Simple Quasi-static Control of Functional Electrical Stimulation-Driven Reaching Motions

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***Abstract*— Functional electrical stimulation is a promising technology for restoring functional reaching motions to individuals with upper limb paralysis. We present a control architecture that combines static models of a paralyzed arm and its response to stimulation with a PID controller. The controller is used to drive the wrist of an individual with tetraplegia to a desired wrist position. We compare the performance of our controller with a feedforward component and with no feedforward component. The combined feedforward-feedback controller produced an average accuracy (defined as the distance away from the target wrist position) of 4.9 cm, and the feedback controller produced an accuracy of 4.3 cm. The combined feedforward-feedback controller produced initially larger errors than the feedback controller, but the end performance was similar. Overall, the control architecture presented has the potential to be used for arbitrary reaching motions.**

I. Introduction

Individuals living with upper limb paralysis are limited in their ability to self-feed, groom themselves, and perform other activities of daily living. For these individuals, regaining the ability to use their arms and hands is the greatest priority for rehabilitation [1]. Functional electrical stimulation (FES) has become a promising technology for achieving this goal.

FES generates functional motion in individuals with spinal cord injuries (SCI) by stimulating the paralyzed muscles to activate in desired patterns. While hand function has been achieved with FES [2], success using FES to achieve full-arm reaching motions has been limited by the complexity of the shoulder and arm mechanics. Many upper limb control methods have been developed in simulation including using an optimized proportional-derivative controller [3], reinforcement learning [4], and threshold control [5]. To this point, however, much of the practically implemented controllers for the upper extremity have focused on controlling limited degrees of freedom (controlling elbow extension [6]), limited muscles [7], or treating each degree of freedom independently [8]. These methods are difficult to generalize to full-arm reaching motions because of the coupling of the degrees of freedom and their interaction with the muscles.

Multi-muscle models that treat the arm as a complete system have been developed for this reason [9]. Due to the complexity and dimensionality of a dynamic model, static models have been developed for control. These static models have been shown to be accurate in controlling 2-dimensional motions as long as the velocities remain low [10]. Our previous work has shown that a similar control strategy can be used to maintain static 3-dimensional wrist positions [11].

Expanding on the idea of using a static model-based controller to achieve planar motions, we propose using our previously developed controller to produce dynamic motions. We aim to use a quasi-static approach in which a series of static wrist positions are followed to move from a starting wrist position to a goal wrist position [12].

The purpose of this paper is to report the preliminary results of the quasi-static strategy to inform later improvements. We present a simple quasi-static controller where the only position on the path is goal wrist position. We use the controller to move the wrist to a series of target wrist positions after starting at a resting position.

II. Methods

We assessed the efficacy of our model-based controller over a single day of experiments. The controller used a model of the arm of an individual with tetraplegia to automatically determine the stimulation commands necessary to achieve a desired wrist position. The experiments took place over approximately two hours.

A. Experimental Setup

A single human participant who has high tetraplegia participated in our experiments. The participant was a 60-year-old female who sustained a hemisection of the spinal cord at the C1-C2 level. She cannot voluntarily move her right arm (the arm with which we performed our experiments) but does have sensation. She experiences hypertonia in some of the arm muscles. A passive arm support produces a comfortable and achievable workspace by using elastic bands to assist against the force of gravity. The arm support creates a resting equilibrium position with the wrist approximately at the height of and centered upon the participant's chest. More details can be found in [13] (Subject 1).

The participant is implanted with a stimulator-telemeter in her abdomen [14][15][16]. The device has leads which transmit current to intramuscular electrodes [17] and nerve cuff electrodes [18] activating muscles in her right arm and shoulder complex. We refer to each muscle or group of muscles stimulated by a single electrode as a muscle group. In this experiment, we controlled nine muscle groups including the triceps, deltoids, latissimus dorsi, serratus anterior, biceps

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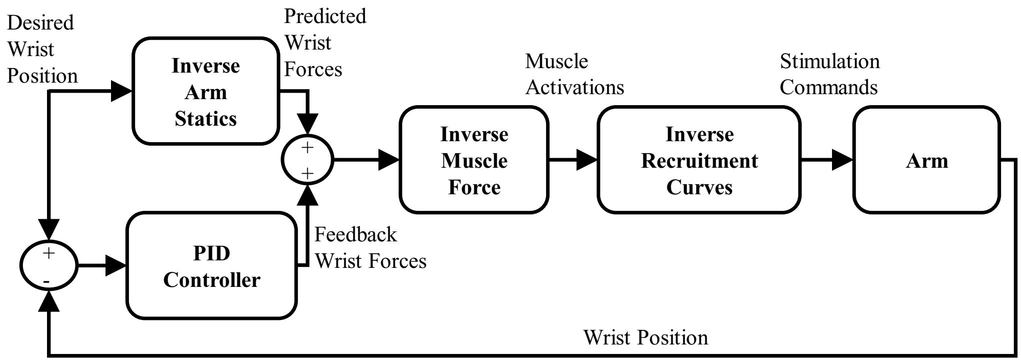

Fig. 1. Controller block diagram

and brachialis, supraspinatus and infraspinatus, rhomboids, lower pectoralis, and upper pectoralis. A computer sends power and control signals to the implanted device via an inductive radio-frequency link. Muscle stimulation uses biphasic, charge balanced pulses delivered at 13 Hz. The amplitude of the pulses is constant for each muscle group. The force generated by each muscle group is controlled by varying the pulse-width (referred to as the stimulation command) from 0-250 μs. The maximum stimulation command for each muscle was determined as the point when no additional muscle force was achieved or the participant reported discomfort. The control input is the vector containing the stimulation command for every muscle group. Stimulation commands are sent to the implant using real-time control code on a computer. Protocols used for this research were approved by the institutional review boards at Cleveland State University (IRB NO. 30213-SCH-HS) and MetroHealth Medical Center (IRB NO. 04-00014).

An Optotrak Certus Motion Capture System (Northern Digital, Inc.) captured data used to estimate the arm's configuration. The arm's configuration was defined by the position and orientation of the wrist relative to the thorax. The motion capture system was also used to measure the real-time position of the wrist to be used for feedback during the static hold experiments. A third-order moving-average filter was used on the wrist position signal to achieve smooth velocities.

The experiment was controlled using MATLAB xPC target on a Dell Dimension 8400 PC with a Pentium 4 3.20 GHz processor. The control and data collection occurred at 52 Hz, but stimulation inputs were updated at the stimulation frequency of 13 Hz.

B. Controller

Our controller (Fig. 1) aims to automatically determine the stimulation commands necessary to achieve a desired static wrist position. The controller was developed in detail in [11], but some details are repeated here for clarity. While the controller was developed and has been tested for static wrist positions, this study used the same control structure to achieve movement from a starting position to a different goal position.

The controller uses a subject specific, data-driven model of the arm's statics and response to stimulation. The model identification details are presented in [11]. The resulting model consists of three parts: the inverse arm statics (the mapping from a wrist position and orientation to the forces needed to maintain that wrist position), muscle force production (the amount of force each muscle can produce in a given configuration), and recruitment curves (the mapping from stimulation command to the muscle group activation). This model forms the basis for the blocks in the controller.

The input to the controller is the desired wrist configuration (defined as the position and orientation), q, which corresponds to the desired wrist position. The inverse arm statics model is then used to predict the wrist forces necessary to hold the desired static wrist position. The forces are then added to the compensation forces determined by a PID controller based on the error of the wrist position. The resulting output of the summation are the desired forces, $\mathbf{f}^*$, at the wrist needed to achieve the desired wrist position.

The desired force is mapped to the muscle activations needed to achieve the desired force in the "inverse muscle force" block. The muscle force production model forms the basis of this block. The output of the model is the linear mapping, $\mathbf{R}(\mathbf{q}) \in \mathbb{R}^{\mathbf{3}\times\mathbf{9}}$, of the muscle activation to the force at the wrist in each Cartesian direction for a given arm configuration. The j^{th} column of $\mathbf{R}(\mathbf{q})$ represents the force produced in each Cartesian direction by 100% activation of the j^{th} muscle group. To determine the required muscle activations, $\boldsymbol{\alpha} \in \mathbb{R}^{9\times 1}$ to achieve the desired force, we must find a solution to the equation

$$\mathbf{f}^* = \mathbf{R}(\mathbf{q})\boldsymbol{\alpha}. \tag{1}$$

$\mathbf{R}(\mathbf{q})$ is not square because there are more muscle groups than degrees of freedom. We resolve the redundancy in real-time using the quasi-Newton method to find the $\boldsymbol{\alpha}$ that minimizes the penalty function,

$$\begin{aligned} &||\boldsymbol{\alpha}||_2^2 + c_1\|\mathbf{R}(\mathbf{q}_*)\boldsymbol{\alpha} - \mathbf{p}(\mathbf{q}_*)\|_\mathbf{2}^\mathbf{2} + c_2 K \\ &K = \sum k_i \text{ where } k_i = \begin{cases} \boldsymbol{\alpha}_i^2 \text{ if } \boldsymbol{\alpha}_i < 0 \\ (\boldsymbol{\alpha}-1)^2 \text{ if } \boldsymbol{\alpha}_i > 1 \\ 0 \text{ if } 0 \le \boldsymbol{\alpha}_i \le 1 \end{cases}, \end{aligned} \tag{2}$$

where $||\boldsymbol{\alpha}||_2^2$ minimizes the muscle activations, $c_1\|\mathbf{R}(\mathbf{q}_*)\boldsymbol{\alpha} - \mathbf{p}(\mathbf{q}_*)\|_\mathbf{2}^\mathbf{2}$ penalizes activations that do not produce the desired force, and c_2K penalizes activations which do not belong to $\alpha_i \in [0, 1]$. c_1 and c_2 were chosen to be 500 and 50,000 respectively.

Once the required activations are found, the modeled recruitment curves are inverted to map the muscle group activations to the stimulation commands which produce those activations. These stimulation commands are then applied to the arm and the resulting wrist motion is tracked.

C. Desired Wrist Position Experiments and Analysis

We quantified the performance of the controller to achieve desired wrist positions at various targets throughout the participant's workspace. For each trial, the participant's arm began at the resting equilibrium. A desired wrist target position was selected and used as the input for the controller. It is important to point out that the predicted wrist forces and $\mathbf{R}(\mathbf{q})$ within the inverse muscle force block are based on the target configuration and are constant for a trial. The controller applied the determined stimulation commands to the arm for seven seconds with the goal of driving the wrist to the target wrist position. The position of the wrist was recorded throughout the trial.

Thirteen targets were selected from a 3×3 grid of targets which filled the subject's reachable workspace. The chosen targets were selected based on subject comfort as well as maintaining a targets spread throughout the workspace. Within each set of thirteen targets, each target was repeated twice, once with the predicted wrist forces all equal to zero (referred to as the feedback trials) and once with a combined feedforward-feedback controller (referred to as feedback+ trials). The experiment was repeated over four sets with a random order of the targets in each one.

To analyze the performance of the two controllers, the accuracy for each trial was determined by the Euclidean distance of the mean wrist position over the final second of the trial and target wrist position. The accuracy for a controller was calculated as the mean accuracy across all trials using that control strategy. Additionally, the system response was quantified by calculating the 5% settling time and the maximum error for each trial. The 5% settling time was determined as the time after which the distance between the wrist position and the target wrist position stayed within 5% of the final accuracy. The performance of the feedback and feedback+ controllers were compared using t-tests.

III. Results

The feedback+ and feedback controllers generally performed with similar accuracy and settling time. The feedback controller performed with a lower maximum error than the feedback+ controller. The maximum error is defined as the largest Euclidean distance of the wrist position from the target during a trial.

A time history of the wrist position during representative trials using each controller for a single target is shown in figure 2. As seen, the feedback+ controller initially drove the wrist away from the target position, but the feedback was able to compensate for that movement and drive the wrist back in the direction of the target. The feedback controller is able to more immediately drive the wrist towards the target without the large initial movement. Both controllers moved towards the desired position and finished with similar accuracy.

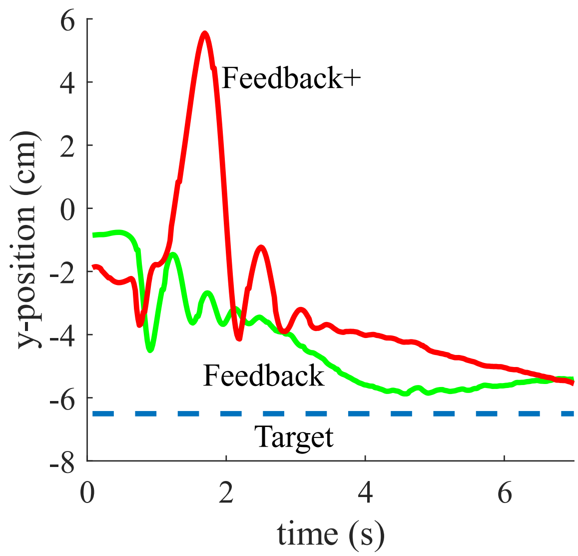

Fig. 2. Representative time history for a single target.

TABLE I
Comparison of Controllers

Mean (standard deviation)	Feedback	Feedback+	p-value
Accuracy (cm)	4.3 (3.3)	4.9 (3.5)	0.41
Maximum error (cm)	9.5 (4.5)	119.5 (5.3)	0.02
5% settling time (s)	5.7 (1.8)	5.8 (1.4)	0.84

The overall performances of each controller across all trials are compared in Table I. As seen, the accuracy of 4.3 cm for the feedback and 4.9 cm for the feedback+ were not significantly different ($p = 0.41$). The settling time of 5.7 s for the feedback and 5.8 s for the feedback+ were also not significantly different ($p = 0.84$). The only significant difference between the controllers was the maximum error of 9.5 cm for the feedback and 119.5 cm for the feedback+ ($p = 0.02$).

IV. Discussion and Conclusion

Overall, the control architecture presented in this study shows promise in controlling reaching movements, but some improvements are necessary. The average accuracy of the feedback controller of 4.3 cm is slightly worse than the accuracy of 2 cm which has been achieved in 2-dimensional reaching motions with a similar controller [10]. The additional degrees of freedom in 3-dimensional reaching motions, however, significantly increases the complexity of the problem. Thus, the limited decrease in accuracy with a significant increase in difficulty is encouraging moving forward with this control structure.

When comparing the two controllers presented in this study, the feedback+ controller often caused more aggressive initial movements which was seen by the overall difference in maximum error as well as in Fig. 2. This rapid movement away from the target is due to errors in the model. Due to the fact that the accuracy and settling times of the two controllers were similar, it may make sense to move forward with the feedback controller to eliminate the large error caused by the predicted forces in the feedback+ controller. However, it has been shown in simulation that using a combined feedforward-feedback controller results in smoother muscle activation time histories during a trial and better performance in general [19]. Smooth activation profiles are preferable as they would be more comfortable for the subject. Therefore, improving the feedback+ controller is a worthwhile pursuit.

A major reason for the error in the feedback+ controller may be because the configuration was assumed to be at the goal configuration. Since the feedback was able to push the wrist in the correct direction after a while in both controllers, this assumption seems valid for the inverse muscle force production block. However, the initial movement away from the target position seen in the feedback+ controller would be dominated by the predicted forces from the inverse arm statics block. It is likely that assuming the target configuration led to large errors in the predicted forces (i.e. the forces our model expects are necessary to achieve the position) relative to what was actually needed to move towards the target wrist position. Using the same control structure to maintain static wrist positions after starting in the desired configuration produced a better accuracy of 3.7 cm [11]. This also points to the fact that the controller performs better when the wrist is in the configuration expected by the model.

To improve the controller moving forward, we plan to use a quasi-static control method. In this method, a path of static points will be chosen from the starting position to the goal wrist position. By creating the path of points, the assumed configuration will be closer to the true current wrist position and thus the model should produce more accurate forces and thus better performance. Further research is necessary to determine the required distance between each static position and the amount of time necessary to move between positions. While dynamic trajectories could be useful and may improve the performance, dynamic models of the arm require significantly more data and model complexity. While the velocities must remain low, moving forward with a quasi-static controller would allow for many reaching motions to be achieved with a much simpler model and less demanding system identification.

Overall, we have demonstrated that our control architecture, with the stated improvements, is capable of moving the wrist to a desired wrist position.

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