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Tactile stimuli affect long-range correlations of stride interval and stride length differently during walking

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Abstract

Sensory feedback below the sole of the foot using sub-threshold mechanical noise significantly reduced postural sway in patients with diabetes and stroke. However, the effects of tactile parameters on walking are still elusive. Specifically, the effects of such parameters on human gait variability need to be studied because of possible rehabilitation outcomes in terms of bringing improvement in temporal and spatial gait parameters. The purpose of this study was to investigate whether different frequency and amplitude combinations of vibro-tactile stimulation of feet would affect stride-to-stride variability in healthy young adults. Ten healthy subjects walked on a treadmill at self-selected pace while wearing customized insoles fitted with tactors that vibrated at selected frequencies and amplitudes. The results show that the frequency manipulations of tactile stimulation altered the long-range correlations (LRCs) in stride length while amplitude manipulations affected the LRCs in stride interval without having any effect on the amount of gait variability. Our findings suggest that independent neural mechanisms may be responsible for coordinating LRCs of gait parameters in the spatial and temporal domains.

Keywords

Tactor; Detrended fluctuation analysis; Variability; Locomotion; Somatosensory

Introduction

Sensory feedback is crucial for postural control during standing and walking. This includes visual, vestibular, and proprioceptive feedback. Previous studies have shown that perturbation of sensory feedback through the application of sub-threshold mechanical noise below the foot significantly reduced postural sway during standing in older adults (Priplata

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et al. 2003), and in patients with diabetes and stroke (Priplata et al. 2006). Moreover, when sensory feedback below the sole of the foot was removed by injecting an anesthetic in healthy young subjects, the base of support and the margin of stability which is an indicator of dynamic stability during walking, were significantly altered (Hohne et al. 2011).

When healthy young subjects walked with reduced sensory feedback after immersing the feet in an ice bath, there was no significant alteration in spatial gait parameters, however, temporal gait variability increased (Sawa et al. 2013). Therefore, tactile feedback may be important for the temporal components of stride-to-stride variability during walking. Moreover, the application of a sub-threshold mechanical noise on the sole of the foot during walking in healthy older adults had no effect on the stride-to-stride variability of stride length although step width variability was reduced (Stephen et al. 2012). Interestingly, a similar sensory feedback manipulation in fallers also resulted in a significant reduction in the stride-to-stride variability in stride interval (Galica et al. 2009). However, in patients with Parkinson's disease, the application of a 70 Hz supra-threshold mechanical noise on the foot sole during the stance phase resulted in a significantly decreased stride-to-stride variability not only in the temporal domain (stride interval) but also in the spatial domain (stride length) in comparison to age-matched healthy controls (Novak and Novak 2006). It is possible that the discrepancies in the results from the different studies were because of the different tactile parameters used.

Indeed, results from static standing studies have shown that manipulating different tactile parameters have characteristic effects. For example, when a supra-threshold mechano-vibration is used to stimulate the sole of the foot, postural sway is increased in healthy young adults (Kavounoudias et al. 1999), however, when such stimulation is provided at sub-threshold levels, postural sway is reduced in healthy and pathological subjects (Priplata et al. 2003, 2006). Moreover, stimulating at different locations on the sole of the foot, the direction and amplitude of body sway can also be affected. Applying the stimulus on the anterior sole, the body sway is directed posteriorly and vice versa. If stimulus is only applied on the right sole, the body sways towards the left and vice versa (Kavounoudias et al. 1999). In addition, administering supra-threshold mechano-vibration stimulus on the sole complemented the ankle muscle inputs during quiet standing to maintain an erect stance (Kavounoudias et al. 2001). It is therefore clear that tactile parameters are important for postural control during standing. However, such effects during walking still remain to be elucidated. Specifically, the effect of such parameters on human gait variability remains to be investigated.

Human gait variability patterns have been shown to have fractal-like properties including characteristic self-similar and emergent properties. One consequence of the self-similar characteristic is that gait variability shows distinct Long-Range Correlations (LRCs). This means that stride-to-stride fluctuations in gait parameters at any given moment are statistically related to those that occur over many different time scales (Hausdorff et al. 1995). Moreover, pathology of the CNS has also been demonstrated to affect such changes during gait. The strength of LRC of gait variability was significantly smaller in older adults, in patients with Huntington's and Parkinson's diseases than in healthy controls (Hausdorff et al. 1997; Delignières and Torre 2009).

When there are alterations in task constraints (e.g., visual stimuli, auditory stimuli, and gait speed changes), we observe changes in the LRC of stride-to-stride fluctuations (Jordon et al. 2007; Katsavelis et al. 2010; Kaipust et al. 2013; Hunt et al. 2014). For example, auditory stimulation, specifically, changes in the temporal properties of the auditory stimulus (e.g., inter stimulus intervals) caused characteristic changes in the LRC of gait variability in healthy young and older adults (Kaipust et al. 2013; Hunt et al. 2014). Another example is the effect of changing gait speed on LRC of gait parameters. A U-shaped relationship was determined where the lowest LRC existed at the preferred walking speed (PWS) and LRC became stronger both above and below the PWS indicating a tighter control of the stride-to-stride fluctuations when task constraints became more complex (Jordon et al. 2007). These studies demonstrate that with changes in age or with gait speed, there is a shift in the long-range relationships between gait events that in turn expresses a shift between more constricted (higher LRC with reduced degrees of freedom) and less constricted (lower LRC with increased degrees of freedom) gait. The strength of LRC, therefore, can reveal the adaptive flexibility that exists in human sensorimotor control and how it is impacted by task constraints, aging, and disease.

The purpose of this study was to investigate whether different frequencies and amplitudes of vibro-tactile stimulation affected the LRC of gait variability in healthy young adults. In this study we explored LRC using Detrended Fluctuation Analysis (DFA). We hypothesized that first, different frequency and amplitude combinations of vibro-tactile stimulation would affect stride-to-stride variability. Second, such combinations of stimulations will also affect the strength of LRC.

Methods

Ten healthy adults were informed about the experimental protocol, and signed a consent approved by the Institutional Review Board of the University of Nebraska Medical Center. At the beginning of the study, all participants walked on a treadmill (Bertec Corp. Columbus, OH) for a 5-min period of familiarization. For the tactile simulation conditions, subjects wore a tactor control unit on their back secured by a belt. This unit sent voltage signals to tactors that vibrated the insoles of the subject's feet (Fig. 1). Six C2 tactors (Engineering Acoustics, Casselberry, FL), embedded in customized insoles at two sites such that they made contact with the ball of the foot and the heel, were made to vibrate at selected frequencies (30, 150, 250 Hz) and amplitudes (8, 12, 17.5 db). Ten walking conditions in this study (3 frequency $\times$ 3 amplitude conditions plus one control/no stimulus trial) were randomly assigned to subjects. Determination of self-selected pace: Prior to the data collection each subject walked for 5 min on the treadmill to determine their self-selected pace. First, the subjects stood on the side of the treadmill. Subsequently, the treadmill belt velocity was incremented from 0 to 0.8 m/s. Then the subject was asked to step on the treadmill while holding the handrail. After the subject started walking on the treadmill, experimenters asked the subject to assess if the treadmill speed was comfortable and similar to walking around a grocery store. The treadmill speed was either increased or decreased based on the subject responses. Once a comfortable self-selected pace was attained, the subject walked continuously for 5 min at this pace for familiarization. Subjects walked at their self-selected pace i.e., PWS for 5 min in each condition. The reason why we selected

the subject's PWS was for two reasons: first, we wanted to study the subject's natural walking pattern. Second, it has been shown that LRC for the PWS is different from gait speeds higher or lower than PWS (Jordon et al. 2007). In this situation fixing the speed would have resulted in the fixed gait speed being higher than PWS for some subjects and lower than PWS for others. This consequently would have affected the LRC analysis for the different subjects. The participants were given the option to rest between conditions. Participants were also instructed to maintain a steady gaze in front of them at their eye level during the walking trials.

A visual analogue scale (VAS) (della Volpe et al. 2006) was adopted to quantify each participant's perception of tactile sensation. This tactile perception test was performed during three separate postural configurations: two-legged standing, one-legged standing, and sitting with feet raised above the floor. These configurations represented different weight-bearing conditions for the foot that could affect tactile perception: full, partial, and no weight bearing. The test was done in this way to make subjective assessments more uniform since such tests during walking are not homogenous (Gravano et al. 2011). They are affected by different foot regions in contact as well as different gait phases (stance, swing, double support) with differing weight-bearing characteristics. The tactile stimulations that were generated by six C2 tactors during the perception test had the same frequency and amplitude combinations as in the walking trials (3×3). The maximum frequency–amplitude combination was 250 Hz and 22.5 db, respectively. The subjects were made to perceive this condition and rank it on the VAS as 100. Subjects were instructed to rank 0 on the VAS scale for any condition where they could not perceive any tactile vibration.

Three-dimensional kinematic data were collected using the NDI motion capture system (Northern Digital Inc, Waterloo, Canada) at 100 Hz. Two rigid bodies with three active infrared markers on each rigid body were attached on the lateral side of both feet. The bony landmarks of each foot (heel, toe, 1st and 5th metatarsal) were digitized with respect to these two rigid bodies. Leg dominance was determined by asking the subject about their preferred leg for kicking a ball. The gait characteristics of the dominant leg were processed and calculated by a custom-written MatLab code (MathWorks Inc., Natick, MA). The kinetics data for the right leg was collected by an instrumented treadmill at 300 Hz (Bertec Corp. Columbus, OH), down-sampled, and synchronized with kinematics data in order to calculate the instant of heel contact. Heel contact was considered to occur at the first frame in which the vertical component of the ground reaction force exceeded a threshold level of 10 N and continuously exceeded this threshold for 40 ms. Toe off was considered to occur at the first frame in which the vertical component of the ground reaction force fell below the 10 N threshold, and was sustained continuously for 40 ms. This 10 N threshold was calculated as three times the standard deviation of the vertical ground reaction force during the initial 100 ms (100 frames) of the trial (Mickelborough et al. 2000; Zeni et al. 2008; Kiss 2010).

Stride length was defined as the length between two consecutive heel contacts of the same leg and was determined by adding the horizontal distance from heel contact to toe off and the horizontal distance from toe off to the subsequent heel contact of the same leg. Stride interval was defined as the time between consecutive heel contacts of the same leg. Stride-

to-stride variability was quantified using standard deviation (amount of gait variability) and LRC (structure of gait variability) for stride length and stride interval.

The LRC was measured using DFA. This is a modified random-walk analysis that makes use of a long-range correlated time series. The LRC can be mapped to self-similar calculations through simple integration. First, the time series is integrated and then divided into window sizes of length n . A least squares fit line is fit to the data in each window, and the data are detrended by subtracting the integrated time series from the least squares fit line. Root mean square is then calculated for each window and summed for the entire time series, $F(n)$. The process is repeated with smaller and smaller window sizes. Finally, the $\log F(n)$ is plotted against the $\log n$ (the root mean square versus the window sizes). The slope of this plot is the reported α -value. If the α -value is greater than 0.5, the LRC is positively persistent which means that increases are followed by increases and decreases are followed by decreases. Whereas if the α -value is smaller than 0.5, the LRC is anti-persistent, meaning that increases are followed by decreases and vice versa. The range of window sizes for the DFA selected in the current study was from $n_1 = 4$ to $n_m = N/4$, where N is the number of stride intervals (Chien et al. 2016).

Statistical analysis

Separate two-way repeated measure ANOVAs were applied using SPSS 18.0 (IBM Corp., Armonk, NY) to examine the effect of amplitude and frequency on the standard deviation and the strength of LRC in stride interval and stride length time series. Post hoc tests with Bonferroni corrections were performed if a significant interaction was found. The level of significance was set at 0.05. In addition, separate two-way repeated measure ANOVAs were applied to examine the amplitude and frequency effects on the tactile sensory perception test during the two-legged standing, one-legged standing, and sitting conditions. Post hoc tests with Bonferroni corrections were performed if a significant interaction was found.

Results

Subject demographics

There were ten healthy young adults (four female) in this study. The average age was 24.7 ± 4.7 years. The average height was 1.76 ± 0.08 m. The average mass was 73.2 ± 16.19 kg. In addition, the average PWS was 0.97 ± 0.18 m/s.

Visual analogue scale

In each of three postural conditions, statistical analysis gave similar results. There was a significant main effect of amplitude (two-legged standing: $F = 37.43$, $p < 0.0001$; One-legged standing: $F = 223.28$, $p < 0.0001$; sitting: $F = 70.45$, $p < 0.0001$) and frequency (two-legged standing: $F = 50.13$, $p < 0.0001$; One-legged standing: $F = 133.36$, $p < 0.001$; sitting: $F = 210.35$, $p < 0.0001$) (Fig. 2). For each posture, the amplitude effect was because the 17.5 db conditions caused higher VAS scores than 8 and 12 db conditions. Also, for each posture, the frequency effect was because the 250 Hz conditions caused higher VAS scores than 30 and 150 Hz. There was a significant interaction effect between frequency and amplitude for each postural condition (two-legged standing: $F = 14.68$, $p < 0.0001$; One-legged standing: F

= 30.62, $p < 0.0001$; sitting: $F = 41.10$, $p < 0.0001$). This was because the VAS at 250 Hz/17.5 db was higher than all other frequency-amplitude combinations except the 150 Hz/17.5 db combination.

Spatial domain of gait variability

There was no significant main effect of either amplitude or frequency of tactile stimulation on the standard deviation of stride length. In addition, there was no significant interaction between amplitude and frequency for the standard deviation of stride length (Table 1).

There was a significant main effect of frequency of tactile stimulation on the LRC of stride length ($F = 8.863$, $p = 0.002$). The 250 Hz conditions had significantly stronger LRC than the 30 Hz ($p = 0.041$) and the 150 Hz conditions ($p = 0.006$). There was no significant interaction between frequency and amplitude on the LRC of stride length (Table 2).

Temporal domain of gait variability

There was no significant main effect of either amplitude or frequency of tactile stimulation on the standard deviation of stride interval. In addition, there was no significant interaction between amplitude and frequency on the standard deviation of stride interval (Table 3).

For LRC, there was a significant main effect of amplitude of tactile stimulation ($F = 14.87$, $p < 0.001$), but no significant main effect of frequency ($F = 1.755$, $p = 0.205$). Post hoc tests showed that the amplitude effect was because strength of LRC was significantly lower at 12 db than at 8 db ($p = 0.05$) and at 17.5 db ($p < 0.0001$). There was a significant interaction between amplitude and frequency of tactile stimulation ($F = 6.123$, $p = 0.001$). For LRC, the general distribution of strength of LRC (α -value) of stride interval is around 0.6 to 1.0 (Table 4).

Moreover, a significantly higher strength of LRC at 250 Hz/12 db and a significantly lower strength of LRC at 250 Hz/17.5 db were determined in comparison with the no stimulus condition ($\alpha = 0.83$). These mean differences are shown in Fig. 3 and LRCs of stride interval in one participant for the different tactile conditions are shown in Fig. 4.

Discussion

We had hypothesized that first, different frequency and amplitude combinations of vibro-tactile stimulation would affect stride-to-stride variability. Second, such combinations of stimulations would also affect the strength of LRC. The results supported our hypotheses partially. We found that altering the frequency of vibro-tactile stimulation affected the strength of LRC in the spatial domain and manipulating the amplitude affected the strength of LRC in the temporal domain. However, frequency or amplitude conditions didn't affect the standard deviation of stride-to-stride variability in either the spatial or temporal domains.

Tactile manipulations affected long-range correlations during gait

Our study shows that it is possible to use tactile stimulus below the feet to either reduce the strength of LRC below that at the PWS or to increase it above that at the PWS. While the strength of LRC increased significantly in the 250 Hz/12 db, condition, it decreased

significantly in the 250 Hz/17.5 db condition in comparison with the no stimulus condition. It has been shown previously that motor response can be directionally driven in a similar manner through structured auditory stimulation (Kaipust et al. 2013; Hunt et al. 2014) and structured support surface oscillations (Rand et al. 2015). Those studies, through the use of entrainment to structured sensory stimuli, demonstrated the adaptive flexibility of the sensorimotor system to switch between task constraints that require shifts in the LRC of gait parameters. In our study, we demonstrated such sensorimotor flexibility during gait when tactile stimulations below the feet differed in amplitude and frequency.

The strength of LRC depends on the attractor dynamics of gait (Jordan et al. 2007) and the characteristic LRC shifts underpin the search to find the nearest attractor that provides stability and efficiency with low energy costs (Brisswalter and Mottet 1996) given the task, environmental, and individual constraints. It has been shown that at PWS, the LRC is the lowest allowing for maximum flexibility with high degrees of freedom while both above and below this speed, LRC increases, restricting the degrees of freedom and consequently reducing movement flexibility (Jordan et al. 2007). There have been suggestions that such ubiquitous observations reflect the intrinsic timing characteristics of human movement (Riley and Turvey 2002), a state of optimal movement variability (Harrison and Stergiou 2015; Stergiou and Decker 2011), or an optimal state of coordination of the degrees of freedom across the individual and task environments (Van Orden et al. 2009).

Tactile manipulations did not affect the magnitude of variability during gait

The application of sub-threshold stimulation in healthy older and older fallers (Galica et al. 2009; Stephen et al. 2012), and supra-threshold vibro-tactile stimulation on the foot in healthy middle-aged adults (Novak and Novak 2006) have been shown to reduce the magnitude of gait variability. However, in the current study, different combinations of frequencies and amplitudes of vibro-tactile stimulation that included both sub- and supra-threshold stimuli, didn't affect the magnitude of gait variability in healthy young adults. There may be several reasons for this. First, in comparison to previous studies that have shown a reduction in gait variability due to insole vibration predominantly in an older age group: patients with Parkinson's Disease and age-matched controls (Novak and Novak 2006), healthy older adults (Stephen et al. 2012), older fallers, and age-matched controls (Galica et al. 2009), our sample was relatively younger. This could be due to the decrements in walking performance resulting from the age-related deterioration in somatosensation (Mold et al. 2004; Deshpande et al. 2008; Cruz-Almeida et al. 2014). Second, the type of stimulation may have played a role because gait variability was shown to be increased after foot-cooling (Sawa et al. 2013). In the current study, while the stimulation did not make the feet anesthetic, it made tactile perception noisy. Third, we used a fixed-stimulus paradigm for the different phases of gait as opposed to other studies (Galica et al. 2009; Stephen et al. 2012) where different stimulus amplitudes were provided during the different phases of gait cycle. The use of continuous sensory stimuli also has evidence from previous studies (Katsavelis et al. 2010). In that study the magnitude of variability during treadmill walking did not show differences when the speed of optic flow remained the same within each condition and changed between conditions. Finally, the application of tactile stimulation for 5 min in each trial may have caused the system to adapt to the stimulus thereby having no

effect on the magnitude of variability. In a previous study (Katsavelis et al. 2010), we have observed that when optic flow speed was manipulated in such long trials, the magnitude of variability did not change. However, in the same study, changes in LRC were demonstrated as a result of the sensory manipulations. We also demonstrated similar results when the effects of auditory stimulation on gait variability were studied (Kaipust et al. 2013). When walking to different structures of auditory stimuli, healthy young adults showed changes in LRC but not in the magnitude of gait variability.

Separate control for spatial and temporal parameters of gait variability

Our results showed that the frequency manipulations of tactile stimulation affected the long-range correlations in stride length while amplitude manipulations affected the long-range correlations in stride interval without having any impact on the amount of gait variability. Although in general terms there is a relationship between spatial and temporal gait parameters, being controlled by the same CNS, there is evidence that independent control exists for these parameters. Such independent control has also been demonstrated in neuronal populations in the spinal cord (Lafreniere-Roula and McCrea 2005; McCrea and Rybak 2008). Hemispherectomy patients showed a preservation of spatial but not temporal adaptability on a split-belt treadmill task (Choi et al. 2009) which indicate distinct neural mechanisms modulating these two aspects of gait. While cerebellar structures are implicated in spatial control, both cerebellar and cortical structures have importance in temporal control of gait (Choi et al. 2009; Bretzner and Drew 2005). In addition, specific tasks/conditions may affect spatial and temporal parameters independently—for example, on a split-belt treadmill, metabolic power was significantly associated with step length asymmetry but not with step time asymmetry (Finley et al. 2013). Visual perception of self-motion provided during split-belt adaptation showed preferential effects on spatial components of gait but not temporal components (Finley et al. 2014). During spatial navigation, anticipatory orienting movements are driven preferentially by spatial rather than temporal cues (Grasso et al. 2000). Additionally, conscious processes such as voluntary conscious corrections or distractions preferentially impact spatial components over temporal (Malone and Bastian 2010). Taken together, all these separate findings implicate the independent control of spatial and temporal components of gait. Our findings suggest that such characteristics may also underlie the control of long-range correlations of gait parameters. Such distinctions in locomotion control may have implications on rehabilitation in terms of bringing about improvement in temporal/spatial aspects of locomotion by manipulating amplitude/frequency of sensory stimuli.

The effect of tactile stimulation on the neural control of gait variability

In addition, we have previously shown that tactile stimulation at 250 Hz and 17.5 db leads to enhanced transfer effects in a split-belt adaptation task (Mukherjee et al. 2015). This was speculated to be due to increased proprioceptive gain (in response to noisy tactile feedback) through multisensory integration (Campos et al. 2014; Green and Angelaki 2010). Therefore, in the current study, we attribute the reduced LRC during preferred walking gait at the supra-threshold tactile stimulation of 250 Hz and 17.5 db to increased proprioceptive gain resulting in greater degrees of freedom and increased flexibility. However, availability of external visual cues limits any changes in proprioceptive gain and the same stimulation

(250 Hz and 17.5 db) does not demonstrate enhanced transfer effects in a split-belt adaptation task (Eikema et al. 2016). It is possible that 17.5 db is a threshold level at or above which vibro-tactile stimulation is perceived as noise requiring alternate sensory gain changes (e.g., proprioception) in the absence of external visual cues. However, below this threshold, there is no increase in proprioceptive gain resulting in a constriction of the degrees of freedom. This was shown by an increase in the strength of LRC in comparison to the baseline walking condition for the 250 Hz and 12 db conditions. We showed that perception of this stimulation (250 Hz and 12 db) was lower than the stimulation at 250 Hz and 17.5 db in all three conditions (sitting, one-legged and two-legged standing; Fig. 2).

The LRC derived through the use of DFA in this study has been considered for some time now as a hallmark of complex self-organizing dynamical systems (Bak and Chen 1991). While LRC (specifically $1/f$ scaling) has been discovered in a wide range of human behavior (Aks et al. 2002; Delignières and Torre 2009; Gilden 2001; Van; Orden et al. 2003), such complex behavior has also been observed in brain activity (Zhigalov et al. 2016; Bédard et al. 2006; Novikov et al. 1997) especially during sensorimotor tasks (Samek et al. 2016) leading to the suggestion that behavioral variability stems from the intrinsic dynamics of neuronal oscillations in the brain (Palva et al. 2013). Given that neuronal oscillations in the brain are strongly related to the modulation of tactile (Michail et al. 2016) and proprioceptive (Nikulin et al. 2008) feedback, it is intuitive to hypothesize that neural control of gait variability, as in our study, was impacted by the tactile perceptual disturbance brought about by continuous foot stimulation.

There were some limitations in this study. First, many other combinations of frequency/amplitude manipulations were possible. However, our intention was to test a range of parameters from low to high amplitude/frequency without fatiguing the subjects. Second, the effect on other gait patterns such as step width variability remains to be investigated. Third, other parameters of the stimuli such as intermittency were not investigated. These will be the focus of future studies. While a case can be made for sensory stimulation on other parts of the body affecting gait dynamics, our intention was to specifically target the tactile stimulus that is felt under the foot during gait. Stimulation at other locations may lead to similar effects but these locations will probably be those that have a role to play in gait control, for example, proprioceptive stimulation on the neck, thighs, or Achilles tendon (Ivanenko et al. 2000a, b), vestibular stimulation on the mastoid process (Chien et al. 2016), or tactile stimulation below the feet (Mukherjee et al. 2015; Eikema et al. 2016). We showed that continuous foot perturbation through vibratory tactors enhanced split-belt locomotor adaptation (Mukherjee et al. 2015). While performance on the split-belt did not change, tactile perturbation led to enhanced overground transfer effects. This effect was attributed to enhanced proprioceptive gain in response to the reduced tactile feedback. In a subsequent study (Eikema et al. 2016), we demonstrated that such effects may be reversed by the presence of simulated optic flow during split-belt learning. It is unlikely that such specific effects which require feedback of alternate treadmill belt velocities could be affected by tactile stimulation at any random part of the body.

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Fig. 1. Instrumentation for the study. **a** One subject walking on the treadmill with the tactor controlling unit carried in a pouch around the waist. The tactor controlling unit sends stimulus signals to six c2 tactors (3 in each insole); **b** the figure shows the three C2 tactors embedded in two sites of each insole—ball of the foot and heel

Fig. 2.

Tactile perception in the tested subjects for the three different postures. The visual analogue scale (VAS) was applied during **a** two-legged standing, **b** one-legged standing, and **c** sitting with feet raised above the floor. The *blue circle* represents the maximum amplitude–frequency combination provided to the subject with the instrument at 17.5 db/250 Hz. The subjects were instructed to score this condition as 100 on the VAS and rank the other conditions accordingly

Fig. 3.

Comparison of the strength of long-range correlation (LRC) of stride interval among 250 Hz/12 db, 250 Hz/17.5 db, and non-tactile stimulus (control) conditions. The strength of LRC was significantly higher at 250 Hz/12 db ($p = 0.009$) but significantly lower at 250 Hz/17.5 db ($p = 0.014$) in comparison with the no stimulus condition (* $p < 0.05$, ** $p < 0.01$, *** $p < 0.001$)

Fig. 4. Long-range correlations of stride interval in one participant for the different tactile conditions: **a** no stimulus, **b** 250 Hz/17.5 db, **c** 250 Hz/12 db

Table 1

The mean gait variability (standard deviation) in meters in the spatial domain (stride length) for the different conditions

Frequency/amplitude	30 Hz	150 Hz	250 Hz
8.0 db	0.042 ± 0.09	0.040 ± 0.01	0.043 ± 0.01
12.0 db	0.044 ± 0.01	0.043 ± 0.01	0.044 ± 0.01
17.5 db	0.043 ± 0.07	0.048 ± 0.01	0.040 ± 0.01

Control: 0.0324 ± 0.01 m

Table 2

The mean fractal scaling index (α) in the spatial domain (stride length) for the different conditions

Frequency Ψ /amplitude	30 Hz [!]	150 Hz [!]	250 Hz
8.0 db	0.68 $\pm$ 0.09	0.72 $\pm$ 0.11	0.89 $\pm$ 0.26
12.0 db	0.76 $\pm$ 0.22	0.78 $\pm$ 0.15	0.79 $\pm$ 0.20
17.5 db	0.73 $\pm$ 0.11	0.65 $\pm$ 0.17	0.79 $\pm$ 0.16

Control = 0.76 $\pm$ 0.16

Ψ Main effect ($p < 0.01$)

[!] Significant difference exhibited when compared to the 250 Hz conditions

Table 3

The mean gait variability (standard deviation) in milliseconds in the temporal domain (stride interval) for the different conditions

Frequency/amplitude	30 Hz	150 Hz	250 Hz
8.0 db	26.7 ± 12.0	28.4 ± 12.8	29.9 ± 14.1
12.0 db	29.4 ± 15.4	29.7 ± 14.7	31.9 ± 17.7
17.5 db	29.5 ± 10.6	32.6 ± 14.5	28.8 ± 10.5

Control: 24.6 ± 9.2 ms

Table 4

The mean fractal scaling index (α) in the temporal domain (stride interval) for the different conditions

Frequency/amplitude Ψ	30 Hz	150 Hz	250 Hz
8.0 db ^{!!}	$0.73 \pm 0.09^{\#}$	$0.75 \pm 0.12^{\#}$	0.85 ± 0.24
12.0 db	0.80 ± 0.15	0.80 ± 0.17	$1.00 \pm 0.08^{*\$}$
17.5 db ^{!!}	$0.77 \pm 0.15^{\#}$	0.77 ± 0.21	$0.65 \pm 0.08^{\#\$}$

Control (0.83 ± 0.07 ; $N = 10$)

Ψ Major effect ($p < 0.01$)

[!] Significant difference exhibited when compared to 12 db

^{*} Significant difference exhibited when compared to 250 Hz/17.5 db

[#] Significant difference exhibited when compared to 250 Hz/12.0 db

^{\$} Significant difference exhibited when compared to control (0.83 ± 0.07 ; $N = 10$)

Single symbols ($p < 0.05$) and double symbols ($p < 0.01$)

Interaction between frequency and amplitude ($p = 0.001$)