Sway during quiet standing was not significantly affected by vibration at any tested amplitude. Absolute CoP velocity did not change significantly when vibration amplitude was set as a percentage of sensory threshold (A) or by RMS amplitude (B). Similarly, mean spectral power (s.p.) in the 0.05–0.5 Hz frequency range did not change significantly for vibration amplitudes set by sensory threshold (C) or RMS amplitude (D). Data are plotted as the change from the baseline no vibration condition, with data points representing means and error bars representing 95% confidence interval. Negative values indicate decreased sway. Baseline values are presented on each panel as mean (standard deviation). Pound signs (#) indicate an insignificant trend ($P < .05$) for an effect of vibration.

Figure 4 —.

Improvements in active postural positioning accuracy were dependent on the set vibration amplitude. Accuracy was not significantly influenced by any tested vibration amplitude when these amplitudes were set as a percentage of sensory threshold (A). Accuracy was significantly and consistently improved when vibration amplitude was set to 30 μm (B). Data are plotted as the change from the baseline no vibration condition. Negative values indicate smaller errors between the target position and the actual position. Baseline values are presented as mean (standard deviation). Data points represent means, error bars represent 95% confidence interval, and the asterisk (*) indicates a significant difference from baseline ($P < .01$).

Figure 5 —.

Improved active postural positioning was a result of reduced variability. Directional error was not significantly influenced by any tested vibration amplitude (A). Error variability was significantly reduced when vibration amplitude was set to 30 μm (B). Data are plotted as the change from the baseline no vibration condition. Baseline values are presented as mean (standard deviation). Data points represent means, error bars represent 95% confidence interval, the asterisk (*) indicates a significant difference from baseline ($P < .01$), and the pound sign (#) indicates an insignificant trend ($P < .05$).

Table 1

Comparison of Participant Characteristics in Groups A and B

	Group A	Group B	<i>P</i>-value
Gender	9F/1 M	7F/3M	.58
Age (y)	24 ± 2	25 ± 4	.22
Height (cm)	168 ± 8	171 ± 11	.63
Mass (kg)	67 ± 9	65 ± 15	.84
Sensory Threshold (µm)	19 ± 10	18 ± 8	.78

Note. The presence of statistically significant differences between groups was tested using Fisher's exact test (for gender) and unpaired *t*-tests (for other characteristics). Data are presented as mean ± standard deviation.