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Automatic Control of a Drop-Foot Stimulator Based on Angle Measurement Using Bioimpedance

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Abstract: The topic of this contribution is iterative learning control of a drop-foot stimulator in which a predefined angle profile during the swing phase is realized. Ineffective dorsiflexion is compensated by feedback-controlled stimulation of the muscle tibialis anterior. The ankle joint measurement is based on changes in the bioimpedance (BI) caused by leg movements. A customized four-channel BI measurement system was developed. The suggested control approach and the new measurement method for the joint angle were successfully tested in preliminary experiments with a neurologically intact subject. Reference angle measurements were taken with a marker-based optical system. An almost linear relation between joint angle and BI was found for the angle range applicable during gait. The desired angle trajectory was closely tracked by the iterative learning controller after three gait cycles. The final root mean square tracking error was below 5°. **Key Words:** Iterative learning control—Functional electrical stimulation—Bioimpedance—Drop foot.

Stroke is a major cause for disability and death in Western countries. Many people have walking deficits after a stroke. Ineffective dorsiflexion during the swing phase (drop foot) is present in 20% of the population with partial recovery (1). A conventional treatment is an ankle–foot orthosis. Electrical stimulation for correction of drop foot was introduced by Liberson in 1961 (2). There, stimulation was applied via surface electrodes and was synchronized with the gait phases by means of a simple heel switch inside the shoe. Electrical stimulation was active during the swing phase when the heel was not in contact with the floor. Many drop-foot stimulators have been developed since Liberson's first study. Most commercially available stimulators still use a heel switch and are open loop during the swing phase (3).

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A disadvantage of such drop-foot stimulators is the manual adjustment of the stimulation intensity, which must be repeated when a training effect of the paretic muscles is observed or when the electrode positions have been slightly changed. An automatic compensation of slowly varying disturbances like muscle fatigue is also not possible. For safety reasons, the stimulation intensity of open loop devices is often set higher than required in order to guarantee sufficient dorsiflexion. However, this leads to a more rapid fatigue of the electrically stimulated muscle.

Closed-loop (feedback) control of the stimulation intensity has not been achieved in a satisfactory way until now, and would require a small sensor for measuring the ankle joint angle or the angle of the foot with respect to the ground. In a feedback-controlled system, the actual ankle joint angle is measured and compared to a desired angle profile. Based on the resulting angle error, a suitable stimulation intensity can be generated by the feedback controller. A closed-loop scheme, which is particularly easy to implement, is obtained when the stimulation profile is not adjusted permanently but only once after each step. Based on the difference between desired and measured angle trajectory, the stimulation profile

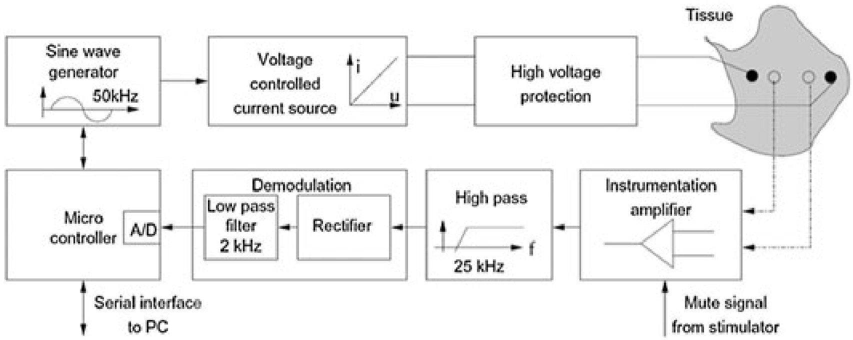

FIG. 1. Bioimpedance measurement system.

may either be just scaled in height and length or entirely updated. Such an approach is often called adaptive feedforward control. The reader should note, however, that it is only feedforward during the gait phase, and feedback adjustment occurs between successive steps.

In Veltink et al. (4) and Negård et al. (5), an inertial sensor, mounted on the shoe, is used to estimate the angle of the foot with respect to the ground. Based on this information, a controller adjusts the amplitude of the stimulation intensity profile to maintain the sagittal orientation of the foot before foot strike at a desired level. This approach has been successfully tested with one stroke patient (5).

Fuzzy control was used by Arifin et al. (6) to tune the duration of rectangular stimulation profiles during functional electrical stimulation gait while keeping the amplitudes constant. The aim of the control system was to obtain certain angle values at predefined time instants. This scheme was only tested in simulation; hence a real sensor was not used.

This contribution introduces bioimpedance (BI) as a tool to measure the ankle joint angle directly. The feasibility of employing such angle measurement for automatic control of the ankle joint angle is demonstrated. Iterative learning control (ILC) is applied to update the stimulation profile from step to step during gait to realize a desired angle trajectory. ILC represents a particular form of adaptive feedforward control and has been successfully used for control of the electrically stimulated upper limbs by Dou et al. (7).

MATERIALS AND METHODS

BI measurement

The passive electrical properties of tissue are summarized as BI. BI is measured by the voltage drop caused by a sinusoidal current flow through tissue. The use of real-time BI measurements to determine joint angles was initially proposed by Song et al. (8). Movements of the ankle joint cause changes in the tissue within the shank, which have an effect on BI. In a series of experiments, an almost linear relationship between the absolute value of the complex BI and the joint angle could be observed.

To realize the BI measurement, a four-electrode BI measurement system was developed, which could be used during activated electrical muscle stimulation. A sinusoidal current of 50 kHz with constant amplitude of 0.25 mA was generated by a programmable function generator connected to a voltage-controlled current source and was applied via two current electrodes. BI changes were measured by two voltage-sensing electrodes. The voltage was amplified by a customized instrumentation amplifier with high common-mode rejection. To allow measurement during stimulation periods, the amplifier inputs were short-circuited by PhotoMOS relays AQV248 (Matsushita Electric Works Ltd., Tokyo, Japan) during the stimulation pulses. Additionally, the amplifier possessed a fast artifact recovery below 1 ms to allow immediate BI measurements after the mute period. As the changes in BI were modulated to 50 kHz, low-frequency disturbances, which were caused by electromyographic and movement artifacts, were suppressed by a high-pass filter of 25 kHz. Finally, the absolute value of the BI was obtained by using an amplitude demodulation circuit. The measurement system was operated by a microcontroller that controlled the offset and gain of the BI signal before sampling in order to tune the signal quality. This resulting signal was sampled by a 12-bit analog-to-digital converter and was transmitted via an optically isolated serial interface to a control device (PC). The measurement frequency was set to 50 Hz to allow real-time control of the electrical stimulation. The structure of the BI measurement system is shown in Fig. 1.

Figure 2 shows the electrode placement for the muscle stimulation and the BI measurement. Two current excitation electrodes were placed on the

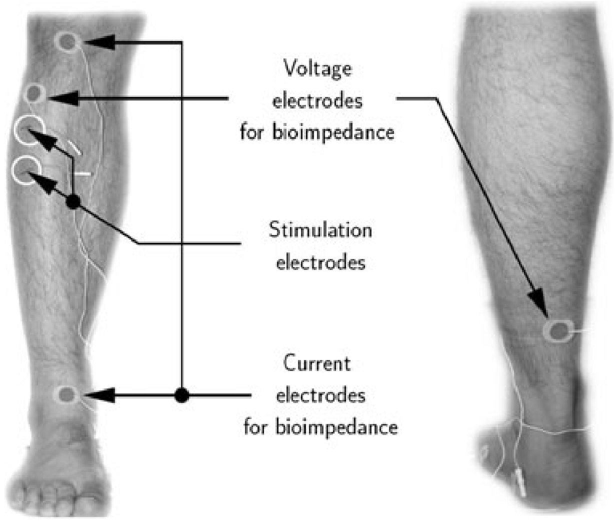

FIG. 2. Electrode positions.

anterior surface of the shank below the patella and on the transverse crural ligament. Voltage detection electrodes were attached to the muscle tibialis anterior close to its origin and to the posterior surface of the lower leg below the calf. The electrodes can be attached without taking off the trousers and shoes.

Experimental setup

Preliminary experiments with a neurologically intact subject were performed. The subject was one of the investigators. All experiments were in accordance with local ethical guidelines. As a first validation of the measurement system and the control strategy, a stationary walking pattern was investigated with the legs being lifted alternately. The ankle joint musculature was relaxed to simulate a drop foot.

A PC running Linux with RTAI extension[1] was used for data acquisition and real-time control. Automatic real-time code generation was performed by the tool chain RTAI-Lab (9) in combination with the open source program system Scilab/Scicos[2]. The externally controllable stimulator RehaStim[3] (HASOMED GmbH, Magdeburg, Germany) was employed to stimulate the dorsiflexor. This device was connected to the PC via an optically isolated USB interface and provided a trigger signal for the BI measurement system to mute the instrumentation amplifier during stimulation pulses. A stimulation frequency of 50 Hz was chosen. Gait phase detection at 50 Hz frequency was realized by means of the insole pressure measurement system Parotec (paromed Vertriebs GmbH & Co. KG, Markt Neubeuern, Germany) to synchronize swing phase and electrical stimulation.

The joint angle was measured via BI and, to provide a reference, by the optical motion capture system AS 200 (LUKOtronic Lutz-Kovacs-Electronics OEG, Innsbruck, Austria), which operates with active infrared markers.

Iterative learning controller

The control problem is to track a predefined angle trajectory for the ankle joint by stimulation of the muscle tibialis anterior during the swing phase of gait. The fact that walking is a cyclic movement can be conveniently exploited by applying an adaptive feedforward control scheme, namely, ILC. This works as follows: within one cycle, a fixed predetermined stimulation intensity profile is applied to the system (feedforward control). Before the next cycle starts, the error signal of the last cycle is analyzed. Based on the error signal, there will be an update of the stimulation profile for the next cycle. It should be noted that feedback is only applied between steps (cycles). There exist extensions of ILC with feedback during the cycles, but these are not considered here.

For controller design, a system model for each step is needed. The angle at time k in step j is regarded as the output $y_j(k)$; the stimulation intensity (modulated pulse width for constant frequency and amplitude) at time k in step j is the control input $u_j(k)$. The system can be approximately described as an asymptotically stable, discrete-time transfer function $P(q)$ of second order

$$\begin{aligned} y_j(k) &= \frac{B(q^{-1})q^{-m}}{A(q^{-1})} u_j(k) + d(k), \\ \underline{u} &< u_j(k) < \bar{u},\ j = 1, 2, \ldots, \end{aligned} \tag{1}$$

where q^{-1} is the backward shift operator ($q^{-1}u(k) = u(k-1)$); A and B are polynomials in q^{-1}; q^{-m} represents a time delay of m sample intervals, and d describes a disturbance that is independent of the cycle. The value of the control signal $u_j(k)$ is constrained to the interval $[\underline{u},\bar{u}]$. The lower-bound $\underline{u}$ describes the pulse width for which a recruitment of motor units starts. The upper-bound $\bar{u}$ represents the maximal tolerated stimulation intensity. Such a simple model is sufficiently accurate because of the limited range of the ankle joint motion during gait. This is demonstrated by the experimental results.

With the cycle-independent reference signal $r(k)$, the error signal $e_j(k)$ is defined as

1. http://www.rtai.org
2. http://www.scilab.org
3. http://sciencestim.sourceforge.net

$$e_j(k) = r(k) - y_j(k). \qquad (2)$$

ILC uses iterative search methods to find an input signal that forces the accumulated error $\sum_k |e_j(k)|$ in step j to decay over j. One possible ILC updating formula is given by

$$u_{j+1}(k) = \mathrm{sat}(Q(q)(u_j(k) + L(q)e_j(k+m))), \qquad (3)$$

where

$$\mathrm{sat}(u) = \begin{cases} \underline{u} & \text{for} \quad u < \underline{u} \\ \overline{u} & \text{for} \quad u > \overline{u} \\ u & \text{else} \end{cases}$$

is a model of the control signal saturation, and $L(q)$ and $Q(q)$ are linear discrete-time filters. The transfer function $L(q)$ is the so-called learning filter. It should be noted that both filters may be noncausal because Eq. 3 will only be processed after the jth cycle is terminated. Based on the model $P(q)$, the filters are chosen in such a way that the control system is stable (bounded and convergent control signal profile) and that monotone convergence of the error is guaranteed under the following conditions: (i) the number of sampling instants during each cycle is fixed and equal to N; (ii) the initial conditions of the system are the same at every step j, and (iii) the control signal is never saturated. More information about the controller design can be found in Bristow et al. (10).

In this application, the learning filter is set to a positive constant $L(q) = l_0$. Larger values of l_0 speed up the learning process but may also lead to instability. The Q filter was chosen as noncausal low-pass filter $Q(q) = Q_1(q)Q_1(q^{-1})$ without phase shift, where $Q_1(q)$ is an infinite impulse response filter of second order with a cutoff frequency of 15 Hz. This choice for the Q filter eliminates learning of the ILC at high frequencies, which improves the robustness of the ILC.

Implementation

The stimulation starts for the drop-foot stimulator with heel rise and ends with heel contact. In reality, cycle duration will vary from cycle to cycle. The number of sample instants N_j within cycle j is

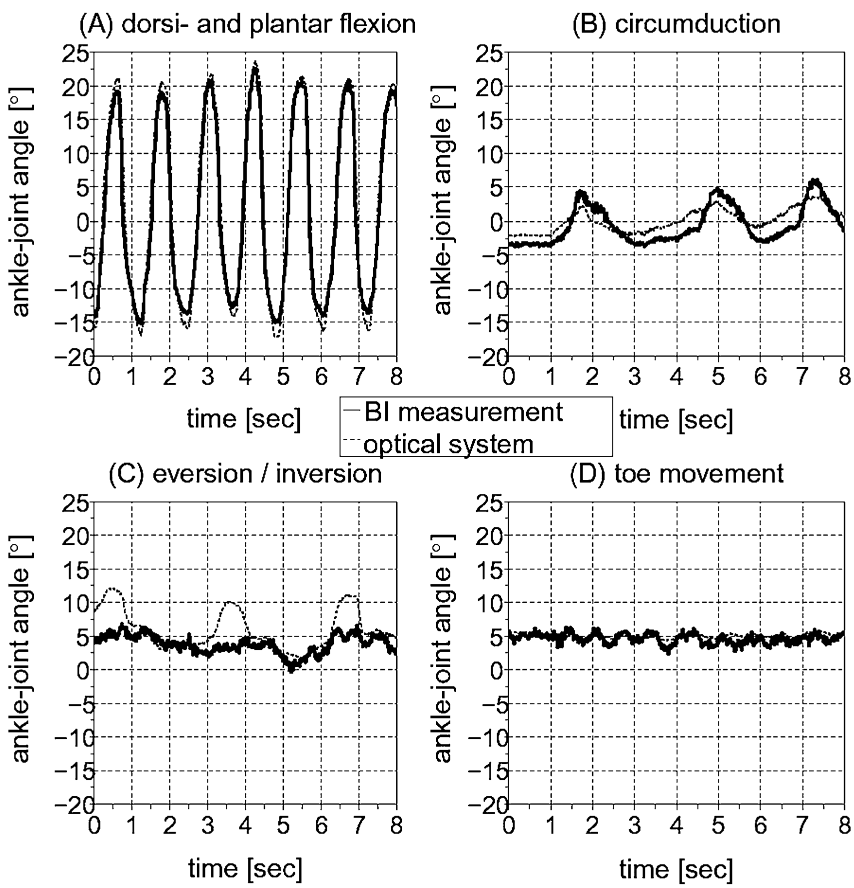

FIG. 3. Results of the sensor validation test.

therefore modeled as a random variable with mean N and a small SD compared to the mean. The mean N is determined by moving average. Two cases must be distinguished when the actual cycle duration N_j differs from the expected mean N:

Case 1 ($N_j < N$): The swing phase is shorter than expected. An update of the control signal $u_{j+1}(k)$ occurs only for $k \in \{1, \ldots, N_j\}$.

Case 2 ($N_j > N$): The swing phase is longer than expected. The applied control signal for the jth step is set to $u_j(k) = u_j(N)$ for $k > N$. An update of $u_{j+1}(k)$ occurs for $k \in \{1, \ldots, N\}$.

Experimental procedure

The following test procedure has been applied:

1. calibration of the insole pressure sensors;
2. calibration of the BI measurement using three static reference angle measurements (–20, 0, and 10°);
3. validation of the BI measurement for different movements of the leg, foot, and toes (sensor validation test);
4. experimental identification of the transfer function $P(q)$;
5. selection of l_0 yielding stability and convergence;
6. test of the ILC during stationary walking (ILC test): the subject was instructed to relax his dorsiflexor and to walk with constant cadence. Neither the desired reference trajectory nor the actual foot angle could be seen by the subject.

RESULTS

The results of the sensor validation test are shown in Fig. 3. In all subplots, the angle measured via BI is shown as a solid line. For comparison, the reference measurement by the optical system is shown as a dotted line. The upper left subplot shows an experiment, where active movements of the unconstrained ankle joint were recorded in a standing position (other leg on a pillar). The root mean square difference between both measurement signals for this test was 2.9°. The other three subplots illustrate the effect of different leg, foot, and toe movements on the ankle joint angle measurement. The circumduction of the leg (Fig. 3B), the inversion and eversion of the intertarsal joint (Fig. 3C), and movements of the toes (Fig. 3D) were investigated.

The results of an ILC experiment are shown in Fig. 4. The upper part shows the desired reference trajectory $r(k)$ (dashed line) and the BI signal $y_j(k)$ (solid line) for the swing phases of the first five steps.

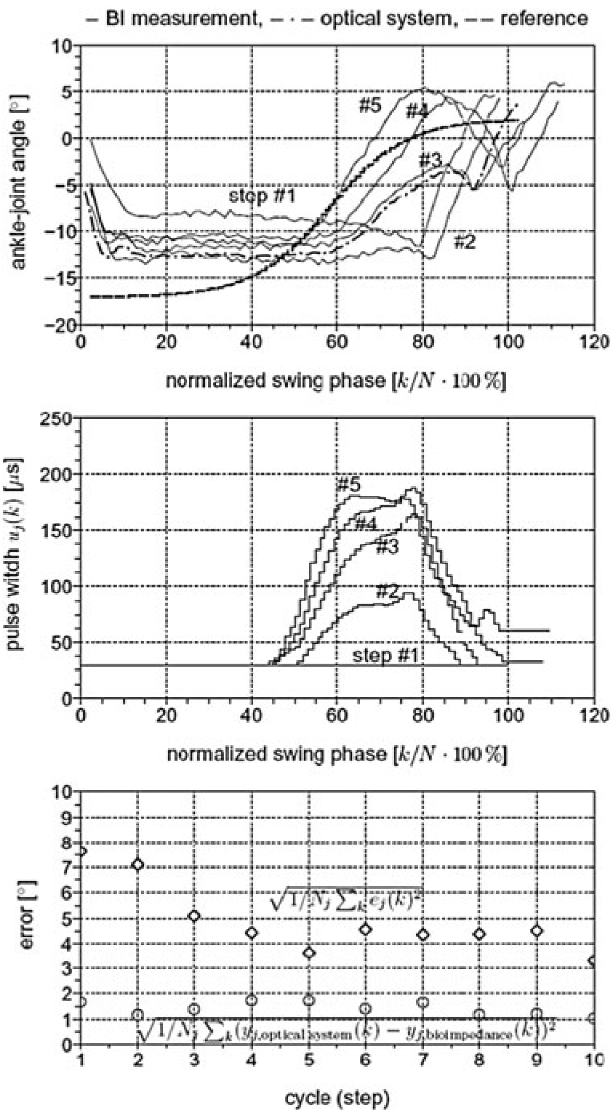

FIG. 4. Results of an ILC experiment with $l_0 = 8$.

The swing phases were normalized to the mean duration N. For the third step, the reference measurement of the angle is also plotted as dash-dotted line for comparison.

The control signal $u_j(k)$ for the five steps is presented in the middle part of Fig. 4. Lower and upper bound of the control signal are 30 and 400 µs, respectively, for a current amplitude of 20 mA.

The accumulated tracking errors of the ILC are shown in the lower graph as diamonds for 10 steps. The circles in the same graph represent the differences between the two measurement signals during the swing phase. The root mean square of the difference between the measurement signals over 10 steps including swing and stand phases was 1.4°.

DISCUSSION AND CONCLUSIONS

These first results indicate that BI is a promising tool to measure the ankle joint angle in a feedback-controlled drop-foot stimulator. The measurement is accurate enough for the considered application and is reliable for at least 2 h. Longer tests have not been performed until now. Other movements than dorsi- or plantar flexion have in general little or no effect on the recorded BI. An exception is the inversion of the foot, which causes errors in the angle measurement of up to 10°. Therefore, stimulation electrodes should be placed carefully to avoid an inversion of the foot.

The suggested feedback control approach worked reliably in tests with a healthy subject. The angle error profile converged within three steps to an acceptable value. Perfect tracking of the desired angle trajectory could not be achieved in the first phase of the swing phase as the joint angle was above the desired value so that plantar flexion would be required to decrease the angle.

The obtained results still need to be verified with stroke patients and under real walking conditions. A larger number of test subjects is necessary. A disadvantage of the presented approach is the large number of electrodes to be attached. An elastic stocking with integrated stimulation and BI electrodes is under development. This will greatly simplify the application of the presented technology. Future work will furthermore focus on gait phase detection based on BI recordings.

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