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Multivariable Control of Foot Motion During Gait by Peroneal Nerve Stimulation via two Skin Electrodes ⋆

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Abstract: Foot drop appears, for example, in stroke patients and is characterized by the limited ability to lift (the lateral and/or medial edge of) the foot, which leads to a pathological gait. We consider treatment of this syndrome via Functional Electrical Stimulation (FES) of the peroneal nerve during the swing phase of the paretic foot. We introduce a novel stimulation pulse waveform modification technique that allows us to manipulate the recruitment of m. tibialis anterior and m. fibularis longus almost independently via two single surface electrodes without violating the zero net current requirement. A piecewise linear controller output mapping is applied in order to cope with the nonlinearities in patients' stimulation intensity tolerance. The pitch and roll angle of the foot are estimated by means of an Inertial Measurement Unit (IMU) and controlled via a decentralized Iterative Learning Control (ILC) scheme. We demonstrate the effectiveness of our approach in experimental trials with stroke patients walking on a treadmill. Starting from conventional stimulation parameters, the controller automatically achieves physiological foot pitch and roll angle trajectories within only two strides.

Keywords: multivariable control systems, iterative learning control, biomedical systems, rehabilitation engineering, drop foot syndrome, neuroprosthetics, functional electrical stimulation, pulse waveform modification, validation by experiments

1. INTRODUCTION

In many cases, stroke leads to impaired motor function. Even after weeks of rehabilitation, many patients suffer from a limited ability to lift the inner (medial) or the outer (lateral) edge, or both, of the foot by voluntary muscle activation. This syndrome is known as drop foot (or foot drop), and it also appears in patients with other neurological disorders. As Figure 1 indicates, foot drop leads to a pathological gait with an increased risk of fall. A common treatment is to fix the foot in the lifted (dorsiflexed) position by an orthosis. While this approach may improve safety and stability in the patient's gait, it promotes muscle atrophy and joint stiffness.

Drop foot neuroprostheses, also known as peroneal stimulators, represent an alternative treatment that aims at generating a natural foot lift via activation of the patient's shank muscles, cf. Ring et al. (2009). The technology known as Functional Electrical Stimulation (FES) facilitates the artificial generation of action potentials in subcutaneous efferent nerves by applying tiny [1] electrical pulses via skin electrodes or implanted electrodes. By modulating the intensity of these pulses, one can control the contraction of paretic muscles and induce movements in the affected limbs. Unfortunately, FES may also trigger action potentials in afferent nerves, causing discomfort at medium and pain at high stimulation intensities. In most subjects, however, the sensation is weak enough to allow the generation of functional movements without discomfort. Abundant research demonstrates the potential of FES in neuroprosthesis design, beyond the application of drop foot treatment, see for example Peckham et al. (2005) and references therein.

1.1 Challenges in FES-based Drop Foot Treatment

For drop foot treatment, a few commercially available solutions make use of FES, some via skin electrodes, others via implanted electrodes. Most of these devices employ heel switches to detect two gait phases: one when the heel is on the ground and the other when it is not (Melo et al., 2015; Peckham et al., 2005). As soon as the heel is

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[1] When using skin electrodes, e.g., 20–50 rectangular current pulses are applied per second, each having an amplitude of less than a tenth of an Ampere and a pulse width of less than half a millisecond.

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lifted, the peroneal nerve is stimulated in each stride with a predefined, manually adjustable stimulation intensity. This approach is sufficient to lift the foot, but it ignores at least two fundamental issues. One is that the ankle joint has two degrees of freedom. Due to the anatomy of the shank muscles explained in Figure 2, it is challenging to find a stimulation that generates a straight foot lift without eversion or inversion of the foot.

Another important issue is that FES-activated muscles fatigue rapidly (Popovic et al., 2000). Moreover, residual voluntary muscle activity as well as, for example, the muscle tone (spasticity) in antagonistic muscles often change within a few strides. Therefore, it is just as challenging to maintain a physiological foot motion, once it has been achieved. An obvious escape strategy is to choose larger stimulation intensities and accept exaggerated foot lift and eversion. While this strategy provides a certain amount of safety and functionality, it accelerates muscular fatigue and leads to a noticeable peculiarity in the patient's gait.

1.2 Benefits of Feedback Control in Peroneal Stimulation

These challenges can be faced in a much more effective and elegant way by the use of feedback control. More precisely, the electrical stimulation can be adjusted automatically to generate a physiological foot motion. This requires measurement of the foot motion via, for example, a foot-mounted inertial sensor, as suggested by Veltink et al. (2003) for the calibration of an implantable drop foot stimulator. Closed-loop control of peroneal stimulation was successfully employed, for example, by using fuzzy control methods (Mourselas et al., 2000), run-to-run control (Negård, 2009), and predictive control (Hayashibe et al., 2011), respectively. However, most approaches aim only at providing a minimum foot clearance or a desired foot-to-ground angle at initial contact, and validation is often performed in simulations or in sitting subjects only.

Beyond these achievements, it was demonstrated by Nahrstaedt et al. (2008) that Iterative Learning Control (ILC) can be used to control the entire foot motion trajectory of the swing phase, i.e. from the last to the first ground contact. In Seel et al. (2013), ILC is used to track a desired pitch angle trajectory by FES of the tibialis anterior muscle, while eversion/inversion of the ankle joint is ignored. In Seel et al. (2014a), a three electrode setup is used to shape the roll angle trajectory of the foot at constant dorsiflexion.

In the present contribution, we extend these results by two major advancements. In Section 3, the number of required electrodes is reduced to two by using a novel stimulation pulse waveform modification technique. In Section 4, we introduce a piecewise linear control output mapping that exploits the entire range of stimulation intensity combinations accepted by the patient. With this mapping in place, a decentralized ILC scheme for the pitch and roll angle is designed in Section 5, which is then validated in a chronic drop foot patient walking on a treadmill in Section 6.

2. PERONEAL STIMULATION VIA SURFACE ELECTRODES DURING GAIT

The human ankle includes the talocrural joint and the subtalar joint. The former admits dorsiflexion and plan-

Fig. 1. In the pathological gait of drop foot patients, foot lift (pitch) is weak, and increased muscle tone often leads to an (additional) inwards roll of the foot. Both angles are defined with respect to level ground.

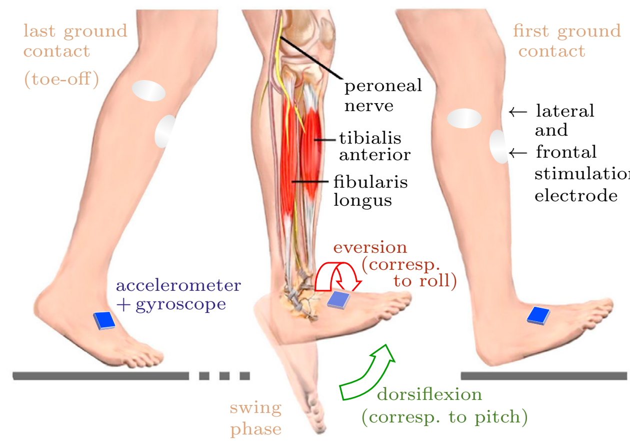

Fig. 2. Two muscles lift the foot during swing phase – one lifts mainly the inner edge, the other one the outer edge. Via electrical stimulation and inertial sensing, this motion can be controlled even in paretic limbs.

tarflexion, i.e. lift and drop of the foot with respect to the tibia. In contrast, the subtalar joint allows for eversion and inversion [2] to occur, which corresponds to a roll motion of the foot around its longitudinal axis, cf. Figure 2. The peroneal nerve divides into a superficial and a deep branch, which innervate the m. fibularis longus and m. tibialis anterior, respectively. Both muscles can be activated by FES via skin electrodes placed on the shank as depicted in Figure 2. Abundant experiments have taught us that it is hardly possible to predict precisely what motion is caused by FES in a specific patient on a specific day with a specific electrode placement. However, the following holds in almost all subjects: Stimulation via the frontal electrode causes dorsiflexion and often inversion. In contrast, stimulation via the lateral electrode causes eversion and, as one increases the stimulation intensity, often additional dorsiflexion. Therefore, a balanced stimulation via both electrodes is required to generate a straight, physiological foot lift without eversion or inversion. In terms of control theory, this defines a coupled multivariable system, in which the two stimulation intensities are the adjustable inputs. When inertial sensors are attached to the shank and foot, the ankle joint angle can be determined, see e.g. Seel et al. (2014b). In this case, the dorsiflexion and eversion angle of the foot represent the measurable outputs, as in Valtin et al. (2014a). If only the foot is equipped with an inertial sensor, the foot orientation with respect to the horizontal plane is known, and the pitch and roll angle of

[2] For the sake of brevity, we omit the distinction of pronation/supination versus eversion/inversion in this contribution.

the foot are the measurable outputs, cf. Figure 1. Since we find foot clearance with respect to the ground practically more relevant than ankle joint angles, we only consider the latter case in the following sections.

The primary objective of FES-based gait support in drop foot patients is to produce a physiological foot motion whenever the foot is in swing phase, i.e. from toe-off to initial contact. Since the dynamics of FES-induced motions are slow, the stimulation must start sufficiently early, i.e. typically already at heel-rise. Let the trajectories of the stimulation intensities $q_{\mathrm{fro}}(t)$, $q_{\mathrm{lat}}(t)$ applied via the frontal and lateral electrode, respectively, be denoted by

$$\begin{aligned}\mathbf{q}_{\mathrm{lat},j} &:= [q_{\mathrm{lat}}(t_{\mathrm{hr},j}), q_{\mathrm{lat}}(t_{\mathrm{hr},j}+t_{\mathrm{s}}), \ldots, q_{\mathrm{lat}}(t_{\mathrm{ic},j})]^T,\\ \mathbf{q}_{\mathrm{fro},j} &:= [q_{\mathrm{fro}}(t_{\mathrm{hr},j}), q_{\mathrm{fro}}(t_{\mathrm{hr},j}+t_{\mathrm{s}}), \ldots, q_{\mathrm{fro}}(t_{\mathrm{ic},j})]^T,\end{aligned} \tag{1}$$

where j is the stride index, t_{s} is the sample time, and $t_{\mathrm{hr},j}, t_{\mathrm{ic},j}$ are the heel-rise instant and the initial contact instant of the considered stride j, which are detected from the measured accelerations and angular rates of the foot (Seel et al., 2014c). We further denote the resulting foot pitch and roll angle trajectories as follows:

$$\begin{aligned}\mathbf{\Phi}_j &:= [\phi(t_{\mathrm{hr},j}+\delta t_{\mathrm{s}}), \phi(t_{\mathrm{hr},j}+\delta t_{\mathrm{s}}+t_{\mathrm{s}}), \ldots, \phi(t_{\mathrm{ic},j})]^T,\\ \mathbf{\Psi}_j &:= [\psi(t_{\mathrm{hr},j}+\delta t_{\mathrm{s}}), \psi(t_{\mathrm{hr},j}+\delta t_{\mathrm{s}}+t_{\mathrm{s}}), \ldots, \psi(t_{\mathrm{ic},j})]^T,\end{aligned} \tag{2}$$

where δt_{s} is the approximate delay of the input-output dynamics. In accordance with Figure 1 and without loss of generality, we define $\phi(t) > 0$ when the toes are above the heel, and we define $\psi(t) > 0$ when the outer edge of the shoe is above the inner edge. By further defining desired (physiological) pitch and roll angle trajectories $\mathbf{r}_\phi, \mathbf{r}_\psi$, we obtain a repetitive trajectory tracking task.

Due to increased muscle tone, there is an additional torque[3] acting on the ankle joint in many drop foot patients. This torque, as well as the residual voluntary muscle activity of the patient, act as a disturbance on the previously defined repetitive control task. Finally, please note that large acceleration and deceleration of the leg occur during swing phase. Therefore, also the (translational and rotational) motion of the shank imposes a disturbance on the considered system dynamics. Since the amount of force that can be generated by FES is typically very limited, these disturbances represent a major challenge.

From the preceding discussion we conclude that peroneal stimulation via surface electrodes during gait requires solving a repetitive, multivariable control task with large delays and disturbances. As discussed in Section 1.2, such control tasks have been solved in the single-input single-output case by applying Iterative Learning Control. Therefore, we tackle the present problem by defining a decoupling controller output mapping in Section 4 and designing a decentralized ILC scheme in Section 5.

3. STIMULATION PULSE WAVEFORM MODIFICATION

The force generated by FES increases monotonously with the frequency and the charge (i.e. the product of pulse width and amplitude) of the applied current pulses. Therefore, adjusting the stimulation intensity typically relates to adjusting either (or both) of these quantities. For the sake of brevity, we assume a fixed pulse frequency of 50 Hz and manipulate only the pulse charge. In order to avoid high and narrow pulses as well as low and wide pulses, we implement all stimulation intensity changes in such a manner that pulse width and amplitude are always increased or decreased by the same factor, i.e. their ratio remains constant.

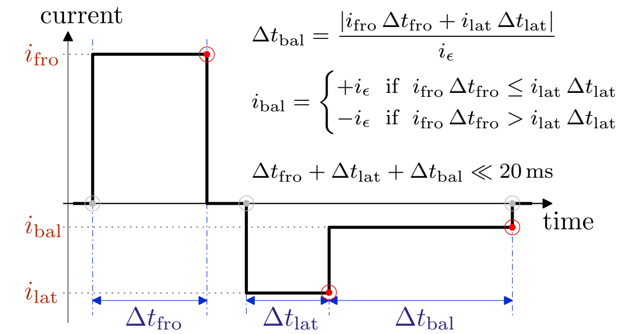

Fig. 3. Tri-phasic pulse wave for individual manipulation of the motion-triggering pulse charges $q_{\mathrm{fro}} = |i_{\mathrm{fro}}\,\Delta t_{\mathrm{fro}}|$ and $q_{\mathrm{lat}} = |i_{\mathrm{lat}}\,\Delta t_{\mathrm{lat}}|$, while assuring a zero net current via a third pulse below the motor threshold of FES.

Most FES devices employ two electrodes per stimulation channel and apply symmetric bi-phasic pulse waveforms, i.e. two current pulses of equal dimensions but opposite sign are applied subsequently. Thereby it is assured that, under each electrode individually, the balance of charge pumped into or out of the body is zero. This is a fundamental requirement, since a non-zero net current through the body is known to cause electrolysis and tissue damage in the long term.

An intuitive solution for the considered application is to choose the lateral electrode as counter-electrode (anode) for the pulse applied to the frontal electrode, and vice versa. However, as discussed in Section 2, it is required to find a pair of stimulation intensities (for the frontal and the lateral electrode) that yields a straight foot lift. Unless the required intensities are exactly equal, this demand is incompatible with the zero-net-current requirement. Therefore, a three-electrode setup was used in Seel et al. (2014a) in which the third electrode serves as common counter electrode for the frontal and the lateral electrode. Due to recent developments, this setup can be reduced even further. Using the versatile stimulator introduced in Valtin et al. (2014b), we can apply the tri-phasic pulse waveform depicted in Figure 3. While the intensity of the first two pulses are chosen independently, the third pulse is defined such that it balances the sum of charge. Due to the underlying biological effects, each FES pulse triggers action potentials only if its current amplitude exceeds a threshold of about 5 mA. Therefore, by choosing a sufficiently small amplitude $i_\epsilon < 5\,\mathrm{mA}$, the third pulse can balance the charge without triggering additional action potentials and, thus, without influencing the foot motion.

4. PIECEWISE LINEAR CONTROLLER OUTPUT MAPPING

Before FES can be applied in a control system that automatically adjusts stimulation intensities, it is advisable to identify the maximum tolerated intensity of the sub-

[3] e.g. m. tibialis posterior may cause an inversion-promoting torque

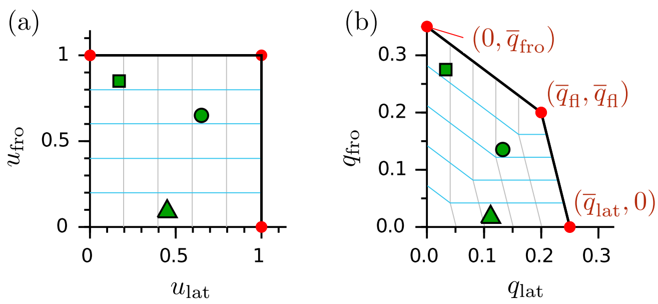

Fig. 4. The controller outputs $(u_{\mathrm{fro}}(t), u_{\mathrm{lat}}(t))^T \in [0,1]^2$ are mapped into the domain of admissible stimulation intensities defined by the subject's maximum tolerated values $\bar{q}_{\mathrm{lat}}, \bar{q}_{\mathrm{fro}}, \bar{q}_{\mathrm{fl}}$. Green markers indicate example value and respective mapped values, cf. (3).

ject, i.e. the maximum pulse charge that causes neither discomfort nor pain, and make sure that this value is never exceeded. In the current application, these values are easily identified for both stimulation channels individually. However, when stimulation is applied to both electrodes, the subject feels the combined sensation and typically tolerates only 70–80% of the single-channel maximum intensities. More precisely, let $\bar{q}_{\mathrm{lat}}$, $\bar{q}_{\mathrm{fro}}$, and $\bar{q}_{\mathrm{fl}}$ be the maximum tolerated stimulation intensities[4] for the lateral electrode, for the frontal electrode, and for equal stimulation via both electrodes, respectively. Then, with linear interpolation, Figure 4b depicts the domain of admissible stimulation intensities $q_{\mathrm{lat}}(t), q_{\mathrm{fro}}(t)$ of the lateral and frontal electrode.

This gives rise to the question how the intensity should be limited in multivariable control of foot orientation via FES. If the stimulation intensities $q_{\mathrm{lat}}, q_{\mathrm{fro}}$ are used as controller outputs, then the saturation limit of each depends on the current value of the other. This problem can be avoided by suitable parameterization of the domain of admissible intensities. To this end, we define a two-dimensional, real-valued controller output $(u_{\mathrm{fro}}(t), u_{\mathrm{lat}}(t))^T \in [0,1]^2$ and set $q_{\mathrm{lat}}(t), q_{\mathrm{fro}}(t)$ according to the following piecewise linear mapping at each sample instant t (cf. Figure 4):

$$\begin{pmatrix} q_{\mathrm{fro}}(t) \\ q_{\mathrm{lat}}(t) \end{pmatrix} = \begin{cases} \begin{pmatrix} \bar{q}_{\mathrm{fl}} u_{\mathrm{fro}}(t) \\ \bar{q}_{\mathrm{fl}} u_{\mathrm{fro}}(t) + \bar{q}_{\mathrm{lat}}(u_{\mathrm{lat}}(t) - u_{\mathrm{fro}}(t)) \end{pmatrix} & \text{if } u_{\mathrm{fro}} \le u_{\mathrm{lat}} \\ \begin{pmatrix} \bar{q}_{\mathrm{fl}} u_{\mathrm{lat}}(t) + \bar{q}_{\mathrm{fro}}(u_{\mathrm{fro}}(t) - u_{\mathrm{lat}}(t)) \\ \bar{q}_{\mathrm{fl}} u_{\mathrm{lat}}(t) \end{pmatrix} & \text{if } u_{\mathrm{fro}} > u_{\mathrm{lat}} \end{cases} \tag{3}$$

With this mapping in place, we can now manipulate the new controller outputs $u_{\mathrm{fro}}(t), u_{\mathrm{lat}}(t)$ independently of each other within the range [0,1] without causing pain or discomfort.

5. DECENTRALIZED ITERATIVE LEARNING CONTROL SCHEME

From the previous sections, recall that $\phi(t)$ and $\psi(t)$ are the measured pitch and roll angle of the foot, respectively, and that the control objective is to adjust $\mathbf{u}_{\mathrm{fro},j}, \mathbf{u}_{\mathrm{lat},j}$ such that the pitch and roll angle trajectories $\mathbf{\Phi}_j, \mathbf{\Psi}_j$ resemble the desired foot motion $\mathbf{r}_\phi, \mathbf{r}_\psi$. Furthermore, we discussed in Section 2 that, despite some cross-coupling, stimulation via the frontal electrode causes mainly pitch, while stimulation via the lateral electrode causes mainly eversion. Therefore, we choose a decentralized approach and pair $\mathbf{u}_{\mathrm{fro},j}$ with $\mathbf{\Phi}_j$ as well as $\mathbf{u}_{\mathrm{lat},j}$ with $\mathbf{\Psi}_j$. For each of these two single-input single-output control tasks, we design an iterative learning controller in the following.

We assume that the pitch and roll angle measurements are available immediately. However, as discussed in Section 2, the system dynamics are slow and model knowledge is typically poor. Therefore, instantaneous feedback, i.e. using current measurement information to adjust stimulation intensities, is practically useless. Instead, the stimulation intensities for each stride must be chosen primarily based on measurement information from previous strides. In terms of ILC theory this means that we neither employ current-iteration feedback nor design methods that rely on a (precise) system model.

Let the number of samples between $(t_{\mathrm{hr},j} + \delta t_{\mathrm{s}})$ and $t_{\mathrm{ic},j}$ of stride j be denoted by n_j, and assume an upper bound $\overline{n}$ such that $n_j + \delta \le \overline{n}\, \forall j$. Then $\mathbf{\Phi}_j, \mathbf{\Psi}_j \in \mathbb{R}^{n_j \times 1}$. Since n_j is not known before the initial contact of stride j occurs, we must prepare full-length controller output trajectories $\mathbf{u}_{\mathrm{fro},j}, \mathbf{u}_{\mathrm{lat},j} \in \mathbb{R}^{\overline{n} \times 1}$ for each stride j:

$$\begin{aligned} \mathbf{u}_{\mathrm{fro},j} &:= [u_{\mathrm{fro}}(t_{\mathrm{hr},j}), \ldots, u_{\mathrm{fro}}(t_{\mathrm{hr},j} + (\overline{n} - 1)t_{\mathrm{s}})]^T, \\ \mathbf{u}_{\mathrm{lat},j} &:= [u_{\mathrm{lat}}(t_{\mathrm{hr},j}), \ldots, u_{\mathrm{lat}}(t_{\mathrm{hr},j} + (\overline{n} - 1)t_{\mathrm{s}})]^T, \end{aligned} \tag{4}$$

although only the first $n_j + \delta$ samples of both trajectories will actually be applied to the system.

In the first stride, when no measurement information from previous strides is available, a conservative strategy is advisable: We set both $\mathbf{u}_{\mathrm{fro},0}$ and $\mathbf{u}_{\mathrm{lat},0}$ to trapezoidal profiles with large amplitudes and a considerably short rise time. This assures a safe swing phase with sufficient (but most likely exaggerated) foot lift.

At the beginning of each following stride, we determine the deviations $\mathbf{e}_{\phi,j}, \mathbf{e}_{\psi,j} \in \mathbb{R}^{\overline{n} \times 1}$ between the measured angle trajectories $\mathbf{\Phi}_j, \mathbf{\Psi}_j \in \mathbb{R}^{n_j \times 1}$ and the respective reference trajectories $\mathbf{r}_\phi, \mathbf{r}_\psi \in \mathbb{R}^{\overline{n} \times 1}$ in the following manner:

$$\mathbf{e}_{\phi,j} := \begin{bmatrix} \mathbf{r}_\phi(1:n_j) - \mathbf{\Phi}_j \\ \mathbf{0}_{1 \times (\overline{n} - n_j)} \end{bmatrix}, \ \mathbf{e}_{\psi,j} := \begin{bmatrix} \mathbf{r}_\psi(1:n_j) - \mathbf{\Psi}_j \\ \mathbf{0}_{1 \times (\overline{n} - n_j)} \end{bmatrix}.$$

These error information vectors are then used to update the controller output trajectories $\mathbf{u}_{\mathrm{fro},j}, \mathbf{u}_{\mathrm{lat},j}$ according to the standard ILC learning law:

$$\begin{aligned} \mathbf{u}_{\mathrm{fro},j+1} &= \mathrm{sat}\left(\mathbf{Q}(\mathbf{u}_{\mathrm{fro},j} + \lambda \mathbf{I}_{\overline{n} \times \overline{n}}\, \mathbf{e}_{\phi,j}) \right), \\ \mathbf{u}_{\mathrm{lat},j+1} &= \mathrm{sat}\left(\mathbf{Q}(\mathbf{u}_{\mathrm{lat},j} + \lambda \mathbf{I}_{\overline{n} \times \overline{n}}\, \mathbf{e}_{\psi,j}) \right), \end{aligned} \tag{5}$$

where $\lambda \in \mathbb{R}_{\ge 0}$ is the learning gain, i.e. an adjustable controller gain, and $\mathrm{sat}(\cdot)$ is the element-wise saturation to the interval $[0,1]$. Furthermore, $\mathbf{Q} \in \mathbb{R}^{\overline{n} \times \overline{n}}$ is a symmetric Toeplitz matrix containing the Markov parameters of a second-order Butterworth low pass filter with a cutoff frequency denoted by $f_{\mathbf{Q}}$. Multiplying a trajectory vector by $\mathbf{Q}$ results in a non-causal low-pass filtering of the trajectory, but without introducing the time delay that is typically observed in causal filtering, see for example Elci et al. (2002). The low-pass filtering yields two advantages: It avoids the discomfort that is usually associated with sudden large increases in stimulation intensity. Beyond this, it improves the robustness of the ILC by restricting the learning process to the low frequency range in which at least some model knowledge is available, see for example Bristow et al. (2006).

The advantage of the above learning scheme is evident:

[4] scaled to the stimulator maximum pulse charge of about 20 μC

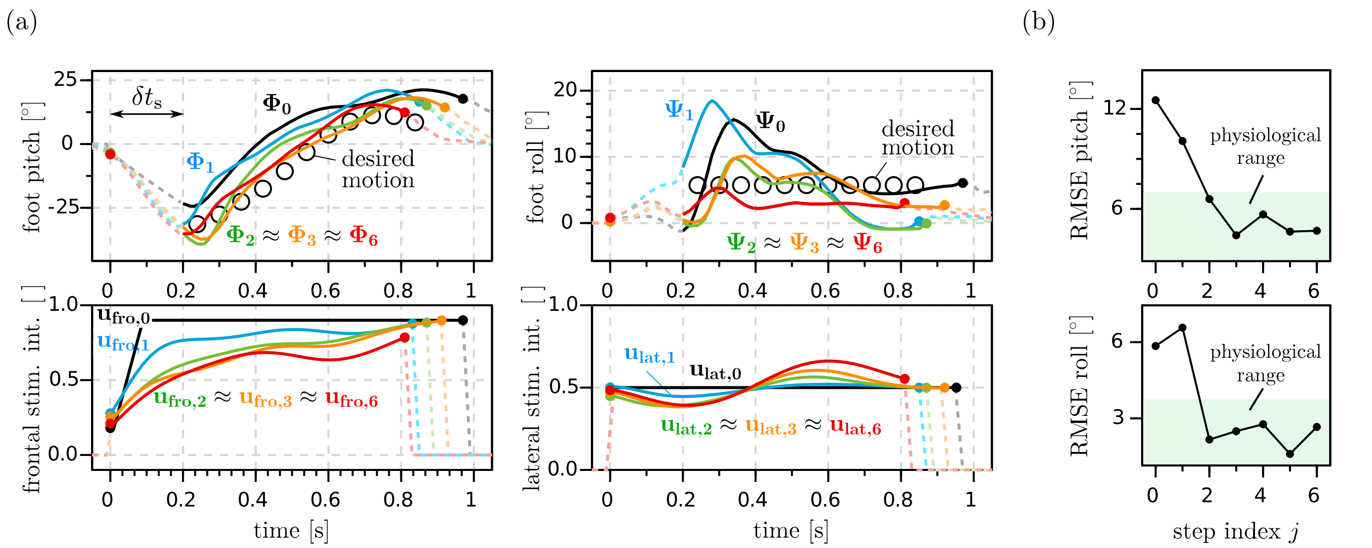

Fig. 5. Experimental results from a trial with a chronic drop foot patient. (a) The stimulation intensities are adapted from stride to stride in order to generate the desired foot motion during swing phase. Dots mark heel-rise and initial contact of each stride. (b) Root-mean-square errors of the foot pitch and roll (data from same trial) are quickly reduced to the ranges of natural variance found in healthy subjects walking at similar velocities.

When a certain section of an angle trajectory is lower than it should be, the update law (5) increases the corresponding section of the respective controller output trajectory. On the contrary, whenever the foot pitch or roll is stronger than necessary, the respective stimulation intensity is reduced. Please note that the time shift δt_s in the trajectory definitions (2),(1), together with the diagonal learning gain in the update law (5), partially compensates the delay in the system dynamics. If, for example, strong inversion occurs during the first five samples of the swing phase in stride j, the controller increases the first five entries of $\mathbf{u}_{\text{lat},j+1}$, which will be applied δ samples before the next stride's swing phase. Moreover, by integrating the control action saturation directly into the learning law (5), we prevent integrator windup effects that would otherwise occur whenever the stimulation intensities reach the boundary of the domain in Figure 4b.
The learning law (5) has been employed successfully for pure dorsiflexion and eversion control in Seel et al. (2013, 2014a). Thus, we utilize the following three results found therein: Choosing a Q-filter cutoff frequency of $f_{\mathbf{Q}} = 5\,\text{Hz}$ limits the learning to a reasonable frequency range; a phase-lead of $\delta t_s = 0.2\,\text{s}$ compensates most of the slow FES and muscle dynamics; and a learning gain of $\lambda = 0.5$ yields fast convergence without overshoot.

6. EXPERIMENTAL VALIDATION

In a first step, we compare the proposed system with two electrodes and pulse waveform modification technique to the three-electrode system employed in Seel et al. (2014a). While the setup time is found to be slightly shorter for the new system, we observe no differences in the maximum tolerated stimulation intensities $\overline{q}_{\text{lat}}$, $\overline{q}_{\text{fro}}$, and $\overline{q}_{\text{fl}}$ of drop foot patients and healthy subjects. Comparing the muscular response of both systems reveals only one difference: While both systems yield a functional foot lift with strong eversion at large q_{lat}-values, the novel system causes less inversion at combinations with large q_{fro} and small q_{lat}.

In a second step, we validate the decentralized ILC scheme in combination with the controller output mapping in experiments[5] with stroke patients. Each patient's individual values $\overline{q}_{\text{lat/fro/fl}}$ are determined and a wireless inertial sensor is attached to the paretic foot. The patient then walks on a treadmill at constant, self-selected speed. The measured accelerations and angular rates are used to calculate the current foot pitch angle $\phi(t)$ and roll angle $\psi(t)$ in realtime, cf. Seel et al. (2014a). For the first stride $j = 0$, we choose $\mathbf{u}_{\text{fro},0}$ and $\mathbf{u}_{\text{lat},0}$ as trapezoidal input profiles with amplitudes 0.9 and 0.5, respectively. After each stride $j \geq 0$, during the short period of time for which the heel and toes of the paretic foot are on the ground, the controller uses the measured pitch and roll angle trajectories $\mathbf{\Phi}_j, \mathbf{\Psi}_j$ to adjust the stimulation intensities $\mathbf{u}_{\text{fro},j+1}, \mathbf{u}_{\text{lat},j+1}$ automatically according to the update law (5). During the subsequent stride $j+1$, stimulation intensity trajectories $\mathbf{q}_{\text{fro},j+1}, \mathbf{q}_{\text{lat},j+1}$ are applied via the frontal and lateral electrode, according to the piecewise linear controller output mapping (3).
Throughout a large number of trials, the iterative learning controllers consistently adjust the stimulation intensities of the frontal and the lateral electrode in a way that the foot motion resembles the desired physiological motion. Results from a representative trial are depicted in Figure 5. The conservative initial stimulation intensity trajectories $\mathbf{u}_{\text{fro},0}, \mathbf{u}_{\text{lat},0}$ induce a foot motion with exaggerated foot pitch $\mathbf{\Phi}_0$ and roll $\mathbf{\Psi}_0$. Consequently, the stimulation intensities $\mathbf{u}_{\text{fro},1}, \mathbf{u}_{\text{lat},1}$ are reduced in the next stride. Since the resulting foot pitch $\mathbf{\Phi}_1$ still lies entirely above the desired reference trajectory, the intensity of the frontal electrode is further reduced. The foot roll $\mathbf{\Psi}_1$, however, is too high during the first half of swing phase and too low during the second half. Therefore, the iterative learning controller slightly reduces the first half of the intensity profile $\mathbf{u}_{\text{lat},2}$ and slightly increases its second half. In all

[5] Informed consent of the patients was obtained and the trials have been approved by the ethics committee of Charité Universitätsmedizin Berlin.

following strides $j > 1$, the deviations of foot pitch and roll were within the natural range that is observed in healthy subjects' gait. Due to the natural fluctuation of many FES and gait parameters, tracking accuracies below the few-degrees level are not achieved. By persistent adaptation, however, the controller maintains the physiological foot motion, even when the muscles fatigue or when the patient modifies his/her walking style [6]. An identical series of trials with another chronic drop foot patient yielded very similar results and confirmed all of the above findings.

7. CONCLUSIONS AND FURTHER RESEARCH

FES-based drop foot treatment via surface electrodes has been considered. Benefits of a closed-loop approach were discussed, as well as the challenges arising from the multidimensionality of this task and from large delays and disturbances. Furthermore, we characterized the domain of admissible stimulation intensities that is defined by the patient's maximum tolerated intensities. A piecewise linear controller output mapping was applied to facilitate decentralized iterative learning control of foot pitch and roll with independent input saturation limits and anti-windup. Furthermore, we have extended the state of the art by introducing a novel stimulation pulse waveform modification technique that allows us to recruit m. tibialis anterior and m. fibularis longus independently via only two single surface electrodes without violating the zero-net-current requirement. Experimental trials with chronic drop foot patients revealed that predefined pitch and roll angle trajectories are achieved by the controller within only two steps of learning. Unlike existing drop foot stimulators, the proposed system adapts the electrical stimulation to the needs of a specific patient on a specific day with a specific electrode placement and maintains a physiological foot motion even in the presence of disturbances.
Further research will focus on combining our achievements with recent results in peroneal stimulation via array electrodes (Valtin et al., 2014a). Moreover, in subjects with low FES tolerance, we will investigate the effect of input saturation and anti-windup schemes on the controller performance. Finally, variable gait velocity and walking stairs will be a focus of future research.

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[6] e.g. by increasing knee flexion and decreasing circumduction