Increasing muscular participation in robot-assisted gait training using FES

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Abstract

Robot-assisted gait training provides automated gait training for individuals with gait impairments. One limitation is that active muscle contractions are not necessarily induced. Muscular contractions that are synchronized with the robot, which we term muscular participation, are required for effective rehabilitation. To induce muscular participation, we propose the use of functional electrical stimulation (FES) controlled by iterative learning control (ILC). ILC takes advantage of the repetitive nature of gait training by automatically adjusting the stimulation intensity according to errors occurring during previous stride cycles. In such a manner, stimulation can be modulated to account for individual differences in physiology. The force trajectory of the right leg obtained during a high effort task was used as a reference trajectory for the ILC to follow. We compared walking within a robot while exerting no volitional effort, normal effort, high effort and no volitional effort but FES applied. We found that the use of FES resulted in less error, higher correlation, and higher muscular participation compared to normal effort and especially to no volition effort. However, we also found that use of FES resulted in more error, lower correlation, and lower muscular participation compared to high effort, implying that additional means are required to maximize training.

***Keywords**: FES, Lokomat, ILC*

Introduction

Robot-assisted gait training of individuals with gait impairments provides reproducible gait patterns and relieves strenuous work of therapists. The legs are attached to a robot, such as the Lokomat (Hocoma AG, Switzerland) [1], that provides support and guidance during training. Only passive movements of the limbs that do not necessitate active contractions of the muscles are provided. To overcome this limitation, we propose adding functional electrical stimulation (FES) that induces active muscle contractions synchronized with the robot, a property we term muscular participation.

Because the therapist loses physical interaction that occurs during manual training, sensors integrated within the robot are relied upon to quantify the interaction between patient and robot. These measures can be shown to patients through visual biofeedback to encourage active participation [2,3]. The same biofeedback values can be used to assess the effectiveness of FES within the robot.

Our approach was to modulate stimulation to reproduce forces exerted within the Lokomat during high effort participation. A feedback control approach based on iterative learning control (ILC) was chosen because it could produce stimulation that compensated for unexpected limb movements and did not require extensive tuning or training of control parameters specific to an individual [4]. ILC automatically adjusts control inputs based on errors of previous cycles of a repetitive task.

Methods

The Lokomat guides the user along a predefined gait trajectory. Force transducers measure the interaction force between robot and user at each joint. Positive and negative interaction forces indicate movement by the user that is corrected by the robot in the direction of extension and flexion respectively. By considering the robot's direction of movement as well, one can determine whether the robot is assisting or resisting the movement. For example, positive interaction force during flexion indicates movement resistance by the robot, effectively providing resistance training. Such desired active movement was calculated with a muscular participation index (MPI), for each stride, similar to the biofeedback determined in [2]:

$$\mathrm{MPI} = w \cdot F + b$$

where w is a constant weighting vector, F is a vector of interaction forces for one stride, and b is an offset used for calibration. The weighting vector was chosen as in [2] so that active contractions in the direction of robot movement were positive.

Two channels of an FES stimulator (RehaStim, HASOMED GmbH, Germany) were used to induce muscular contractions about the right knee joint through 5 cm by 10 cm transcutaneous electrodes. For knee extension, one electrode was placed on the belly of the rectus femoris covering part of the vastus lateralis, and another was placed on the distal end of the rectus femoris covering part of both vastus lateralis and medialis. For knee flexion, one electrode was placed over the head of the biceps femoris and semitendinosus and another over the long head of the biceps femoris and the belly of the semimembranosus.

Symmetric biphasic pulses were delivered at 40 Hz. For each individual, the maximally tolerated electric current and pulse width were determined at the beginning of the experiment. Thereafter, the current was set constant at this value, but the pulse width was modulated between zero and maximum by ILC, which was implemented using custom software (Scicos, INRIA, France and RTAI, Politecnico di Milano, Italy). Mean current was 38 mA and mean maximum pulse width was 425 µs.

Using ILC, the pulse width profile was updated as:

$$PW_{i+1}(n) = Q \cdot [PW_i(n) + L \cdot e_i(n+k)]$$

where subscripts indicate stride number, n the sample index within a stride, k a constant sample offset, PW the stimulation pulse width (positive and negative indicate knee flexor and extensor stimulation respectively), Q a matrix representing a filter, L a constant learning matrix, and e the error between the actual and reference force trajectories. For the first stride, the pulse width was zero so that the controller could increase stimulation gradually.

A control frequency of 40 Hz was used. Constant sample offset, k, was required to take into account delay between pulse width change and its effect. The Q matrix low-pass filter acted to prevent high frequency components of stimulation that could cause discomfort. The same empirically derived controller parameters were used for all individuals.

The interaction forces from the Lokomat's sensors were measured and recorded during four different conditions. During the first condition (no effort, NO), the individual was asked to completely relax and allow the robot to move the legs. During the second condition (normal effort, NE), the individual was asked to walk normally such that the leg movements matched that of the robot. During the third condition (high effort, HE), the individual was asked to walk with high effort to induce high muscular participation. During the fourth condition (FES), the individual was asked to relax the legs while FES was applied. 20 strides were recorded for each condition, with adequate resting time in between. Two healthy individuals unaccustomed to FES participated in this study. For each individual, the reference trajectory was obtained by averaging the force profile during HE.

Results

Typical results of force tracking with FES for one individual are shown in Fig. 1. Normalized pulse width (NPW) is normalized to the maximum pulse width obtained at the beginning of the experiment. Although convergence occurred within approximately 7 strides, force did not reach desired values at certain portions of stride, despite maximum pulse width intensity being delivered.

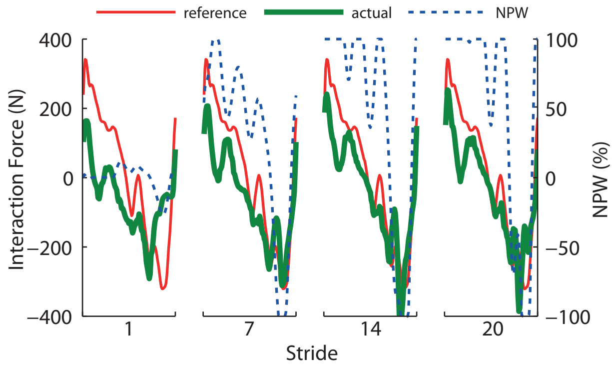

Fig. 1: Representative profiles of one individual during FES within the Lokomat. Stride begins with heel strike.

The stimulation pulse widths delivered with FES are shown in Fig. 2. From the beginning, the pulse widths quickly increased and began to plateau approximately to 90% at stride 15.

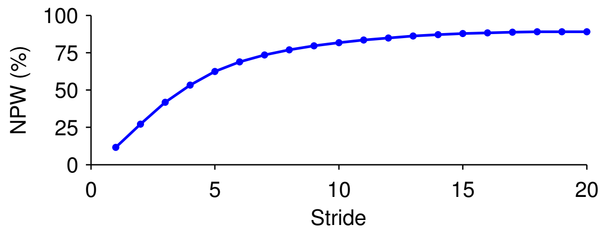

Fig. 2: Mean NPW of all individuals during FES within the Lokomat.

Error is expressed as normalized root-mean-square error (NRMSE), which is normalized between the minimum RMS error during HE and the maximum RMS error during NO. The mean NRMSE and correlation during all conditions are shown in Fig. 3 and Fig. 4, respectively. As expected, lowest NRMSE and highest correlation occurred during HE. During FES, the correlation increased and NRMSE decreased quickly from the beginning to obtain a stable value after approximately stride 5. Lower NRMSE and especially higher correlation was achieved compared to NO and even NE.

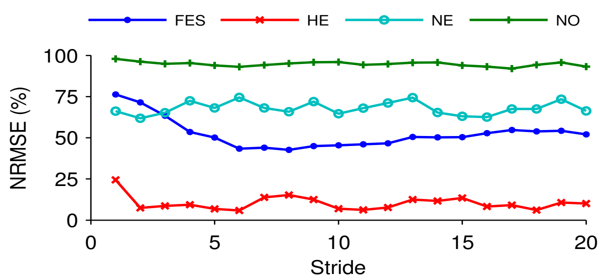

Fig. 3: Mean NRMSE of all individuals during all conditions.

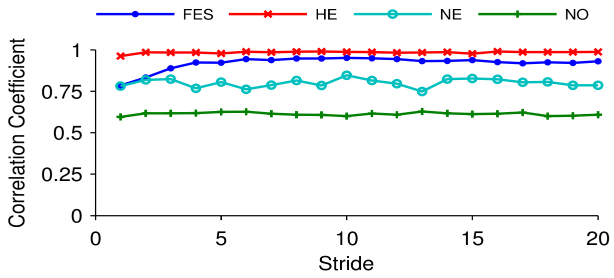

Fig. 4: Mean correlation coefficients between actual and reference force trajectories during all conditions.

The normalized MPI (NMPI) is normalized between the minimum value obtained during NO and the maximum obtained during HE. The mean NMPI values are shown in Fig. 5. Here too, the lowest values occurred during NO and the highest during HE. Intermediate values were during FES, which was higher than both NO and NE.

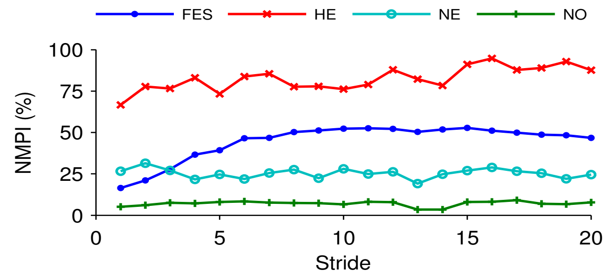

Fig. 5: Mean NMPI values of all individuals during all conditions.

Discussion and Conclusions

The obtained results show good force tracking by ILC. Although convergence is fast, performance is limited by saturating stimulation pulse width as can be seen in Fig. 1 during the first 25% of stride. The cause appears to be the inability of even maximum pulse width to increase force further. One way to overcome this limitation would be to conduct regular sessions of FES so that the individuals would become more accustomed to stimulation and thus be able to tolerate higher current amplitudes and longer pulse widths.

ILC resulted in converged stimulation that was synchronized with the robot and adapted FES to differences between individuals without the need to specify predefined stimulation patterns. With FES, the error, correlation, and muscular participation were all better than during NO, and even NE, showing that FES can be effective within the Lokomat to induce better training. Yet, FES alone was unable to achieve the values obtained during HE. This was expected since the participants were not accustomed to stimulation and thus only low stimulation intensity could be delivered comfortably. However, the high correlation obtained during FES indicates that ILC was able to stimulate with correct timing, but was hindered by the inability to produce strong contractions.

In this study, the reference trajectories were obtained from force recordings of the knee of healthy individuals during HE. In future work with patients, trajectories will be specified indirectly by therapists, who assist walking within the robot, which meanwhile records the interaction forces. Thus, the level of muscular participation will effectively be specified by the therapist who will modify it depending on individual ability and tolerance to FES. It is expected that as therapy progresses, higher MPI, well above zero, will be achievable. Moreover, the method will be extended bilaterally to all joints.

For the control of FES, joint angle trajectory following has thus far been the main focus of research [5]. Although such an approach is valid, the combination robot-assisted gait training with FES allows the focus of angle control to be shifted to the robot while FES can be used for muscle force control. The combination offers opportunities to explore different paradigms for FES such as force trajectory following and muscular participation to improve automated gait training.

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