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Effects of Sensory Augmentation on Postural Control and Gait Symmetry of Transfemoral Amputees - A Case Description

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Abstract Despite recent advances in leg prosthetics, transfemoral amputees still experience limitations in postural control and gait symmetry. It has been hypothesized that artificial sensory information might improve the integration of the prosthesis into the human sensory-motor control loops and, thus, reduce these limitations. In three transfemoral amputees, we investigated the effect of Electrotactile Moving Sensation for Sensory Augmentation (EMSSA) without training and present preliminary findings. Experimental conditions included standing with open/closed eyes on stable/unstable ground as well as treadmill walking. For standing conditions, spatio-temporal posturographic measures and Sample Entropy were derived from the Center of Pressure. For walking conditions, step length and stance duration were calculated. Conditions without feedback showed effects congruent with findings in the literature, e.g. asymmetric weight bearing and step length, and validated the collected data. During standing, with EMSSA a tendency to influence postural control in a *negative* way was found: postural control was less effective and less efficient and the prosthetic leg was less involved. Sample Entropy tended to decrease, suggesting that EMSSA demanded increased attention. During walking, with EMSSA no persistent positive effect was found. This contrasts the positive subjective assessment and the positive effect on one subject's step length.

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1 Introduction

Modern transfemoral prostheses already offer far-reaching restoration of motor function after amputation [2, 31, 45, 47]. However, amputees still show imbalanced use of prosthetic and sound limb [10], which may lead to osteoarthritis in the knee and the hip joint of the unaffected side, osteoporosis in the residual limb, as well as back pain [15]. Furthermore, postural control after lower-limb amputation is impaired [16, 33]: one in two above-knee amputees falls in the first year after amputation and even two in three experienced prosthesis users fall at least once within twelve months [29]. Besides actuation constraints of current prostheses, decreased muscle volume and force as well as limited load-bearing of the pain-sensitive stump [25, 34], one reason for these limitations could be that amputees receive a reduced amount of sensory information compared to able-bodied persons: In addition to vestibular and visual feedback, physiological postural control mechanisms involve cutaneous feedback from the foot sole, as well as proprioceptive input from muscles and joints. Amputees only receive sensory input from their residual limb and rely more on visual input [13]. The reduced prosthetic capability may lead to a reduced engagement in physical and social activities [33] but it can possibly be compensated for: The concept of sensory augmentation hypothesizes that additional information about the prosthesis enhances its integration into the human sensory-motor control loops [27, 40, 50] which eventually promotes the confidence in and the usage of the artificial leg.

For the upper extremities, it has already been shown that artificial proprioceptive feedback benefits users [3, 38]. For the lower extremities, potential beneficial effects of feedback systems on postural control and gait symmetry have been investigated in several studies. Prostheses were equipped with sensors to measure the pressure distribution underneath the prosthetic foot [12, 42], the location of the Center of Pressure (CoP) [52], the knee angle [28], or gait events like heel strike [51], while the user received auditory [5, 18, 51], visual [52], haptic [12, 52], or electrotactile [5, 28, 42, 49] stimuli conveying the information. Some of these solutions have already been tested in unimpaired or amputee subjects, while other concepts are still under development:

Fan et al. [11, 12] developed tactile balloon actuators to be worn around the residual limb which convey information on force loads underneath the prosthetic foot. Perceptual testing was performed in one amputee and six able-bodied subjects and showed that the concept is feasible. Webb et al. [49] used an eight-channel electrical stimulator placed at the residual limb to provide feedback about deviations from the desired hip joint angle. Perceptual testing was successfully performed in 13 able-bodied subjects.

Various studies have tested feedback systems in amputees and addressed the subjective judgment of the subjects. In a case series with four below-knee amputees [52], deviations of CoP location from a gait data-based predefined reference area were communicated via mechanically induced tactile stimuli. Despite the overall positive impression of the subjects, results of a planned follow-up study have not yet been published. In an experiment with three subjects, transfemoral amputees tested auditory feedback on the prosthesis' knee angle [18]. Two subjects appreciated additional information; the third person terminated the study because the feedback system required too much attention. In a study with eight lower-limb amputees, Kawamura et al. [28] reported that seven out of eight amputee subjects considered electrotactile feedback about the pressure distribution underneath the prosthetic foot as useful or partially useful. Using the same stimulation principle, one amputee subject testing electrotactile knee angle feedback could not distinguish between different stimulation locations during walking. Clippinger et al. [6] fitted 13 lower-limb amputees with a sciatic nerve stimulation implant that provided information about heel strike and bending moments in the shank pylon. Subjects reported that ambulation in the dark and stair climbing improved.

Some studies assessed the effect of artificial feedback on amputee postural control or gait symmetry in a quantitative way. Twelve transtibial and twelve transfemoral amputees tested the Sense-of-Feel feedback system [42] where two transcutaneous electrode pairs were placed on the residual limb and elicited stimuli proportional to the pressure at the heel and the metatarsal head, respectively. Statistical evaluation demonstrated positive effects on weight bearing and gait symmetry after a five-hour familiarization phase. In Yang et al. [51], two out of three transtibial amputees improved their gait after six 30 -minute training sessions, using discrete auditory feedback: An acoustic signal sounded when stance time symmetry ratio exceeded preset tolerances [1]. Percentage differences between pre- and post-test were found for symmetry as well as for postural sway.

Available feedback solutions still have several limitations that hinder their application in patients' daily lives: Visual and auditory feedback devices use important sensory channels and may be perceived as cumbersome and confusing [5]. In contrast, electrotactile as well as vibrotactile feedback systems may provide information without overloading common sensory channels. However, using conventional stimulation techniques, the low-to-medium bandwidth of the somatosensory system limits information content [21, 26]. Recently, a novel feedback principle has been introduced: Electrotactile Moving Sensation for Sensory Augmentation (EMSSA) [43]. This solution exploits the tactile phi phenomenon, which creates the illusion of moving point-like sensations between two pairs of electrodes [24, 26]. Allowing continuous spatial feedback instead of modulating stimulation intensity at discrete locations may maximize transmission efficiency and reduce sensory adaptation [21, 26]. It is now possible to convey continuous information about the CoP trajectory or the knee angle during walking.

However, the importance of CoP information for unimpaired postural control during unperturbed stance is still under debate [32]. A theoretical analysis that we recently conducted [35] indicated that available information from receptors in the residual limb, in combination with a dynamical model of the prosthetic leg, allows transfemoral amputees to estimate their CoP location during stance as well as their knee angle during swing with almost the same precision as able-bodied subjects.

Here, we investigate in three transfemoral amputees whether predictions from the theoretical analysis are supported during unperturbed standing and treadmill walking. We also test the effect of feedback in more difficult tasks, i.e. during perturbed standing.

During quiet stance, the effect of EMSSA is assessed by the body weight index (BWI) and three spatio-temporal posturographic measures: Root mean square resultant distance (Δ_{RMS}), mean CoP velocity (v_{CoP}) and Sample Entropy (SEn) calculated from the CoP. Δ_{RMS} and v_{CoP} describe the effectiveness of the postural control system and the effort to maintain the corresponding level of postural stability [37], respectively. SEn describes the regularity of the CoP time series and can help to differentiate between automatic and attentive postural control. This distinction is valuable to determine whether processing artificial information requires additional attention and slows down the volitional response [4]. During walking, stance duration and step length are evaluated.

In the following, the subject sample, the experimental setups, and data analysis methods are described. The obtained results are compared to findings from the theoretical analysis and in the literature, and possible explanations for the observed effects are given.

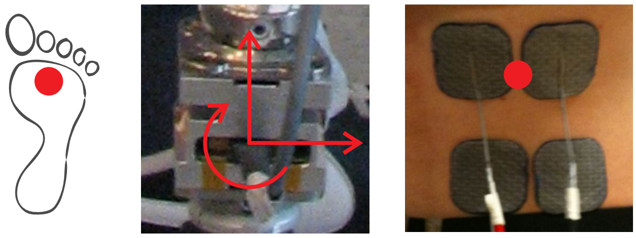

Fig. 1 The CoP underneath the prosthetic foot (left) was calculated from measurements of a modular three degrees-of-freedom force sensor (center). The subject received real-time information via electrotactile stimulation on the lower back (right).

2 Methods

2.1 Participants

Inclusion criteria were a minimum age of 18 years, unilateral transfemoral amputation, constant limb volume and absence of restrictive contractures as well as polyneuropathies. Participants with affected mobility, vision, or balance were excluded. Three male unilateral transfemoral amputees (age: 21, 54, and 73 years; height: 180, 182, and 188 cm; weight: 63, 73, and 75 kg; side of amputation: all left; cause of amputation: two trauma, one bone cancer; time since amputation: 1, 52, and 53 years; regular use of prosthesis: 1, 36, and 53 years) participated voluntarily and gave informed consent. All procedures complied with the Declaration of Helsinki and were approved by the Cantonal Ethics Committee Zurich.

2.2 Experimental Setup

Subjects wore their own prosthesis (all passive with hydraulic damping: two 3R80, one 3R60, both Ottobock, Duderstadt, Germany) with the addition of a modular, lightweight force/moment sensor [46] mounted between prosthetic foot and shank tube (Fig. 1, center). The sensor measured two force components and the bending moment in the sagittal plane. The analog signals were preamplified, low-pass filtered and galvanically decoupled before it was transferred via an A/D converter (NI PCI-6071E, National Instruments) to a computer executing real-time code with xPC Target (The Mathworks Inc., Natick, Massachusetts). From the sensor data, the CoP (Fig. 1, left) was calculated at a sampling rate of 1000 Hz. A goniometer-gyroscope sensor system [14] provided knee and hip joint angles and velocities in real-time at a sampling rate of 1000 Hz.

Subjects stood and walked on an instrumented single-belt treadmill (h/p cosmos, Nussdorf-Traunstein, Germany). Eight force sensors, embedded into two individual plates at the front and the back of the treadmill (Fig. 2, dashed line), recorded ground reaction forces (GRF) at 1000 Hz.

For EMSSA, two pairs of round electrodes (3 cm, PALS Neurostimulation electrodes, Axelgaard Manufacturing CO., Ltd.) were placed on the lumbar area of the subjects' back (Fig. 1, right). For each pair, a stimulator generated charge-balanced biphasic signals with a carrier frequency of 6 kHz [20, 21, 43]. Changing the relative intensities of these two channels elicited the sensation of a traveling stimulus. This effect is known as tactile phi phenomenon [24]. In a study with 37 participants, the recognition rate of the sensation's moving direction and location was about 75% [21]. A subsequent case study has shown that the tactile phi phenomenon can be exploited to display different location sensations that correspond to the CoP during standing and walking [36] with a recognition accuracy of 73% and 83%, respectively.

Two types of feedback were tested: CoP feedback and knee angle feedback. For CoP feedback, the anterior-posterior CoP location underneath the foot was mapped to a continuously moving point-like stimulus on the back, such that the CoP at the heel was associated with a stimulus between the lower pair of electrodes. Shifting the body weight to the toes, the sensation moved upwards along the back. CoP feedback could be provided during quiet stance and during stance phase of walking. For angle feedback, a flexed leg was associated with a stimulus between the lower pair of electrodes. Extending the leg, the sensation moved upwards along the back. Angle feedback was only provided during swing phase of walking, and was switched off as soon as the prosthesis was loaded with 10 % of total body weight. During walking, only one of the two feedback signals was provided.

To ensure the participants' safety, electrical and software range limitations as well as manual emergency stop switches were included. Furthermore, subjects wore a safety harness that would have caught them in the event of a fall. A physiotherapist was present and ready to give support if necessary.

Task	EO	EC	HARD	SOFT	NOCOUNT	COUNT
1		✓	✓		✓	
2		✓	✓			✓
3	✓			✓	✓	
4	✓			✓		✓
5		✓		✓	✓	

Table 1 During quiet stance, six conditions, i.e. eyes open (EO) vs. eyes closed (EC), hard (HARD) vs. soft (SOFT) surface, no counting (NOCOUNT) vs. counting (COUNT), were combined and resulted in five different tasks (T). Tasks are sorted by increasing difficulty. Each task was repeated with FB ON and FB OFF.

2.3 Experimental Protocol

The experiment was split into two sessions: The first session served to fit the sensor onto the prosthesis, to allow subjects to familiarize with the treadmill, and to determine individual stimulation thresholds for EMSSA.

To identify the minimal stimulation threshold, the duty rate for each pair was increased separately until the subject felt a sensation for the first time. To identify the maximal stimulation threshold, the duty rate for each pair was increased further until the stimulation was just uncomfortable. After calibration, subjects tested whether they recognized the stimulus and its movement direction in six randomized trials of seven seconds each. All subjects completed this test successfully.

The second session was scheduled about one week later. Stimulation thresholds were confirmed before measurements during standing and walking were performed. Resting breaks were allowed whenever required.

In the standing experiments, subjects stood in a comfortable, self-selected position with each foot placed on one force plate (Fig. 2, left), having the arms at one's sides. Subjects were instructed to stand as still as possible. To simulate everyday situations and their influence on postural control, in five different tasks the following conditions were combined: (1) Standing still with eyes open (EO) or closed (EC), to investigate the influence of visual input; (2) standing directly on the treadmill (HARD) or on a soft pad placed on the belt (SOFT), to investigate the effect of stable and unstable ground; (3) standing still while not counting (NOCOUNT) or starting at 100 and counting backwards by 3, 7 or 9 (COUNT), to investigate the influence of cognitive load (Table 1). Each of the five tasks was performed both with feedback OFF and ON, resulting in ten trials which were presented in randomized order, lasted 30 seconds and were not repeated.

In the walking experiments, subjects walked on the treadmill at a self-selected speed (1.3, 1.4, and 2.3 km/h, Fig. 2, right) and were instructed to walk as symmetric as possible. In three conditions, which were presented in randomized order and lasted two minutes each, subjects received feedback about either the prosthesis' CoP or the prosthesis' knee angle, or they were not provided with additional information. Prior to each condition, subjects were informed by the experimentor about the type of feedback they were about to receive. After the experiment, subjects filled in a questionnaire on how they experienced the feedback during standing and walking, e.g. whether it was comfortable or whether they profited from it.

Fig. 2 Subject in quiet stance on soft ground with eyes closed (left) and walking with knee angle feedback (right) on an instrumented treadmill. The dashed lines mark the location of the two force plates.

2.4 Data Processing and Analysis

Prior to analysis, force plate data was low-pass filtered using a fourth-order, zero-phase digital Butterworth filter with cutoff frequency 5 Hz. For standing tasks, the body weight index, two spatio-temporal CoP-based measures, and one dynamic measure were calculated. The first five seconds of each measurement were discarded, resulting in $N = 25000$ data points and $T = 25\,\mathrm{s}$ to analyze.

1. BWI is the ratio of body weight (w) borne by the sound (S) and the prosthetic (P) leg. It is a common measure to describe *weight bearing asymmetry*:

$$BWI = \mathbf{w}_S / \mathrm{w}_P \tag{1}$$

2. Δ_{RMS} reflects the *effectiveness* of the postural control system, with low values corresponding to a high effectiveness [37]:

$$\Delta_{RMS} = \sqrt{\frac{1}{N}\sum_{k=1}^{N}\Delta_{RES,k}^{2}} \tag{2}$$

with k the sample index of the recorded time series. $\Delta_{RES,k}$ is the distance time series

$$\Delta_{RES,k} = \sqrt{x_k^2 + y_k^2}, k = 1 \dots N \tag{3}$$

The CoP time series in medio-lateral and anterior-posterior direction, x and y respectively, are mean-centered, i.e. the arithmetic mean of each time series is subtracted before calculating $\Delta_{RES,k}$. Unless otherwise noted, Δ_{RMS} is calculated from the resulting trajectory of the CoP. When calculated from CoP trajectories for each foot, $\Delta_{RMS,S}$ and $\Delta_{RMS,P}$ reflect how effective the involvement of S and P in postural control is [37].

3. v_{CoP} reflects the *efficiency* of the postural control system, with low values corresponding to a high efficiency [37]:

$$v_{CoP} = \sum_{k=1}^{N-1}\sqrt{\frac{(x_{k+1}-x_k)^2 + (y_{k+1}-y_k)^2}{T}} \tag{4}$$

Unless otherwise noted, v_{CoP} is calculated from the resulting trajectory of the CoP. When calculated from CoP trajectories for each foot, $v_{CoP,S}$ and $v_{CoP,P}$ reflect how efficient the involvement of S and P in postural control is [37].

4. SEn reflects the amount of attentive and automated balance control, respectively: low entropy indicates that postural control gets a lot of attention whereas high entropy indicates that it is an automated process [4, 30, 39]. Mathematically, SEn quantifies the regularity of a time series, with low values corresponding to a high regularity:

$$\mathrm{SEn}(m,r,N) = -\log\left(\frac{A(r)}{B(r)}\right) \tag{5}$$

In a data set of length N, $B(r)$ expresses how many times a data subset of m points repeats itself within a tolerance r. With $A(r)$ expressing how many times a data subset of $m+1$ points matches itself within the same tolerance r. Data was downsampled from 1000 Hz to 40 Hz and low-pass filtered (cutoff frequency 5 Hz) before using the algorithm provided on PhysioNet [19]. Following the criteria proposed in [30] resulted in $m = 2$ and $r = 0.1$. Calculated for the resultant distance time series $\Delta_{RES,k}$ (eq. 3), SEn complements spatio-temporal posturographic measures and is a means to assess the influence of augmented feedback on automatism of and attention devoted to postural control. Unless otherwise noted, SEn is calculated from the resulting trajectory of the CoP. When calculated from CoP trajectories for each foot, $\mathrm{SEn_S}$ and $\mathrm{SEn_P}$ reflect how much attention is paid to the involvement of the sound and the prosthetic leg in postural control [8].

To assess asymmetry during standing tasks, the ratio sound limb/prosthetic limb (S/P) was calculated for all three posturographic measures.

For walking tasks, GRF data was first decomposed into individual left and right profiles before calculating gait parameters. The algorithm used here is an advancement of the approach proposed by Davis et al. [7] which uses the mediolateral center of pressure ($\mathrm{CoP_{ML}}$ to differentiate between single and double limb support. Instead of using constant threshold locations, here, for each step they are re-calculated as $0.8 \cdot \max(\mathrm{CoP_{ML}})$ and $0.8 \cdot \min(\mathrm{CoP_{ML}})$ for the left and right leg, respectively. Thus, step width as well as foot placement variability are taken into account.

To quantify effects on amputee gait, stance duration SD and step length SL for each leg as well as the ratio S/P, stance duration ratio SDR and step length ration SLR, respectively, were calculated from the obtained GRF profiles [10, 25].

Owing to the small subject population, measures were calculated and evaluated individually for each subject. For standing tasks, no statistical analysis was conducted as for each condition and subject only one trial was recorded. For walking tasks, mean and standard deviation of the two gait characteristics SD and SL, and of the two corresponding ratios SDR and SLR, were calculated from all steps recorded per condition (52±6 steps). The percentage change between two opposing conditions, e.g. ON and OFF, was calculated for all measures. With restoration of symmetry being the main potential benefit of feedback, the change of the ratio S/P between the different feedback conditions was of particular interest. Only absolute differences larger than 5 % were considered as potential improvement or deterioration. Improvement was defined as restoration of symmetry, and is reflected in convergence of the ratio towards 1 Deterioration, in contrast, is reflected in divergence of the ratio away from 1.

3 Results

3.1 Postural Control

Table 2 summarizes how the five measures for all subjects (columns) and tasks (rows) change when feedback is provided. Improvements are represented by $+$ and $++$, deteriorations by $-$ and $--$ and small or no changes by o. For ratios, improvement implies that with feedback the ratio is closer to 1 than without feedback. For absolute measures, improvement is defined below. All subjects showed the weight bearing asymmetry known from the literature in all tasks, reflected in BWI > 1 [9, 22, 23].

	BWI			ratio Δ_{RMS}			Δ_{RMS}			ratio SEn			SEn		
Task	S1	S2	S3	S1	S2	S3	S1	S2	S3	S1	S2	S3	S1	S2	S3
1	−−	o	−−	++	−−	−−	−−	−−	−−	−−	o	++	−−	−−	−−
2	−−	−−	++	++	++	+	++	−−	−−	-	o	−−	−−	−−	++
3	−	++	+	++	−−	++	++	+	−−	−−	++	o	+	−−	++
4	o	-	-	−−	−−	++	++	−−	++	o	+	−−	−−	−−	++
5	−−	-	-	++	+	++	−−	−−	−−	o	−−	−−	++	o	−−

Table 2 $+/-$: Improvement/deterioration between 5 and 10 %, $++/--$ Improvement/deterioration >10 %. o: change <5 %. For BWI and the two ratios, improvement implies that with feedback the ratio is closer to 1 than without feedback. A decrease of Δ_{RMS} indicates increased effectiveness of postural control. An increase of SEn indicates reduced attentive and increased automated control.

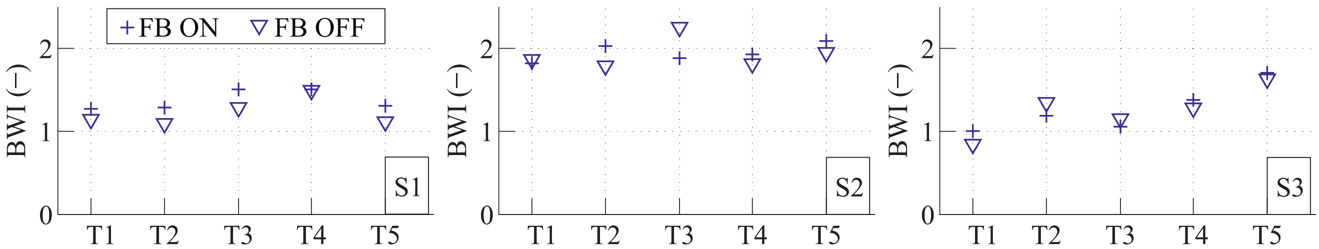

Fig. 3 Development of body weight index BWI for all three subjects (S1-S3) over all five standing tasks (T1-T5). Results for conditions with (cross) and without (triangle) FB are shown. A BWI of 1 reflects that both legs bear the same weight. Higher values reflect that more weight is put on the sound leg. T1: EC HARD NOCOUNT, T2: EC HARD COUNT, T3: EO SOFT NOCOUNT, T4: EO SOFT COUNT, T5: EC SOFT NOCOUNT

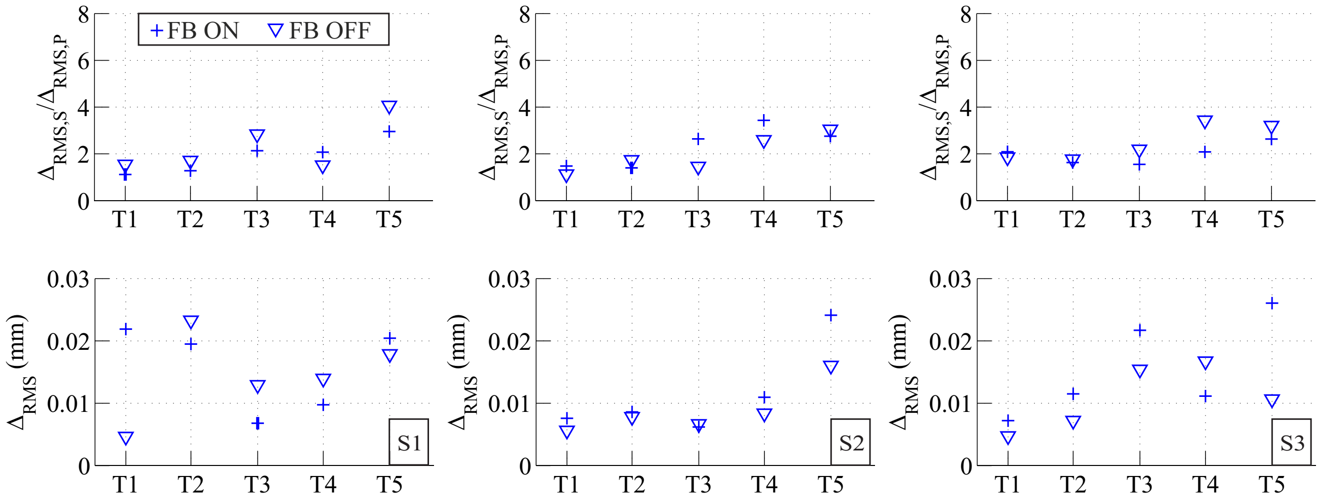

Fig. 4 Development of the ratio of root mean square distance $\Delta_{RMS,S}/\Delta_{RMS,P}$ (top) and resulting Δ_{RMS} (bottom) for all three subjects (S1-S3) over all five standing tasks (T1-T5). Results for conditions with (cross) and without (triangle) FB are shown. Lower Δ_{RMS} reflects more effective postural control. T1: EC HARD NOCOUNT, T2: EC HARD COUNT, T3: EO SOFT NOCOUNT, T4: EO SOFT COUNT, T5: EC SOFT NOCOUNT

In Δ_{RMS}, the asymmetry of postural control was visible as well: For all subjects, the ratio $\Delta_{RMS,S}/\Delta_{RMS,P}$ was around two and tended to increase with increasing difficulty, i.e. T1 to T5 (Fig. 4, top). Also resulting Δ_{RMS} had the tendency to increase with increasing task difficulty (Fig. 4, bottom). An improvement implies that with feedback resulting Δ_{RMS} decreases and, thus, effectiveness of postural control, increases. The ratio $v_{CoP,S}/v_{CoP,P}$ and the resulting mean velocity v_{CoP} qualitatively changed similar as reported for $\Delta_{RMS,S}/\Delta_{RMS,P}$ and resulting Δ_{RMS}, respectively (results not shown).

Asymmetry was also reflected in Sample Entropy, with $\mathrm{SEn_S}/\mathrm{SEn_P} > 1$ for al subjects and conditions (Fig. 5, top). For resulting SEn, no distinct tendency to change with task difficulty was observed. An improvement implies that with feedback resulting SEn increases and, thus, less attention is paid postural control and the amount of automated control increases.

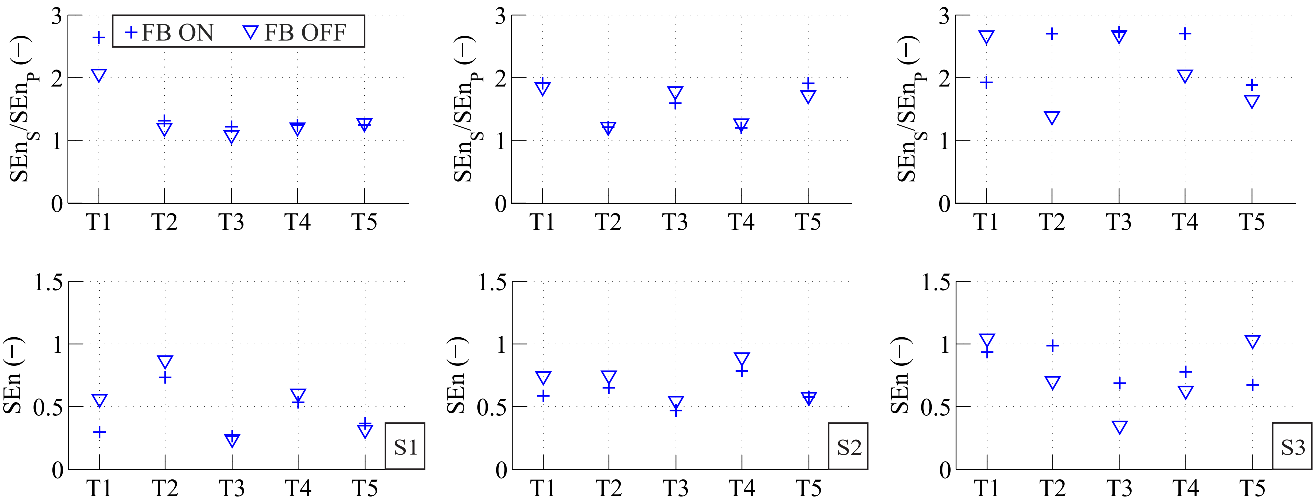

Fig. 5 Development of Sample Entropy SEn_S/SEn_P (top) and resulting SEn (bottom) for all three subjects (S1-S3) over all five standing tasks (T1-T5). Results for conditions with (cross) and without (triangle) FB are shown. Lower SEn reflects that balance control is less automated and that more attention is paid. T1: EC HARD NOCOUNT, T2: EC HARD COUNT, T3: EO SOFT NOCOUNT, T4: EO SOFT COUNT, T5: EC SOFT NOCOUNT

3.2 Gait Analysis

For all subjects and all conditions, SDR > 1 and SLR < 1 was found.

Compared to conditions without feedback, with one exception changes of SDR and SLR were below 10 % when CoP or angle feedback was provided (Fig. 6): For subject 2, SLR increased by 35 and 19 % for CoP and knee angle feedback, respectively. Compared to the reference condition without feedback, the absolute step length for both sides increased as well: By 38 and 48 % for the prosthetic, and by 87 and 76 % for the sound leg (not shown).

According to the questionnaire, all subjects were aware of the feedback during the experiment but did not feel distracted by it. Two subjects thought they had profited from EMSSA, in particular from knee angle feedback during walking. The third subject said he did not profit but also noted that he could not properly evaluate on this question.

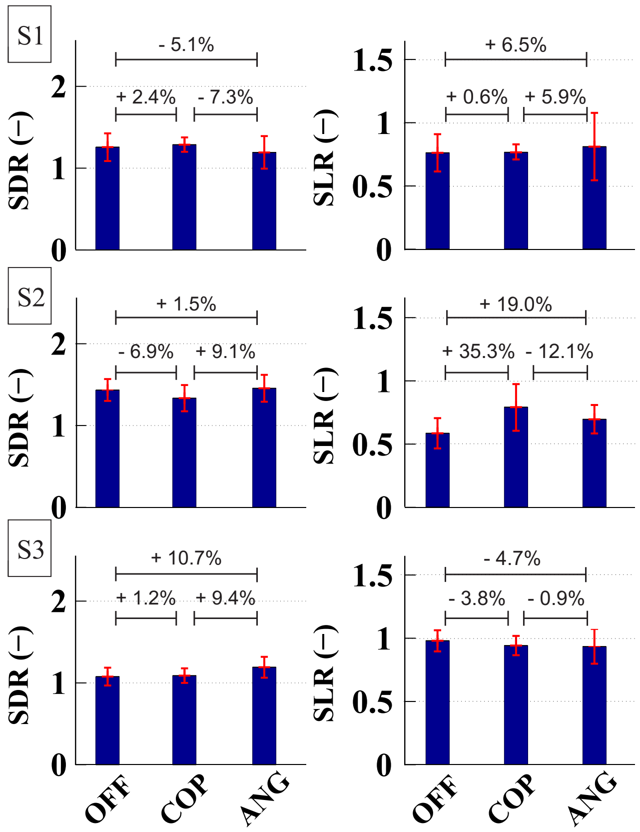

Fig. 6 Stance Duration Ratio SDR (top) and Step Length Ratio SLR (bottom) during treadmill walking for all three subjects S1-S3: mean and std of without feedback (left), with prosthesis' CoP (center) and with prosthesis' knee angle feedback (right). Percentage values indicate the change of the left with respect to the right condition, e.g. whether SDR for COP feedback (center) increased or decreased compared to SDR without feedback (left).

4 Discussion

4.1 Postural Control

In the literature, posturographic measures like BWI, Δ_{RMS} and v_{CoP} have shown that in lower-limb amputees and stroke patients the sound limb is more involved in postural control than the prosthetic limb [23, 41, 48]. Furthermore, it has been reported that the imbalance between impaired and unimpaired leg increases when vision is shut off or when cognitive load is enhanced [8, 23, 41, 48]. Despite the relatively small amount of data and the heterogeneous subject group tested, the evaluation of five standing tasks confirmed these findings.

It has been hypothesized that this asymmetric behavior in amputees and stroke survivors may compensate for the reduced sensory input from the affected leg [23]. Therefore, sensory augmentation using artificial feedback might be beneficial. In this work, we tested in three naïve amputee subjects whether an electrotactile moving stimulus can convey information about the CoP and the prosthesis' knee angle during standing and treadmill walking, even without training.

Providing feedback about the CoP underneath the prosthetic limb did not reduce the weight bearing asymmetry with two exceptions.

Also, Δ_{RMS} did change ambiguously but predominantly increased when EMSSA was provided, thus reduced the effectiveness of postural control.

The individual analysis of the two legs revealed that across all tasks the RMS distance for the prosthetic leg was smaller than for the sound leg, which is in line with the literature. However, the effect of feedback on the sound and on the prosthetic leg differed and predominantly decreased, i.e. the relative involvement of the affected leg increased. These effects cannot be due to the change in weight bearing: For S1 the loading of the sound side increased, whereas for S2 and S3 no change in BWI was found (Fig. 3). Hlavackova et al [22] reported a similar effect of mirror feedback on the CoP displacements underneath the sound limb. In their study, however, no changes were observed underneath the prosthetic limb.

The effects of EMSSA on the mean magnitude of CoP velocity v_{CoP} resemble those found for the RMS distance: Only for S1, feedback increased the efficiency of postural control, as indicated by a decrease of v_{CoP}, as well as the involvement of the prosthetic leg, as indicated by a decrease of $v_{CoP,S}/v_{CoP,P}$. For S2 and S3, feedback had a rather negative effect on efficiency of postural control. The involvement of the prosthetic leg did not change consistently, though: S2 rather decreased the involvement of the prosthesis, whereas for S3 no clear trend was found.

In this experiment, we were interested whether feedback increases or decreases the cognitive involvement, reflected by a change in SEn when EMSSA was provided. Here as well, the effect of feedback differed between subjects: When feedback was provided, S2 showed a clear decrease of SEn for all tasks. Two complementary explanations are possible. First, EMSSA may not be very intuitive, leading to increased attention devoted to postural control. Second, automatism of postural control may be reduced as the subject tries to consciously make use of additional information to influence postural control.

S1 and S3 did not show such a consistent tendency. However, in S3 SEn varied less between tasks when feedback was provided. This could indicate that independent on the natural sensory information available, the subject tried to incorporate EMSSA. This always required a similar amount of attention and postural control was automated to a similar extend.

SEn calculated for individual limbs can assess to which extend the use of the affected and non-affected leg for postural control is automated. For all subjects, we found a trend for SEn to be smaller underneath the prosthetic than underneath the sound limb which is reflected in $\mathrm{SEn_S}/\mathrm{SEn_P} > 1$. This complies with the understanding that impaired systems show greater regularity and decreased complexity [4]: The involvement of the prosthetic foot seems to be more conscious and its use requires more attention. When feedback was provided, in S1 and S2, both $\mathrm{SEn_S}$ and $\mathrm{SEn_P}$ tended to decrease, thus both legs were used more consciously. The effect on the ratio $\mathrm{SEn_S}/\mathrm{SEn_P}$, however, differed for these two subjects: S1 decreased the relative involvement of the prosthesis, i.e. the ratio increased, whereas S2 increased the involvement. In S3, no clear effect of feedback was found. However, similar to resulting SEn, the ratio varied less between the tasks when feedback was provided, suggesting again the conscious use of feedback information.

4.2 Gait Analysis

Asymmetry of amputee gait is well known in the literature [25, 39] and the results presented in section 3.2 are in line with these findings. Providing feedback about either prosthesis' COP or the prosthesis' knee angle, we tried to investigate whether enriched sensory information could potentially change gait characteristics and improve gait symmetry.

In this case description, feedback affected SLR and SDR only marginally and the direction of change, i.e. towards improvement or deterioration of symmetry, was not distinct.

Only in S2, the subject with the most distinct asymmetry, with either type of feedback SLR as well as step length for both legs increased. This result is insofar promising, as transfemoral amputees have a reduced step length compared to unimpaired subjects and feedback could be a means to change the prosthetic gait pattern towards more physiological gait. However, the effect was limited to step length and did not improve stance duration symmetry.

Despite the ambiguous objective results, all subjects had the impression that feedback changed their gait pattern and that they benefited from EMSSA, in particular from knee angle information.

5 Conclusion and Outlook

When EMSSA was provided during standing, the level of stability decreased, whereas the corresponding regulatory activity increased. Thus, in this case description, sensory feedback resulted in less effective and less efficient postural control. In the context of this very limited case description, the effect of feedback was comparable to the effect of closed eyes, increased cognitive load, or soft ground, and indicates that EMSSA was not beneficial. When EMSSA was provided during walking, only in one subject step length increased. Beyond that, no distinct effect was found.

On the one hand, these preliminary findings contradict our theoretical analysis which predicted that feedback information is redundant and has *no* influence on postural control and gait symmetry. On the other hand, they indicate that biofeedback may not exclusively show an effect in the early phase of rehabilitation [17] as two of three subjects were experienced users of a prosthesis.

This case description has shown that the effects of sensory feedback can also be assessed with Sample Entropy. In future studies, this measure could be used to evaluate the usefulness of a feedback modality especially for rehabilitation: the more intuitive a feedback design is, the less attention it requires, which is reflected in higher SEn. Applied to the feedback design tested in this study, decreased SEn indicates that information conveyed with continuously moving stimuli may actually be difficult to interpret and to process without training.

It is well known that it takes time and requires training before a lasting effect of sensory feedback occurs in the brain [40, 44]. Therefore, EMSSA may be beneficial after an appropriate training phase in both experienced and recent amputees. Following up on this preliminary experiment, a long-term study would be needed to enable general conclusions. First, the subject population should be more homogeneous and more participants should participate. Subjects could be randomly allocated to different groups that receive either continuous, discrete or no feedback. Second, the experimental protocol should include intensive training as well as baseline and retention tests. Furthermore, different types of information could be conveyed with EMSAA. For example, the moving stimulus could only be switched on as soon as weight bearing is asymmetric or as soon as the prosthesis can be loaded safely after swing phase. This would meet the preferences of our subjects who would appreciate feedback in situations where the principal focus is not on locomotion, such as gardening or shopping. To test feedback in daily life, a portable feedback system that can be used at home for a longer period of time would be required. Nevertheless, despite the limited amount of data and ambiguous objective findings, subjective reviews were already positive and encourage future work on sensory feedback systems for lower-limb amputees.

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