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A novel approach to mechanical foot stimulation during human locomotion under body weight support

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ABSTRACT

Input from the foot plays an essential part in perceiving support surfaces and determining kinematic events in human walking. To simulate adequate tactile pressure inputs under body weight support (BWS) conditions that represent an effective form of locomotion training, we here developed a new method of phasic mechanical foot stimulation using light-weight pneumatic insoles placed inside the shoes (under the heel and metatarsus). To test the system, we asked healthy participants to walk on a treadmill with different levels of BWS. The pressure under the stimulated areas of the feet and subjective sensations were higher at high levels of BWS and when applied to the ball and toes rather than heels. Foot stimulation did not disturb significantly the normal motor pattern, and in all participants we evoked a reliable step-synchronized triggering of stimuli for each leg separately. This approach has been performed in a general framework looking for “afferent templates” of human locomotion that could be used for functional sensory stimulation. The proposed technique can be used to imitate or partially restore surrogate contact forces under body weight support conditions.

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1. Introduction

Locomotor therapy with body weight support (BWS) is an effective tool in rehabilitation that may help patients regain the ability to walk (Barbeau, Ladouceur, Norman, Pepin, & Leroux, 1999; Dietz & Colombo, 2004; Edgerton, Kim, Ichiyama, Gerasimenko, & Roy, 2006; Grasso et al., 2004; Schmidt, Werner, Bernhardt, Hesse, & Krüger, 2007; Scivoletto et al., 2007). Foot–support interactions and appropriate sensory signals are an integral part of the rhythm-generating networks (Duysens, Clarac, & Cruse, 2000). A variety of sensory receptors can be activated by limb loading. These may include Golgi tendon organs, spindles, cutaneous receptors, and various load mechanoreceptors in the foot arch. However, the particular contribution by each type of receptor is still a matter of debate. Moreover, the ability of any single input to entrain or affect the locomotor rhythm is much reduced when competing with the input from other afferents and descending pathways (Ivanenko, Grasso, & Lacquaniti, 2000; Stephens & Yang, 1999; Whelan & Pearson, 1997). A current view on the role of different mechanoreceptors considers task- and context-dependent contribution of sensory inputs (Pearson, 2004), as well as its maturation in early development (Dominici, Ivanenko, & Lacquaniti, 2007; Musselman & Yang, 2007). In addition, the organization of the interneuronal network for locomotion and the use of corrective reactions during walking points toward a rule-based finite control system rather than a simple additive principle of multisensory fusion (Misiaszek, 2006; Prochazka, 1996). A number of cutaneous reflexes may also participate in the fine control of foot positioning in animals (Guertin, Angel, Perreault, & McCrea, 1995; Schouenborg & Weng, 1994) and humans (Abbruzzese, Rubino, & Schieppati, 1996; Aniss, Gandevia, & Burke, 1992; Sayenko et al., 2007; van Wezel, Ottenhoff, & Duysens, 1997; Yang & Stein, 1990; Zehr & Stein, 1999). Nevertheless, the functional role of specific groups of sensory receptors in regulating human locomotion is still uncertain because they cannot be easily separated since they interact with each other and with central rhythm-generating centers in a complex manner. In addition, support surface and contact with a ground may also be included as components of our ego space in a similar way as external objects and tools can be included in our body scheme (Berti & Frassinetti, 2000; Iriki, Tanaka, & Iwamura, 1996; Ivanenko, Levik, Talis, & Gurfinkel, 1997; Solopova, Kazennikov, Deniskina, Levik, & Ivanenko, 2003).

While load feedback might reflect mostly sensory input from muscle, tendon organ, and joint receptors, cutaneous inputs from the foot also carry limb loading information. Loss of cutaneous sensation may in fact lead to less stable posture and locomotion (Courtemanche et al., 1996; Dingwell & Cavanagh, 2001; Meyer, Oddsson, & De Luca, 2004; Perry, McIlroy, & Maki, 2000; Taylor, Menz, & Keenan, 2004) or affect the foot motion (Bouyer & Rossignol, 2003). In these processes, even minimal contact forces can be functionally critical (Ivanenko, Grasso, Macellari, & Lacquaniti, 2002; Jeka, Schöner, Dijkstra, Ribeiro, & Lackner, 1997). In addition, mechanical foot stimulation may attenuate muscle atrophy induced by prolonged hindlimb unloading (Kyparos, Feeback, Layne, Martinez, & Clarke, 2005) such as that induced in astronauts by prolonged exposure to microgravity, or may enhance voluntary contractions during exercises in leg muscles (Layne, Forth, Baxter, & Houser, 2005). Thus, foot pressure stimulation can be used as an important gait-related sensory input during gait rehabilitation in combination with other methods or therapies. The tactile pressure input from the soles is attenuated and may vary significantly with body unloading (Flynn, Canavan, Chiang, & Cavanagh, 1997; Ivanenko et al., 2002) or when using gait machines enabling the repetitive foot motion (Dietz & Colombo, 2004; Hesse, Schmidt, & Werner, 2006). Several techniques have been previously proposed to increase or modify an input from the foot (Hijmans, Geertzen, Schokker, & Postema, 2007; Priplata, Niemi, Harry, Lipsitz, & Collins, 2003) such as step-synchronized vibration stimulation of soles (Novak & Novak, 2006; Priplata et al., 2006), ankle–foot orthosis (Gordon, Wu, Kahn, Dhaher, & Schmit, 2009), redistributing plantar pressure footwear (Bus, Waaijman, Arts, & Manning, 2009; Duranti, Galletti, & Pantaleo, 1985; Rao, Baumhauer, Becica, & Nawoczenski, 2009), mechanical support stimulation imitating walking in conditions of microgravity or prolonged hypokinesia (Chernikova, Umarova, Saenko, & Kozlovskaya, 2007; Sayenko, Miller, Ivanov, Galanov, & Guekht, 2005) or electrical stimulation of distal nerves. The latter approach has been typically used to test cutaneous reflexes rather than to imitate or restore the sensory input from the foot. Here we developed and tested a new system for phasic stimulation of foot soles and for overcoming the attenuation of foot sole stimulation during BWS. It consists of lightweight pneumatic insoles (PI) placed inside the shoes in combination with body weight support and real-time motion analysis system. To test the system, we measured the spatio-temporal characteristics of foot pressure and gait parameters during mechanical foot stimulation at different levels of BWS. The advantage of this technique is that it represents a non-invasive method for augmenting the afferent input from foot receptors and for testing whether pressure applied to the sole of the foot during the stance phase plays a role in various locomotion conditions. The relevance of foot stimulation for the theoretical understanding of the control of locomotion is discussed.

2. Methods

2.1. General description of the system

The system we developed (Fig. 1) included the following main components and properties:

- treadmill stepping in combination with body weight unloading;
- pneumatic insoles for each leg to mechanically stimulate the foot soles;
- motion analysis system for recording and 3D reconstruction of limb kinematics;

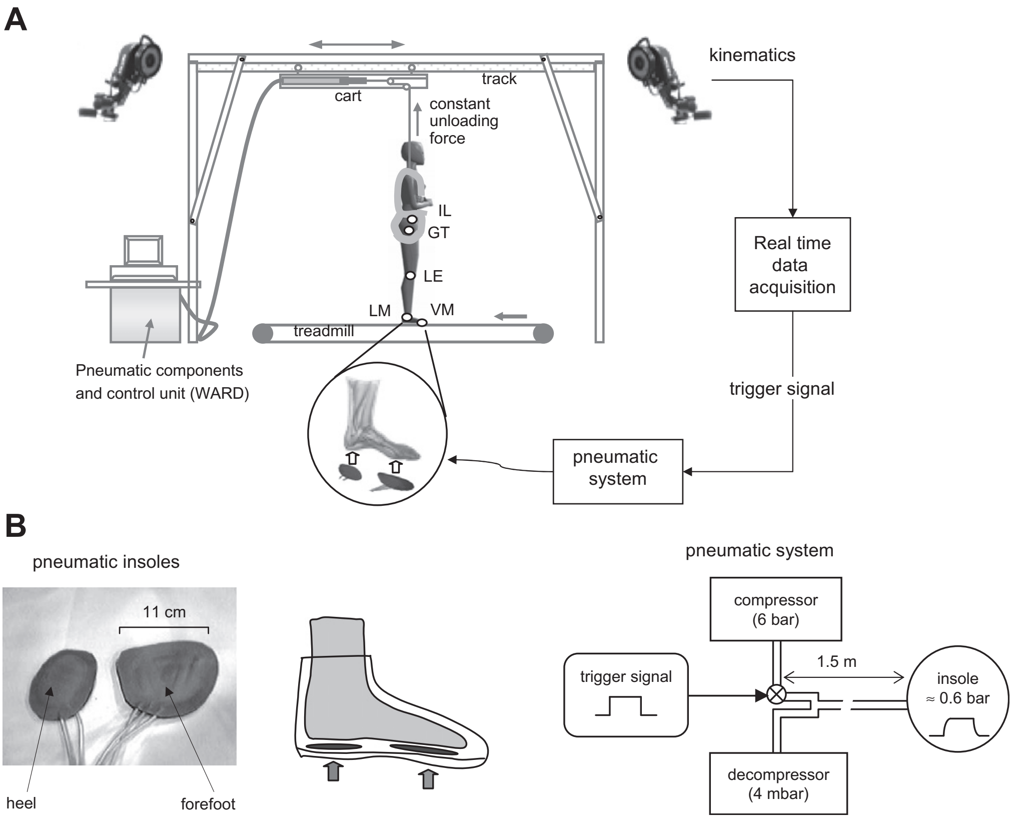

Fig. 1. Experimental setup. (A) Subjects walked on a treadmill with different levels of body weight support (BWS). They were supported in a harness, pulled upwards by a cable connected to a pneumatic device that exerted the preset unloading force. Limb kinematics was recorded by monitoring the coordinates of 5 markers at the following landmarks: the midpoint between the anterior and the posterior superior iliac spine (ilium, IL), greater trochanter (GT), lateral femur epicondyle (LE), lateral malleolus (LM), and fifth metatarso-phalangeal joint (VM). (B) Heel and forefoot pneumatic insoles (left) and a schematic control of the pneumatic system (right). Pneumatic insoles were placed inside the shoes.

– real-time communication protocols providing feedback and an on-line definition of the stance phase of walking for each leg;
– variable treadmill velocity to provide the participant an ability to change his/her walking speed when stepping on a treadmill.

2.2. Experimental setup

The experiments with different levels of body weight support (BWS; 50%, 75%, 90%, and 100% of body weight) were carried out on a treadmill (EN-MILL 3446.527, Bonte Zwolle BV, Netherlands) (Fig. 1A). The walking surface of this treadmill is 1.5 m long, 0.6 m wide, and 0.15 m high above the ground. Treadmill velocity was set either constant (3 km/h) or variable, allowing the participant to modulate his/her walking speed. In the latter case, the instantaneous velocity of the treadmill was recorded (resolution, 0.1 km/h) and controlled by a computer using a real-time kinematic feedback from the participant's horizontal position: forward displacements from the initial position increased the treadmill velocity proportionally, whereas backward displacements decreased it. To monitor a participant's position, we measured the coordinates of the hip marker (a change in position occurred when the velocity of the participant differed from that of the treadmill belt). Position was sampled with an accuracy of 1 mm by means of the VICON-612 motion analysis system (Oxford, UK), and treadmill speed was updated at 20 Hz frequency (after 2 Hz low-pass filtering to avoid jerks). This servo-system provided an efficient and stable control of the treadmill speed. The feedback constant ($G = 3.8$) was selected so as to make participants feel comfortable when they changed their walking speed on the treadmill (Ivanenko et al., 2000).

Pneumatic insoles. Special light-weight pneumatic insoles were designed for mechanical stimulation of the foot. Active components of the PI consisted of two custom-made rubber pneumochambers (one under the heel and another under the ball of the foot and toes) connected to the compressor (AIR-BLOK W250, FIAC, Italy) via nylon tubing (3 mm internal diameter) (Fig. 1B). To make each chamber, the two flat rubber surfaces were glued together over a (8-mm) perimeter so that the central part of the pneumochamber could be inflated via nylon tubing (Fig. 1B). They had a total weight of 0.02 kg (heel chamber) and 0.03 kg (ball chamber), 0.4 cm thickness (when deflated) and dimensions of about 6×9 cm and 9×11 cm, respectively (Fig. 1B). Pneumatic insoles were placed inside both shoes so that the shoes were a critical part of the system since the pressure (0.6 bar) was spread and applied to the different foot regions (Fig. 1B, middle panel). Four parallel pressure regulators (PNEUMAX, Singapore) (to limit the maximal pressure to 0.6 bar) supplied compressed air to four insoles separately (two for the left foot and two for the right foot). The trigger signal opened and closed the solenoid valve connecting and disconnecting the compressor from the insoles. An electromechanical delay for opening and closing the solenoid valve was 50 ms and could be easily taken into account when programming the onset and the end of the actual stimulus. A decompressor (Diaphragm Vacuum Pump MZ 2D, Vacuubrand Gmbh + CO KG, Germany) worked continuously (2.1 m^3/h pumping speed, 4 mbar low ultimate vacuum), helping to exhaust pressure more rapidly after the trigger-off signal closed the valve of the compressor (Fig. 1B). Compressor, decompressor, and airflow produced a low level audible noise which likely had only a minimal effect on the locomotor pattern. Maximum pressure inside the insoles was set at 0.6 bar for the sake of insoles' safeguard and it was comparable with that (0.5 bar) used in static conditions (Chernikova et al., 2007). Higher values may also evoke pain sensations as has been evaluated in preliminary experiments.

BWS. Body weight support was obtained by suspending the participants in a harness connected to a pneumatic device that applied a controlled upward force at the waist, WARD system (Gazzani, Fadda, Torre, & Macellari, 2000). The overall constant error in the applied force and dynamic force fluctuations monitored by a load cell were less than 5% of the body weight (Gazzani et al., 2000).

2.3. Participants

Eight healthy participants (5 males and 3 females, 35 ± 13 (mean ± *SD*) yr of age, 69 ± 11 kg, 1.74 ± 0.08 m) volunteered for the recordings. The studies complied with the Declaration of Helsinki, and informed consent was obtained from all participants according to the procedures of the Santa Lucia Institute.

2.4. Testing the system

To test the system, we measured both the spatio-temporal characteristics of foot pressure and gait parameters during mechanical foot stimulation at different levels of BWS. All measurements were obtained in two experimental sessions for each participant on separate days.

In the first session, we characterized the spatio-temporal characteristics of foot stimulation since it may interfere with the normal foot loading pattern. The rationale was the following. An extra pressure under the soles produced by the pneumatic insoles and the time needed to inflate them can depend on foot loading. For instance, during full limb loading the total pressure due to the body weight is higher than the extra (0.6 bar) PI pressure and would not allow for the insoles to inflate. Therefore, to make an estimate of the temporal and amplitude characteristics of extra foot pressure, we applied foot stimulation during both quiet standing and the stance phase of walking at different levels of BWS. In addition, in these experiments the participants were familiarized with the stimulus and with walking with BWS.

Temporal characteristics of the mean foot pressure at different levels of limb loadings were first tested under static conditions, i.e., during quiet standing. Two forefoot insoles and two heel insoles of both legs were stimulated simultaneously. However, forefoot and heel areas were stimulated independently (10 times, with a 5 s interval) in order to evaluate temporal characteristics separately. The duration of the stimulus was set at 0.5 s for all insoles.

For stepping, we aimed at approximating and imitating a general sequence of foot pressure events. In normal walking, the plantigrade heel-to-toe contact during stance is characterized by a temporal shift of pressure from the heel to the forefoot. To imitate this general picture of temporal pressure distribution, the stimuli were applied separately to the heel and ball zones during the stance phase of each foot using the real-time kinematic feedback from horizontal foot motion. During walking at 3 km/h, the stance duration is about 850–900 ms and cycle duration is about 1.3–1.5 s (Ivanenko et al., 2002). To mimic the loading pattern, the heel insole was activated from 50 to 550 ms (trigger on and off signals, respectively) and the ball insole from 150 to 700 s (see Section 3). Ten strides were recorded without mechanical foot stimulation and 10 strides with stimulation.

After having experienced foot stimulation during walking at all BWS levels, we asked participants for their sensations about the intensity of the pressure stimulus using a 7-point Likert scale (Chang & Troje, 2008; Dawes, 2008), the value 6 corresponding to the maximally experienced sensation across all conditions. We did not ask to discriminate the stimulus for each leg separately since it was symmetric and equal on both sides. However, we asked them to discriminate the intensity for the forefoot and heel areas. Thus, each participant reported 12 values (6 levels of BWS $\times$ 2 foot zones). Although these measurements are subjective, they represent a typical assessment of relative subjective judgments of the stimulus.

In the second session, we tested whether foot stimulation during the stance phase disturbs significantly the EMG patterns and gait kinematics (cycle duration, horizontal foot excursion, and foot path variability) during walking at 3 km/h with different BWS levels. Ten strides were recorded without mechanical foot stimulation and ten strides with stimulation. The presentation of BWS conditions was randomized across the participants.

In addition, in five participants we tested whether foot stimulation could evoke variations in the walking speed under the assumption that foot unloading at the end of stance may accelerate the timing of the stance-to-swing transition and thus increase the walking speed (e.g., in reduced cat preparation, Duysens and Pearson (1980)). To this end we used a feedback-controlled treadmill belt speed (see above). We tested this effect only at 75% BWS level. However, in this protocol we used a slightly different procedure for the control of stimulus duration. While the onset and duration of the heel stimulus was always constant, the forefoot stimulus termination depended on the stance phase duration computed for the preceding step: the trigger-off signal anticipated the stance-to-swing transition by 200 ms (τ_1) independent of the walking speed. In such a way we tried to synchronize the ending of pressure with stance-to-swing transition. The forefoot stimulus was also triggered-off 350 ms (τ_2)

prior to the stance-to-swing transition. Three trials were recorded in each condition (τ_1 and τ_2) and their presentation was randomized across trials and participants. Prior to application of step-synchronized foot stimulation, the participants were asked to adopt an initial walking speed of about 3 km/h. Stimulus was applied for about 10 steps and the mean walking speeds before and during stimulation were compared.

2.5. Data recording

Bilateral kinematics of locomotion was recorded at 100 Hz by means of the VICON-612 motion analysis system. The positions of selected points on the body were recorded by attaching passive infrared reflective markers (diameter 2.5 cm) to the skin overlying the following bony landmarks on both sides of the body (Fig. 1A): gleno-humeral joint (GH), the tubercle of the anterosuperior iliac crest (IL), greater trochanter (GT), lateral femur epicondyle (LE), lateral malleolus (LM), and fifth metatarso-phalangeal joint (VM).

Electromyographic (EMG) activity was recorded bilaterally by means of surface electrodes from the soleus (SOL), lateral gastrocnemius (LG), tibialis anterior (TA), rectus femoris (RF), vastus lateralis (VL), semitendinosus (ST), biceps femoris (BF, long head), and adductor longus (ADD) muscles. EMG-signals were preconditioned at the recording site (active electrodes from DelSys, Boston, USA), transmitted to the remote amplifier (bandwidth was 20–450 Hz), and sampled at 1000 Hz.

A capacitive pressure distribution PEDAR-mobile system (Novel, Germany) was used to quantify changes in the external load on the foot during stepping with BWS. The individual sensor elements were elastic and arranged in a matrix insole (1.9 mm thickness). The local vertical force sensed by each element was recorded at 50 Hz. The PEDAR insole containing 99 capacitive sensors was interposed between the participant's foot and the sole of the shoes. Pressure threshold was set at 1 N/cm^2. All sensors were statically individually calibrated in a simple, computer-aided procedure using homogenous air pressure. Before each trial, the mean level of each sensor was measured while the foot was unloaded (lifted) for 3–5 s and this value was used as a zero level. Both the pneumatic and PEDAR insoles were simultaneously placed in the footwear. The PEDAR insoles were positioned first and the pneumatic insoles were placed on the top of them. In the text we'll use the term "insole" to implicitly refer to the pneumatic insoles, and in the case of foot pressure distribution measurements we will explicitly refer to the "PEDAR insoles".

2.6. Data analysis

The onset of the gait cycle was defined as the timing of the maximum of the horizontal foot (VM marker) motion relative to GT and, accordingly, gait cycle duration was defined as the time interval between two successive maxima of VM_x. In general the difference between the time events measured from horizontal foot motion and kinetics (recorded by the foot pressure PEDAR-mobile system) was less than 3%.

The relative loading of the forefoot and heel regions during mechanical foot stimulation at different BWS levels was estimated using the PEDAR-mobile pressure distribution system. For illustrative purposes, to characterize the general pattern of foot pressure distribution, peak pressure values of the individual sensors during standing or over the stance phase of walking were displayed using the color scale. To quantify an overall loading of the two main supporting areas of the foot, the foot was subdivided into forefoot and heel zones (Fig. 2A), and for each region we computed the mean pressure value of all sensors as a function of time during stance.

Raw EMG data were numerically high-pass filtered (cutoff 30 Hz) to remove motion artefacts, rectified and then low-pass filtered with a zero-lag Butterworth filter (cutoff 15 Hz). Data from several steps (with and without stimulation) were ensemble-averaged after time-interpolation over individual gait cycles to a normalized 200-point time base.

Foot-trajectory spatial variability in the sagittal plane was quantified in terms of normalized tolerance area of VM, computed over the swing phase (Ivanenko et al., 2002). Briefly, VM trajectories (relative to the mean position of GT) were re-sampled in the space domain by means of linear interpolation of the x, y time series (1.5-mm steps) over all gait cycles. All steps under the same walking

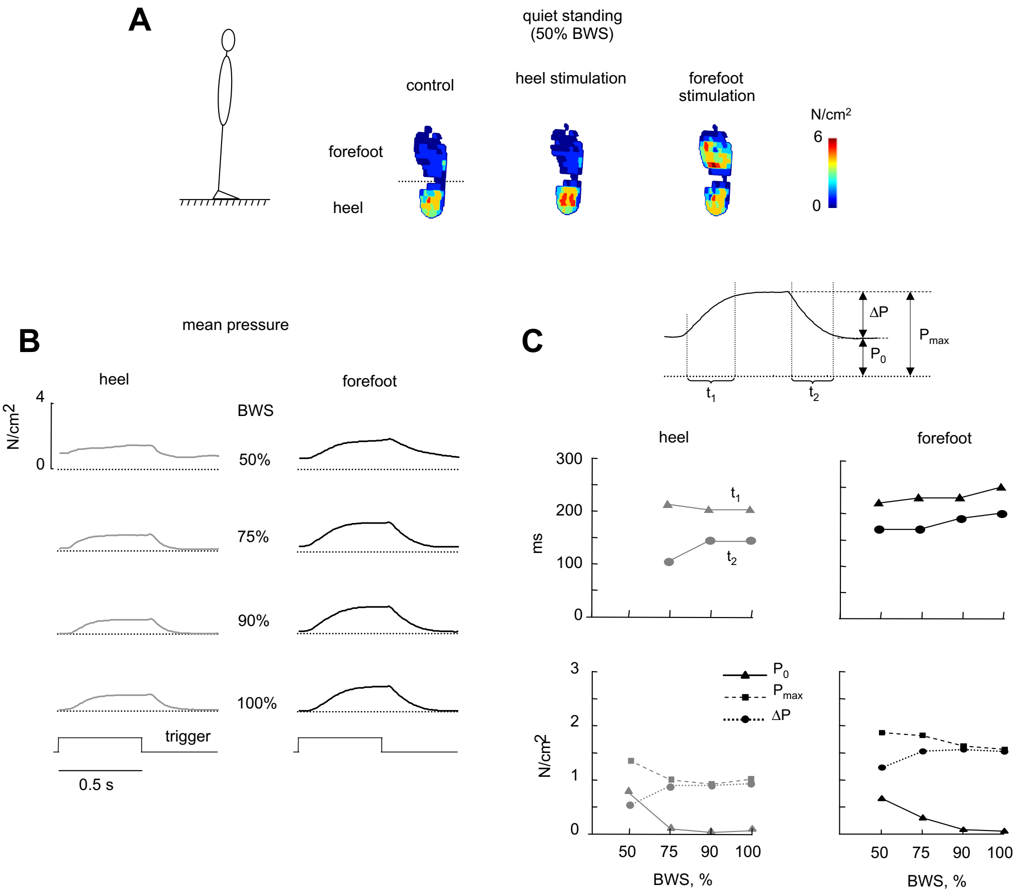

Fig. 2. Amplitude and temporal characteristics of foot stimulation (0.5 s duration) during quiet standing at different levels of BWS. (A) The color pictures illustrate an example of distribution of foot pressure (maximal sensor pressures for the period of 0.5 s) during normal standing and when stimulating heel and forefoot areas in a representative subject. The individual sensor elements were elastic and arranged in a matrix insole (PEDAR-mobile system). The PEDAR insole containing 99 capacitive sensors was interposed between the subject's foot and the sole of the shoes. Both the pneumatic and PEDAR insoles were simultaneously placed in the footwear. (B) An example of changes in the mean pressure as a function of time in a representative subject. The foot was subdivided into forefoot and heel zones, and for each region we computed the mean pressure value of all pressure sensors belonging to this zone. Stimulations of the heel (left) and forefoot (right) areas of both legs were obtained independently (with a 5 s interval). (C) Based on the above measurements, we calculated the mean inflating (t_1) and deflating (t_2) times and pressure characteristics. t_1 and t_2 were computed when the extra pressure exceeded 5% and 95% of the total magnitude of extra pressure ΔP. The time to max ($t_1 \sim 200$ ms) was slightly longer than the relaxation time ($t_2 \sim 150$ ms). Note that the amount of extra pressure (ΔP) increased with body unloading both for the heel and forefoot.

conditions were pooled together for this analysis. For every interval corresponding to 10% of the maximal horizontal excursion, we computed the 2D 95%-tolerance ellipsis of the points within the interval. The areas of all tolerance ellipses were summed and normalized by the mean length of foot trajectory, providing an estimate of the mean area covered by the points along 1 cm of path. This index describes the integrated variability of foot path, including variability in both the vertical and horizontal directions. A greater tolerance area indicates greater variability.

For the protocol with a feedback-controlled treadmill belt speed (when the stimulus was triggered-off at either τ_1 or τ_2, see above), we computed in each trial the mean walking speed before ($n = 10$ strides) and during ($n = 10$ strides) stimulation and then averaged changes in the walking speed across three trials recorded for each condition (τ_1 and τ_2).

2.7. Statistics

Descriptive statistics included means ± standard deviation (*SD*) of the mean. Student's paired *t*-tests were used to compare the foot pressure characteristics and gait parameters across conditions (with and without stimulation): mean plantar pressure, horizontal foot excursion, foot (VM marker) path variability, mean EMG activities, cycle duration, and walking speed (when using a feedback-controlled treadmill belt speed). Reported results were considered significant for $p < .05$.

3. Results

3.1. Characteristics of the stimulus at different levels of foot loading during quiet standing

To characterize the amount and temporal characteristics of the physical action of pneumatic insoles, we applied bilateral stimulation of the forefoot and heel areas during quiet standing at different levels of body weight support. Fig. 2A shows an example of pressure distribution during normal standing and when stimulating heel and forefoot insoles. The stimulus selectively increased pressure under heel or forefoot areas and had only limited mechanical consequences on the pressure under the non-stimulated part of the foot, as can be appreciated from an example in Fig. 2A. For instance, when stimulating the heel insoles during standing with 50% BWS, the mean pressure under the heel increased by 0.47 ± 0.15 N/cm^2 while it remained unchanged under the forefoot (0.01 ± 0.04 N/cm^2). When stimulating the forefoot insoles, the mean pressure under the forefoot increased by 1.29 ± 0.21 N/cm^2 while it even slightly decreased under the heel (-0.09 ± 0.07 N/cm^2). Thus, the crosstalk was minimal and the pressure was simply transmitted through the confines of the shoe.

We measured the amount of the sole pressure by means of the PEDAR-mobile system. To characterize the stimulus, the foot was subdivided into a heel and forefoot (ball and toes) zones, and for each region we computed the mean pressure value of all pressure sensors belonging to this zone. An example of changes in the mean pressure as a function of time at different BWS levels is shown in Fig. 2B. Based on these measurements, we calculated the mean inflating (t_1) and deflating (t_2) times. Both t_1 and t_2 increased with increasing BWS during forefoot stimulation and were relatively constant for heel stimulation (Fig. 2C). However, heel stimulation timings could be reliably estimated only at high levels of body unloading since at 0% BWS the amount of extra pressure was minute; t_1 ($\sim$200 ms) was slightly longer than the relaxation time t_2 ($\sim$150 ms) for all insoles likely due to the functioning of the decompressor, which counteracted inflation but accelerated deflation. A decompressor worked continuously (at 4 mbar), helping to exhaust pressure more rapidly after the trigger-off signal closed the valve of the compressor (Fig. 1B): in its absence t_2 was much longer than 150 ms (about 300–400 ms depending on the BWS level).

The amount of extra pressure (ΔP) increased with BWS both for the heel and forefoot insoles (Fig. 2C). The maximal mean pressure was higher for the forefoot stimulus than for the heel stimulus ($p < .01$ for all BWS conditions tested) likely because the shoes are an important part of the system to counteract and distribute pressure (Fig. 1B) and fixation of the insoles is tighter under the metatarsus than under the heel, due to some vertical freedom of the heel in the shoe.

3.2. Characteristics of the stimulus during walking at different BWS levels

We also estimated foot pressure changes during walking at different BWS levels. During normal walking, the participant makes a typical heel–toe rolling during stance (Fig. 3A). To imitate this general foot contact pattern, we stimulated heel and forefoot regions accordingly (see trigger signals in the bottom of Fig. 3A). Fig. 3B shows an example of the plantar pressure pattern (maximal sensor pressures) in one representative participant and Fig. 3C shows corresponding changes in the mean pressure under the heel (left) and forefoot (right) zones as a function of time. An extra pressure ΔP produced by mechanical foot stimulation was more prominent at high BWS levels in agreement with postural recordings (Fig. 2). To estimate it we subtracted the time courses of the mean pressure obtained during steps with and without stimulation (gray areas in Fig. 3C).

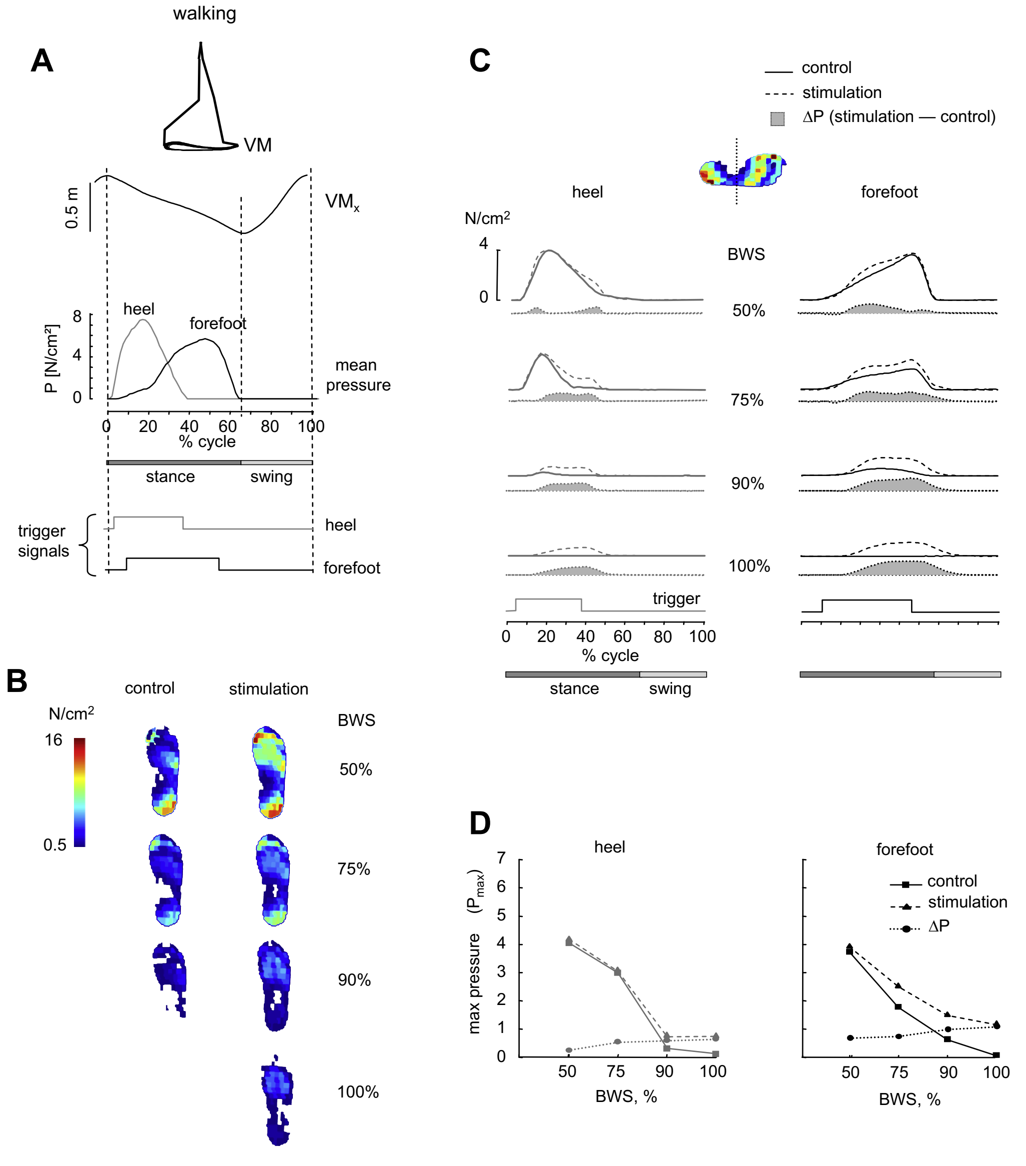

Fig. 3. Amplitude and temporal characteristics of foot stimulation during walking at 3 km/h with different levels of BWS. (A) A typical heel–toe rolling during stance in one representative subject walking at 0% BWS (without activation of pneumatic insoles). To increase or imitate this general foot contact pattern, we stimulated heel and forefoot regions accordingly (see trigger signals in the bottom) using the real-time kinematic feedback from horizontal foot (VM_x) motion. The onset of the gait cycle was defined as the timing of the maximum in VM_x. The heel insole was activated from 50 to 550 ms and the ball insole from 150 to 700 s. The heel strike event was defined as the time of the maximum of VM_x. (B) Examples of the general plantar pressure pattern during stance in one representative subject. Peak pressure values during stance are shown using the color scale. (C) An example of changes in the mean pressure under the heel (left) and forefoot (right) zones as a function of time. (D) Maximal pressure values (averaged across all subjects) of the heel and forefoot zones during stepping with and without stimulation and the amplitude of the extra pressure ΔP. Note an increment of ΔP with body unloading.

Again, ΔP was higher under the metatarsus than under the heel and increased monotonically with body unloading (Fig. 3D), as was the case for the quiet standing trials (Fig. 2). Note, however, that total foot loading during the stance phase in walking was more than twofold higher than during standing since the stance limb bore the full weight of the body while in quiet standing it bears only half the body weight.

3.3. Subjective reports

Since the tactile information from the foot may evoke contact- or load-related sensations (Roll, Kavounoudias, & Roll, 2002), we also asked participants for their perception of the stimulus. All participants reported prominent sensations of an extra foot pressure during walking. While the insole raised the pressure inside the entire shoe, "pressure on the sole" prevailed in the subjective reports. Table 1 contains the mean subjective estimates of the intensity of the stimulus, expressed according to a 7-point Likert scale (6 = max experienced sensation across all conditions). In general, subjective sensations corresponded roughly to the amount of extra pressure. For instance, they increased monotonically with increasing body unloading in accordance with monotonic increments in ΔP (Fig. 3D) and the participants perceived the stimulus under the forefoot as being much stronger than that under the heel.

3.4. Motor patterns and gait parameters during walking at different BWS levels

We tested the system in a step-synchronized manner and validated its functioning during stepping on a treadmill. Real-time communication protocols of the VICON motion analysis system provide a consistent feedback of the bilateral limb kinematics and foot motion. Based on these procedures, we implemented our software that triggered the stimulus during the stance phase to mimic a general pattern of contact with a support. Measuring on-line the heel strike event (the onset of the stance phase) and taking into account an electromechanical delay (50 ms) for opening and closing the solenoid valve connecting and disconnecting the compressor from the insoles, it was possible to apply the stimulus properly for each insole to imitate the heel-to-forefoot rolling pattern during stance (Fig. 3A). In all eight participants we evoked a reliable step-synchronized triggering of stimuli for each leg separately with a precision of 10 ms (corresponding to the sampling rate of the kinematic recordings).

Until now we have described the general characteristics of stimulation in postural and walking conditions. We also tested whether it could significantly disturb the normal gait pattern by recording the kinematics patterns during walking at a constant speed. Mechanical foot stimulation did not significantly affect the foot trajectory characteristics (horizontal foot excursion and path variability, Fig. 4A) or EMG patterns (Fig. 4B), nor did it affect the duration of the gait cycle ($p > .5$ in all cases), which was 1.37 ± 0.07 s, 1.45 ± 0.08 s, 1.60 ± 0.16 s, and 1.46 ± 0.24 s at 50%, 75%, 90%, and 100% BWS, respectively. In air-stepping, an extra foot pressure could evoke some limited changes in the shape of the foot trajectory (possibly, in part due to the small direct mechanical effect of pressure on the shoe volume), however, it did not increase (or decrease) foot path variability as one would have expected from application of surrogate contact forces (Ivanenko et al., 2002).

Walking on a treadmill forces the participant to keep a constant velocity so that foot unloading could not evoke significant changes in the steady state of walking, except for some potential minor

Table 1
Subjective estimates (mean values and range) of the intensity of the pressure stimulus during walking at 3 km/h at different levels of body weight support. Estimates were expressed according to a 7-point Likert scale (6 = max experienced sensation across all conditions).

BWS level (%)	Heel	Forefoot
50	1 (0–3)	2 (1–5)
75	2 (1–4)	4 (3–6)
90	2 (1–4)	6 (4–6)
100	3 (2–5)	6 (6–6)

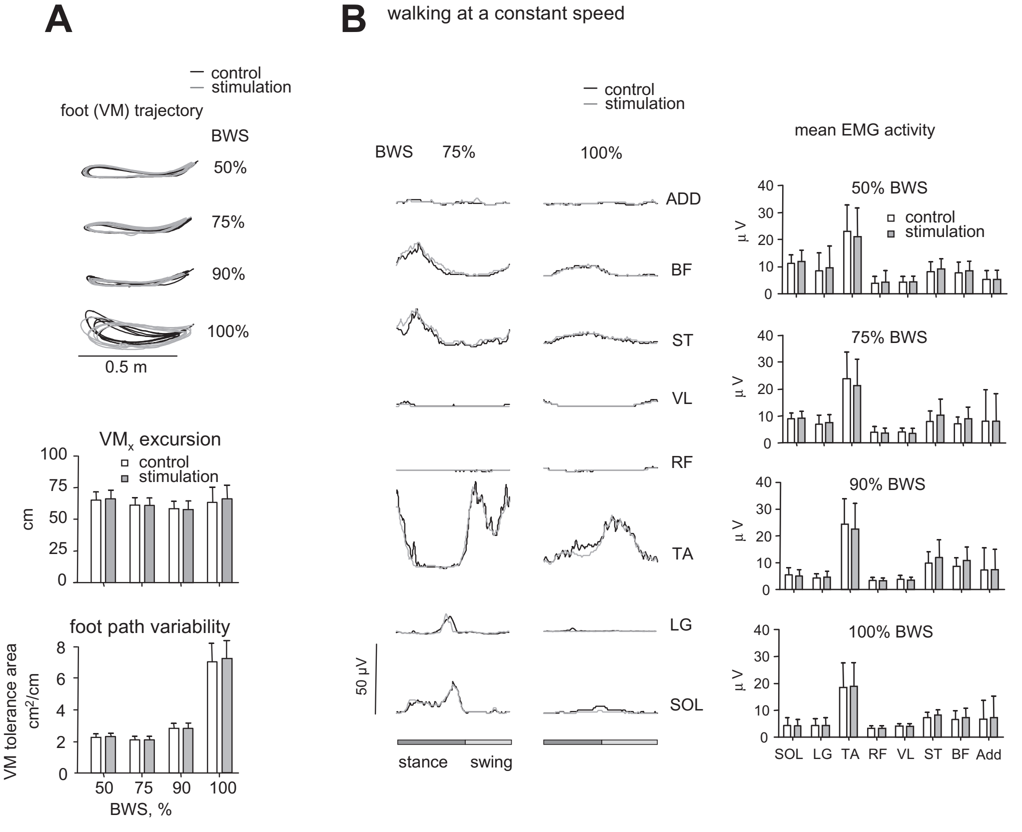

Fig. 4. Effect of foot pressure on the general parameters of foot motion and EMG patterns during walking on a treadmill at 3 km/h. (A) Foot trajectories (10 steps) in a representative subject and horizontal foot excursion and variability (mean ± *SD*, $n = 8$ subjects). (B) An example of ensemble-averaged EMG patterns in one subject (left) and mean EMG activities ($n = 8$ subjects). Note that foot stimulation did not significantly disturb the foot trajectory characteristics and the motor patterns.

changes in the step length or duration. To overcome this limitation, we also developed a feedback-controlled treadmill speed protocol. In this protocol, velocity (v) was proportional to the horizontal position (x) of the participant: $v = 3.8 \cdot x$ (Fig. 5A, see Section 2). Forward displacements increased the treadmill velocity, whereas backward displacements decreased it. The timing of the stimulus under the heel was constant (from 50 to 550 ms), while the duration of that of the forefoot was scaled according to the duration of the stance phase of the preceding step. The forefoot stimulus was triggered-off either 200 ms (τ_1) or 350 ms (τ_2) prior to the stance-to-swing transition. The former was scheduled to slightly anticipate the end of the stance phase (taking into account a delay and duration of insole deflating, Fig. 5A) while the latter was expected to abolish pressure much before the toe off event and served as a control. A schematic extra pressure profile (approximated from Fig. 3C) is shown in Fig. 5A. The initial walking speed was about 3 km/h. Mean velocity prior to and during mechanical stimulation was analyzed. When treadmill speed was set variable, allowing the participant to modulate his/her walking speed, we could observe changes in the walking speed that depended on timing of stimulation (Fig. 5B). There were small but significant increments in speed in the τ_1 condition during stepping at 75% ($p < .05$, paired t-test). There was no significant difference in the walking speed increment between the first and last trials ($p = .3$). When the pressure stimulus ended long before the end of the stance phase (τ_2), no detectable changes were found ($p = .45$).

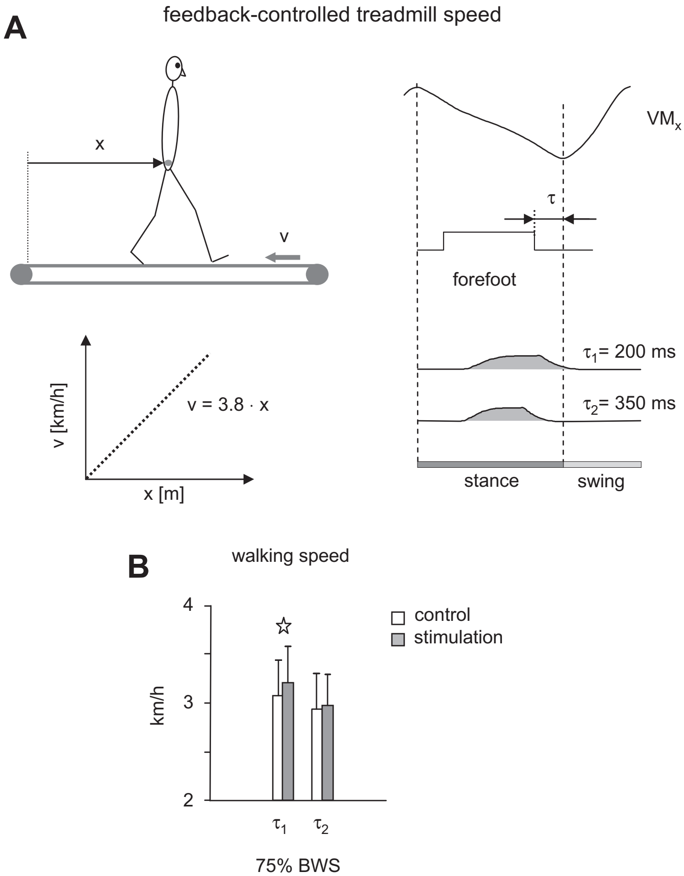

Fig. 5. Effect of foot stimulation on the walking speed. (A) Treadmill speed was set variable allowing the subject to modulate his/her walking speed. Velocity (v) was proportional to the horizontal position (x) of the subject: $v = 3.8 \cdot x$ (see Section 2). Forward displacements increased the treadmill velocity, whereas backward displacements decreased it. The forefoot stimulus was triggered-off either 200 ms (τ_1) or 350 ms (τ_2) prior to the stance-to-swing transition. A schematic extra pressure profile (approximated from Fig. 3C) is shown on the bottom (in gray). (B) Mean velocity prior to and during mechanical stimulation (mean ± *SD*, $n = 5$ subjects). Note small but significant increments in speed in the τ_1 condition during stepping at 75% (denoted by asterisk).

4. Discussion

This study describes a novel system for applying mechanical foot stimulation under BWS. The results represent the first steps toward employing this stimulation and assessing the role of foot pressure for various applications. We have used this system successfully to provide a step-synchronized extra foot pressure during the stance phase of locomotion. Considering natural foot pressure changes during normal gait, 150–200 ms inflation and deflation times (Fig. 2C) represent a reasonable approximation of the rate of total pressure changes in walking. The proposed foot stimulation technique can be used only to partially restore contact forces and provide a temporal sequence of forefoot and heel stimulation under body weight support conditions (for instance, appropriate gait-related directional shear forces are not generated by the inflating insoles). Although in this study we used a fixed upper limit for the stimulus (0.6 bar), the amount of pressure can be adjusted for each insole via pressure regulators as a function of the characteristics of the participant (e.g., in children it could be set proportional to body weight or foot size) or to provide a different stimulation of the forefoot and heel zones or an asymmetric stimulation for both legs (Chaudhuri & Aruin, 2000). Both cutaneous receptors and the proprioceptors of the small muscles and joints of the plantar arch are stimulated by this technique. The shoes are an important element of the system. Fixation of the insoles is tighter under the metatarsus than under the heel, due to some vertical freedom of the heel in the shoe. In fact, the participants perceived the stimulus under the forefoot as being stronger than that under the heel, in part also due to lower thresholds in the ball and arch of the sole than in the heel (Inglis, Kennedy, Wells, & Chua, 2002). Even being selective for the heel and ball zones of the foot (Fig. 2A), stimulation acts not only on the sole but the insole raises the pressure inside the shoes. Nevertheless, "pressure on the sole" prevails in the subjective reports possibly because the sole is designated to interact with a support and it is densely populated by various receptors.

We observed only limited or no changes in the motor patterns of healthy participants during step-synchronized foot pressure application (Fig. 4). One possibility is that foot pressure perturbation does not result in any change in ankle torque. Thus, although load feedback is known to play an important role in the control of locomotion, the findings reported here suggest that load feedback may not depend on pressure receptors in the foot. Nevertheless, the pilot results are not trivial and may be of interest since they shed some light on the effect of foot pressure and central processing. For instance, from various cutaneous reflexes (Aniss et al., 1992; van Wezel et al., 1997; Yang & Stein, 1990; Zehr & Stein, 1999) one might expect to see a direct effect of foot pressure on motor patterns. In addition, it is well known that loss of cutaneous sensation from the foot can compromise stability in locomotion (Courtemanche et al., 1996; Dingwell & Cavanagh, 2001; Perry et al., 2000). So why do foot pressure perturbations have no discernible effect on gait kinematics or muscle activation patterns?

Another possibility is that the load perturbations were insufficient and not large enough to alter gait kinematics. This explanation finds some support in the simple BWS effect on gait kinematics. For unloadings up to 95% of BW, the gait kinematics were found to be quite normal (Ivanenko et al., 2002). Only when BWS reached 100% was there a significant difference in the gait kinematics. Thus if only a major loss of loading is required to alter limb kinematics, increases in loading might be expected to have a minimal effect in healthy participants. However, if kinematics are insensitive to increased loading, that is likely to be a consequence of compensatory changes in muscle activity.

A third possibility is that load sensing involves some combination of pressure sense in the foot together with muscle and joint loading in the limb. This explanation finds some support in animal studies. For example normal cats are able to adapt to a lack of load-related cutaneous information, while spinal cats cannot (Bouyer & Rossignol, 2003). This suggests that loss of information in one channel (foot pressure for example) may not produce a deficit while other channels (muscle or joint receptors for example) are operative. In the case of peripheral neuropathies, the losses are less selective and likely to affect more than one channel. While this might reflect mostly sensory input from muscle, tendon organ, and joint receptors, cutaneous inputs from the foot also carry limb loading information. Loss of cutaneous sensation from the foot may in fact lead to less stable locomotion (e.g., in peripheral neuropathy, or cold feet).

When variations in speed were allowed, foot stimulation could evoke small increments in the walking speed (Fig. 5). There are controversial data in the literature concerning the effect of foot loading on the cycle duration and walking speed. For instance, Stephens and Yang (1999) found that loading during the stance phase of walking in adults increases the extensor EMG amplitude but does not change the duration of the step cycle (see also Misiaszek, Stephens, Yang, and Pearson (2000)), while a large change in timing was seen in the infants (Musselman & Yang, 2007). Furthermore, the changes in duration of phases may be more prominent in reduced preparations than in intact animals (Duysens & Stein, 1978; Stephens & Yang, 1999; Whelan & Pearson, 1997). Therefore, foot loading influences might be subject- and context-dependent. Further studies are necessary to clarify whether foot pressure changes may affect rhythmogenesis under BWS conditions or in patients.

In air-stepping, extra foot pressure did not decrease foot path variability in the absence of contact events (Fig. 4A) as one would have expected from application of surrogate contact forces (Ivanenko et al., 2002). In that previous study we performed a series of experiments in which participants were unloaded at 100% BWS but sensed minimal contact forces during "stance." This was obtained in two different ways. In one protocol, the lower surface of a piece of compliant foam-rubber taped to the participants' feet lightly touched the moving belt of the treadmill during stance. In another protocol, 100%

unloaded participants stepped on a compliant (pillow-like) foam placed over the stationary treadmill belt. Step-by-step variability decreased substantially in the presence of these surrogate contact forces as compared with air-stepping. This raises the question then as to why mechanical foot stimulation during "stance" has no effect on foot path variability in air-stepping (Fig. 4A). One possibility is that foot pressure in each step of air-stepping is applied independently of the vertical foot displacements during "stance" and thus cannot provide reliable sensory cues about the position of a contact surface. In contrast, light touch contact may provide this feedback and enhance control of posture (Jeka et al., 1997) or foot trajectory in walking (Ivanenko et al., 2002).

5. Concluding remarks

The proposed approach for mechanical foot stimulation during gait and recordings should be interpreted in the general context of the role of foot contact in human locomotion. It may be a critical combination of afferent signals that is needed to generate an appropriate locomotor pattern (Dietz & Harkema, 2004; Misiaszek, 2006). In fact, this approach has been performed in a general framework looking for sensory synergies or "afferent templates" of human locomotion (Ellaway, Taylor, Durbaba, & Rawlinson, 2002) that could be used for functional sensory stimulation. The design criteria for the system are based on both neurophysiological findings about the importance of an adequate temporal foot loading pattern and a simple and flexible usage of activators (light-weight insoles, pressure regulators, tuning of activation timings), together with the maximum freedom of movement for the patient. The tools we developed can be applied to other dynamic conditions (including microgravity, see for instance Khusnutdinova, Netreba, and Kozlovskaya (2004), Layne, Forth, and Abercromby (2005)) during walking or standing or even when restrained on the bed (Selionov, Ivanenko, Solopova, & Gurfinkel, 2009; Shapkova & Schomburg, 2001). The tactile information from the main supporting areas of the foot is also used by the brain for perceptual purposes and can evoke strong kinesthetic illusions (Roll et al., 2002) and activations of somatosensory areas and superior parietal lobes (Chernikova et al., 2007). A direct (touch) contact of bipeds with an external world occurs constantly via the foot so that it represents an important element of our interaction with surroundings. The main purpose of this paper is to introduce the system and provide detailed information about its functioning, and further investigations are needed to clarify how sensory inputs interact and to determine the extent to which different patient populations can benefit from this technology.

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