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Motion Control of the Rabbit Ankle Joint with a Flat Interface Nerve Electrode

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Abstract

INTRODUCTION—A flat interface nerve electrode (FINE) has been shown to improve fascicular and subfascicular selectivity. A recently developed novel control algorithm for FINE was applied to motion control of the rabbit ankle.

METHODS—A 14-contact FINE was placed on the rabbit sciatic nerve ($n=8$), and ankle joint motion was controlled for sinusoidal trajectories and filtered random trajectories. To this end, a real time controller was implemented with a multiple-channel current stimulus isolator.

RESULTS—The performance test results showed good tracking performance of rabbit ankle joint motion for filtered random trajectories and sinusoidal trajectories (0.5Hz and 1.0Hz) with less than 10% average root-mean-square (RMS) tracking error, while the average range of ankle joint motion was between $-20.0^\circ \pm 9.3^\circ$ and $18.1^\circ \pm 8.8^\circ$.

CONCLUSIONS—The proposed control algorithm enables the use of a multiple contact nerve electrode for motion trajectory tracking control of musculoskeletal systems.

Keywords

motion control; ankle; sciatic nerve; nerve stimulation; functional electrical stimulation

Introduction

Spinal cord injury (SCI) causes muscle paralysis and loss of volitional movement due to failure of communication between the brain, where the motor commands are generated, and the muscles, the actuators for movement. Although there is no known fully restorative cure for SCI, functional electrical stimulation (FES) has proven to be effective in restoring function of paralyzed muscles by electrically activating either the muscles or the nerves that innervate them¹.

While both modalities are useful, nerve stimulation has several advantages over muscle stimulation. In the case of muscle stimulation, a stimulating electrode is placed on the surface of the muscle (epimysial electrode) or inside the muscle (intramuscular electrode) close to the motor point. Therefore, at least 1 electrode is required for each target muscle,

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and if the muscle covers a wide area, such as the pectoralis major, more muscle electrodes are needed. In contrast, a single multiple-contact nerve electrode can activate many muscles innervated by a single nerve. The reduction of the number of electrodes and connecting wires simplifies the surgical procedure and maintenance after implantation. In addition, nerve electrodes are less susceptible to mechanical failure and length-dependent stimulus threshold change due to muscle contraction. Stimulus threshold variation during muscle contraction is caused by a change of the distance between the electrode and the muscle motor point. Moreover, the threshold amplitude of nerve stimulation is 1–2 orders of magnitude smaller than that of muscle stimulation, which thus enables the application of much less current to achieve useful stimulation².

For selective control of different muscle activation using a single multiple contact nerve electrode, high spatial specificity of the electrode is desired³. A flat interface nerve electrode (FINE) is a multiple contact nerve cuff electrode designed to improve spatial selectivity by reshaping the cross-sectional geometry of a nerve trunk through reducing the distance between the electrode contacts and axons in the nerve⁴. FINE has demonstrated increased fascicular and subfascicular selectivity in both computational simulations and preclinical and clinical trials including intraoperative study of the human femoral nerve^{4–9}.

However, due to the complexities in neuromuscular-skeletal systems and neural interfaces, motion control of neuromuscular-skeletal systems using a FINE has not been demonstrated so far. In addition to the inherent complexities in neuromuscular-skeletal systems including high nonlinearity, time-varying properties, and strong coupling between joints, among others, undesired properties of electrical stimulation hinder development of control algorithms for functional electrical stimulation (FES) systems. In electrical stimulation of peripheral nerves, in contrast to natural neuromuscular-skeletal system control reversal of the recruitment order, charge spreads to untargeted axons, and there is increased muscle fatigue due to stimulation with a higher frequency than natural excitation frequency can occur¹⁰.

The difficulties in obtaining accurate models of the neuromuscular-skeletal systems makes it more complicated to control FES systems. Although the Hill-type muscle model is used most widely, it has been reported that error with this model can be as high as 50% in the physiologically relevant range of motor unit firing rates¹¹. Accurate modeling of a nerve is even more difficult due to the difficulties in finding fascicular distribution inside a nerve and the mapping of each fascicle to target muscles. Because of these modeling complexities, control methods that do not require an analytical model have been used for FES, including both open-loop and closed-loop methods for motion control of the feline ankle joint with sciatic nerve stimulation^{12–15}. However, in these studies, single contact electrodes were used and all nerve branches except for 1 or 2 that innervate the target muscles were cut for simplification.

Previously we developed an algorithm for motion control of musculoskeletal systems using only measurable stimulus parameters and output joint angles without prior knowledge of the systems¹⁶. Our control algorithm separates static properties of the neuromuscular system from dynamic properties, and finds their inverse models sequentially. The separation of

static and dynamic properties reduces system complexity and makes it easier to find an inverse model of a system with redundancy¹⁷. Since this control method does not require a complicated analytical model, it is suitable for FES control with a multiple contact nerve electrode.

In this study, the control algorithm was tested on rabbit ankle joint motion with a FINE on the sciatic nerve. For simplification, 2 contacts were selected: 1 contact for dorsiflexion and the other for plantar flexion. Monopolar biphasic stimulation was used for these contacts with a reference electrode placed on the contralateral hip. A real-time controller was implemented with a custom-built multiple-channel current stimulus isolator and a 14-contact FINE. The controller was tested by measuring the error between the reference and actual trajectories of the rabbit hind leg ankle joint for sinusoidal and filtered pseudo-random command trajectories.

Materials and Methods

A. Animal Preparation

Acute experiments were conducted on 6 New Zealand White rabbits weighing between 3.5 kg and 4.0 kg. The experiments were performed on 8 hindlimbs. All animal protocols were approved by the Institutional Animal Care and Use Committee (IACUC) of Case Western Reserve University. Animals were anesthetized initially with an injection of ketamine (50mg/kg) and xylazine (5mg/kg), and then maintained with 1~3% isoflurane mixed with pure oxygen or medical air.

A surgical incision was made on the posterior thigh to expose the sciatic nerve around the branch point to the common fibular and tibial nerves. Then a 14-contact FINE with 7 contacts on each side was implanted on the sciatic nerve proximal to the branching point. After suturing the incision, the rabbit was placed on the measurement instrument in the prone position. The foot was secured to the armature, and the ankle was allowed to rotate with little load, while the knee joint was maintained at approximately 90°. A needle electrode was inserted under the skin of the contralateral hip as a reference electrode. The experimental setup and the multiple-contact FINE used for the experiments are shown in Figure 1.

B. Instrumentation

Figure 2 shows the block diagram of the motion control of the rabbit ankle joint. The ankle joint angle (dorsiflexion/plantar flexion) was measured with an angle encoder (Futek TRS605) with a resolution of 0.25°. The real-time controller was implemented using LabVIEW in a PC with a hardware clock source in a data acquisition (DAQ) board (National Instruments PCI-6221). A multi-channel stimulator was built using a single-channel analog stimulus isolator (A-M Systems Model 2200) and analog multiplexers (MAX 308, Maxim Integrated Products, Inc.). A detailed description of the multi-channel stimulator and stimulus pulse waveform can be found in the supplementary material. The stimulation frequency was set to 30 Hz (a single biphasic stimulus pulse was delivered around every 33 ms) and only the pulse amplitude was modulated for control.

C. Controller Design

The controller is composed of 3 control blocks: an inverse steady state controller (ISSC), an artificial neural network (ANN) feedforward controller (FFC), and a PD feedback controller (FBC) as shown in figure 3. ISSC is an inverse model of the system at steady state, where the input to ISSC is the desired ankle joint angle and the output of ISSC is the stimulus amplitude of each contact of FINE. Typically, the neuromuscular-skeletal system to be controlled has a larger input dimension, in this case, the number of contacts of FINE, than the output dimension, which is the number of joints. Therefore, the combination of ISSC and the neuromuscular-skeletal system results in a reduced dimensional system. The ANN controller is a dynamic inverse model of the combination of ISSC and the neuromuscular-skeletal system. As a result, the combination of the ANN controller and ISSC becomes a dynamic inverse model of the system to be controlled.

The ISSC was built by linear interpolation of the steady state input-output data, which were obtained by changing the stimulation amplitude of each contact between the threshold amplitude and the maximum amplitude. For the ANN controller, a time-delayed neural network (TDNN) was used to approximate the inverse dynamics of the combination of ISSC and the system to be controlled. The input to the ANN controller at time step k with sampling period of T_s is the future desired trajectory as follows:

$$(r([k+1]T_s), \dots, r) = (r[k+1]), \dots, r[k+p]) \quad (1)$$

where $r(kT_s)$ or $r[k]$ is the desired output at time step k , and p is the number of tapped delays. The sampling frequency was set to 30 Hz, and thus the sampling period T_s was approximately 33 ms. The number of tapped delays was set to 5. For the training of TDNN, offline batch training was adopted with the Levenberg-Marquadt training algorithm in MATLAB® (The MathWorks, Inc., Natick, MA). The feedforward controller was trained with filtered pseudo-random trajectories with a total of 30–50s of data and validated for an additional 10s of data. The trajectories were divided into 10s segments with more than 30s of resting time between them in order to prevent any possible muscle fatigue during training. Filtered random trajectories were obtained by a fourth-order Butterworth low-pass filter with a cut-off frequency of 1 Hz.

A PD controller was used as a feedback controller to compensate for system variations and feedforward controller inaccuracy. The PD gains were tuned by trial and error by increasing the gain beginning from a low gain to prevent output oscillation or output error increment.

Performance of the controller was tested for both filtered pseudo-random reference trajectories and sinusoidal reference trajectories. The effects of each component of the controller were also compared. The performance results were evaluated by the output RMS errors and the time delay between the reference trajectories and the measured output trajectories. The time delay was calculated by finding the maximum cross-correlation between these 2 trajectories.

Results

Steady state responses

The ankle joint angle could be controlled with only 2 contacts: 1 for dorsiflexion and the other for plantar flexion (table 1). The contacts were chosen based on threshold amplitudes. An example of the steady state response is shown in Figure 4, where the stimulation of contact #1 and #6 for 1s generated dorsiflexion and plantar flexion angles, respectively, without load. The average output angular speed between 0.9s and 1.0s after stimulus onset was very slow ($1.6^\circ \pm 2.2^\circ/\text{s}$), indicating that the output angle almost reached steady state within a second. Therefore, the output angle at 1.0s after initiation of stimulation was selected as the steady state output joint angle for simplification. The average range of ankle joint motion was between $-20.0^\circ \pm 9.3^\circ$ to $18.1^\circ \pm 8.8^\circ$, and the shape of ISSC varied with each experiment.

Performance results of dynamic control

The performance of the controller was evaluated for filtered pseudo-random reference trajectories. An example of the control performance is shown in Figure 5. The RMS error was 3.6° , and there was no time delay between the desired trajectory and the measured system output trajectory. In order to show the improvement by adding feedforward and feedback control, the controller was also tested with only ISSC (Figure 5-b). Not only was there a large RMS error (10.0°), but there was also a significant time delay (5 tapped delays, equivalent to 165ms) between the desired command trajectory and the measured trajectory. Figure 5-(c) and 5-(d) show the normalized output errors and time delays for filtered random trajectories. The normalized average RMS error for ISSC only was $13.4 \pm 2.1\%$ ($n=8$), and that for our control method was $6.0 \pm 1.0\%$. These results indicate that dynamic control that includes both feedforward and feedback control improved the controller performance significantly ($P<0.001$, t -test). In addition, the time delay between the desired trajectory and the actual output trajectory was reduced significantly ($P<0.001$, t -test).

Dependency on the reference trajectory frequency

In order to evaluate the frequency characteristics of the controller, the performance of the ankle joint motion control was tested for sinusoidal reference trajectories. The frequencies of the sinusoidal reference trajectories were 0.5 Hz and 1.0 Hz, and their amplitudes were selected to provide the maximum range of motion. Examples of the control performance for the sinusoidal reference trajectories are shown in Figure 6. The RMS errors for the sinusoidal signals at frequencies of 0.5 Hz and 1.0 Hz were 3.5° and 5.6° , respectively, and the time delay was small (<12 ms in average) in both the 0.5 Hz and 1.0 Hz reference trajectories.

The normalized RMS errors were $7.2 \pm 1.6\%$ for 0.5Hz and $9.9 \pm 1.9\%$ for 1.0Hz. Although the output error increased for fast reference trajectories (1.0 Hz) with a significant difference ($P<0.05$), the average RMS error was still within 10% of the range of the reference trajectory. The time delay between the command signal and the actual output was very small in both reference trajectories (0.5 Hz and 1.0 Hz) due to the presence of the feedforward

controller and that ISSC was properly trained to approximate the inverse dynamics of the system.

Feedback control effect

To determine the role of feedback on the performance of the controller, the performance of the controller with or without the PID controller was measured for sinusoidal reference trajectories. The normalized RMS errors without a feedback controller were $6.7 \pm 1.7\%$ for 0.5Hz and $9.1 \pm 1.6\%$ for 1.0Hz. The time delays were 8 ± 11 ms for 0.5Hz and 5 ± 11 ms for 1.0Hz. There was no significant difference ($P>0.5$) in either the RMS output error or the time delay between the control result with and without feedback controller. This result is different from the simulation study of the human computational ankle joint system model, where significant improvement was obtained with incorporation of feedback control¹⁸. This discrepancy between the simulation study and the animal study might be due to small feedback gains used in the animal study. The small output error even without a feedback controller indicates that the inverse dynamics of the system were properly obtained by separation of the steady state properties and dynamic properties.

Discussion

Recently, we developed an algorithm to find an inverse statics model efficiently using measurable input and output data¹⁶, and we tested a motion control algorithm for FES control with a multiple contact nerve cuff electrode on a computational model of a human ankle joint system with good results¹⁸. The proposed control algorithm solved the redundancy problem by separating steady state properties from dynamic properties.

In this study, the algorithm was tested on motion control of the rabbit ankle joint with a 14-contact FINE on the sciatic nerve. Although ISSC reduced redundancy in steady states, ISSC alone showed relatively large error with latency in the actual ankle angular movement. The latency and dynamic error were much reduced by addition of feedforward and feedback controllers. Since input to the feedforward controller is the desired output at a future time, the future desired trajectories should be known *a priori*, similar to other artificial neural network control methods¹⁹.

The average output RMS error for filtered pseudo-random command trajectories was $<10\%$. Compared to the control results with only the ISSC controller, the proposed control method reduced the output error by more than half for the filtered pseudo-random trajectories. In order to evaluate the frequency-dependent performance, the controller was tested for tracking sinusoidal reference trajectories at 2 different frequencies of 0.5 Hz and 1.0 Hz. The output error increased as the frequency increased, but the average RMS error for 1.0 Hz was still $<10\%$. The results indicate that the controller can track reference trajectories with frequency components up to 1.0 Hz with small error. Sinusoidal reference trajectories with a frequency higher than 1.0 Hz were not tested, but it is expected that the output error will increase as the frequency increases. In order to track higher frequency command trajectories, higher torque is needed, and the maximum net torque that electrical stimulation can generate will determine the maximum acceleration and speed of the motion.

Due to inherent time delays in neuromuscular-skeletal systems, feedback control alone has limited bandwidth. Therefore, the addition of a feedforward controller into the feedback controller reduced the output error in similar experimental studies^{20,19}. In our experiment, there was no significant difference between controllers with feedback and controllers without feedback for sinusoidal reference trajectories, which indicates proper training of the feedforward controller. Less dependency on feedback control is beneficial considering the fact that accurate joint angle sensors are difficult to install outside of a laboratory environment. However, in the presence of output disturbance or system parameter variations, the contribution of the feedback controller increases, which was tested in a computational model of the human ankle joint system in our previous simulation study¹⁸.

The FINE has had satisfactory fascicular and subfascicular selectivity in both animal and human experiments^{4,21,9}. However, these selectivity studies were based on either EMG or torque measurements in isometric conditions lacking dynamic motion responses with respect to time. In this study, we demonstrated motion trajectory tracking control using the enhanced selective properties of FINE. Satisfactory performance of the rabbit ankle joint control indicates the applicability of FINE in human ankle joint control with a single FINE on the sciatic nerve. One of the advantages of the proposed control method is that it does not require an analytical modeling procedure. Therefore, it can also be applied to control of different joints, such as wrist and elbow joints, using multiple contact electrodes with little modification.

One limitation of the proposed control method is that the feedforward controller and ISSC were obtained in the training stage and remained fixed during control. Therefore, the time-varying properties of the system can be compensated only by the feedback controller. During the control experiments of the ankle joint, including ISSC development, feedforward controller training and feedback gain settings, which took typically less than an hour, the ankle joint motion response to stimulation did not change substantially. However, during the course of experiments, we noticed changes in the current threshold and range of motion. These changes can possibly be due to physiological change to nerve stimulation in the acute preparation. However, in previous chronic experiments on feline sciatic nerves, the response to the nerve stimulation remained unchanged over a long period of time⁶, suggesting that the control parameters would be more stable for chronic applications. Regarding the selection of contacts for dorsiflexion and plantar flexion, only 2 contacts were used in this study based on the threshold. Employing more than 1 contact for each direction may be beneficial not only for generating higher net torque by recruiting more muscles but also for reducing muscle fatigue by alternating the stimulus contacts. In addition, the contact selection based on the recruitment curve may improve the controller performance, since the output for the stimulation using the contact with the least steep slope of the recruitment curve is less sensitive to stimulus amplitude change. In this study, co-contraction was not measured, although unnecessary co-contraction can cause potential muscle fatigue. In our previous simulation study with direct muscle activation, we showed that co-contraction can be controlled dynamically¹⁶. By measuring individual muscle activation due to electrical stimulation of a peripheral nerve, it may be plausible to control co-contraction in addition to movement.

Another limitation of this study is that a single degree of freedom was controlled. In separate experiments, when 2 degrees of freedom of motion (dorsiflexion/plantar flexion and inversion/eversion) were allowed mechanically, inversion and eversion could not be activated selectively, probably due to anatomical characteristics of the rabbit ankle joint, or limited selectivity of the FINE, due to the small number and size of fascicles inside a nerve trunk, or both factors. Although this experiment concerns single degree of freedom joint control, >1 degree of freedom joint movement was controlled in the simulation study of the human ankle joint system. In this study, individual muscle selectivity of the FINE was not measured, but it would be intriguing to determine how individual muscle selectivity affects control performance in the future. In this simulation study, we show that output errors can be small even with limited individual muscle selectivity of the FINE¹⁸. For control purposes, functional muscle group selectivity seems to be more relevant than individual muscle selectivity⁸.

The acceptable errors in FES control for clinical applications are very important but difficult to determine since the permissible errors depend on the task type. For example, the FES system to prevent foot-drop requires foot clearance during the swing phase of walking, and if the ankle joint angle does not flex enough, the FES system may fail^{22,23}. However, in this case, the acceptable ankle joint error depends on the hip and knee joint angle, and thus control requirements need to be determined in connection with the kinematics and dynamics of the whole musculoskeletal system.

In conclusion, the proposed control algorithm enables the use of a multiple contact nerve electrode for motion trajectory tracking control of musculoskeletal systems. A possible future application is control of the human ankle joint system with a FINE on the sciatic nerve during gait for stroke or spinal cord injury patients.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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Abbreviations

FINE	flat interface nerve electrode
PF	plantar flexion
DF	dorsiflexion
ANN	artificial neural network
MIMO	multiple input multiple output

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Fig 1.

(a) Experimental setup. The rabbit is placed in the prone position, and the foot is fixed to the pivoting armature. A FINE is placed on the sciatic nerve proximal to the branching point to the common fibular and tibial nerves. The ankle joint angle is measured with an angle encoder. (b) The 14-contact FINE used for stimulation. The contacts on the bottom part are numbered from 1 to 7 from medial to lateral, while the contacts on the top are numbered from 8 to 10 from medial to lateral.

Fig. 2.

Block diagram of the closed-loop rabbit ankle joint motion control system. The controller generates control (*Ctrl*) and stimulation (*Stim*) signals, which are converted to a stimulation waveform for each contact of the FINE by the multi-channel stimulator.

Fig. 3.

The controller is composed of an Inverse Steady State Controller (ISSC), an Artificial Neural Network (ANN) feedforward controller, and a PID feedback controller. $r[k]$ is the desired output, $y[k]$ is the system output, $u[k]$ is the system input, $v[k]$ is the input to ISSC, $v_{ff}[k]$ is the feedforward controller output, and $v_{fb}[k]$ is the feedback controller output at time step k . (b) Offline training of the feedforward controller. In training, the time series of the system output is the input to the feedforward controller, and the input to ISSC becomes the desired output of the feedforward controller.

Fig. 4.

Examples of constant pulse amplitude stimulation at a frequency of 30 Hz (a) The output angle reached a stable value within a second after initiation of stimulation. (b) The steady state output angles for different pulse amplitudes. The circles and stars represent the measured output angles and their corresponding pulse amplitudes for dorsiflexion and plantar flexion, respectively, while the amplitudes of other contacts were zero. (c) The ISSC constructed from the steady state input-output data in (b). The solid lines are the interpolated pulse amplitudes of each contact for given output angles, whose information is stored in the ISSC in a table format. (d) Another example of ISSC for another hindlimb experiment.

Fig. 5.

(a) Example of performance results with only ISSC. The RMS error is 10.0° , and the time lag is 165ms ($33\text{ms} \times 5$ step). The desired trajectory is the dotted line, and the measured output trajectory is the solid line. At time 0, the joint angle was at resting state. (b) Example of performance results with feedforward and feedback controller. The RMS error is 3.6° , and the time lag is 0ms. (c) Normalized output RMS error. (d) Time delay between the desired trajectory and the actual trajectory. The RMS error and time delay were reduced significantly with the aid of the feedforward and feedback controllers.

Fig. 6.

Performance results for sinusoidal reference trajectories

The reference trajectories are sinusoidal signals with frequencies of (a) 0.5Hz and (b) 1.0Hz.

The upper graphs show the desired trajectories (dashed line) and the measured trajectories (solid line). At time 0, the joint angle was at resting state. The lower graphs show the pulse amplitudes of the contacts. The range of motion was between -35° and 35° . The RMS errors for 0.5 Hz and 1.0 Hz were 3.5° and 5.6° , respectively. Normalized output trajectories for 8 hindlimb experiments for (c) 0.5 Hz and (d) 1.0 Hz sinusoidal command trajectories. The gray line represents the desired trajectory, and black lines represent each experiment. An

example of intra-trial variations for (e) 0.5 Hz and (f) 1.0 Hz sinusoidal command trajectories. The gray line represents the command trajectory.

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Table 1

List of contacts selected for the stimulation to generate dorsiflexion and plantar flexion motion.

Experiment #	Animal #	Side	Dorsiflexion Contact #	Plantar flexion Contact #
1	1	L	9	11
2	2	L	9	11
3	3	L	9	12
4	4	R	8	11
5	4	L	2	5
6	5	R	1	6
7	6	R	2	5
8	6	L	2	5