Abstract

Introduction: Functional electrical stimulation (FES) has shown effectiveness in restoring movement post-stroke when applied to assist a patient’s voluntary intention during repeated, motivating tasks. However, most commercial upper-limb FES products assist only few muscles and do not use position sensor feedback to adjust the FES. Recent clinical trials have employed advanced controllers that precisely adjust stimulation applied to three muscle groups in order to assist functional reach and grasp tasks, giving rise to statistically significant reduction in impairment. This paper describes the system, focusing on the innovative sensing technology, electrode array based controller and associated hardware.

Method: Stroke participants (N=4) undertook seventeen intervention sessions, each of one hour duration. During each session FES was applied to the anterior deltoid and triceps via single electrodes, and wrist/finger extensors via an electrode array to assist participants in performing functional tasks with real objects and virtual reality. An advanced model-based iterative learning controller used kinematic data from previous attempts at each task to update the FES applied to each muscle on the subsequent trial. This produced stimulation profiles that facilitate accurate completion of each task while encouraging voluntary effort by the participant. Participants completed clinical assessments (Fugl-Meyer and Action Research Arm Test) pre- and post-intervention, as well as FES-unassisted tasks during each intervention session.

Results: Stroke participants’ results showed that FES-assisted performance increased over the course of the intervention for a range of functional tasks.

Conclusion: The feasibility of applying precisely controlled FES to multiple muscle groups in the upper limb using advanced sensors, controllers and array hardware was demonstrated.

1 Introduction

Stroke is the principal cause of adult disability in the UK, with an annual incidence of 152,000 people and a cost of £9 billion [12]. 77% of 1.2 million stroke survivors experience altered arm function, and 40% are left with a non-functional arm [9]. The majority of activities of daily living (ADLs) involve the upper limb. The capacity to achieve them has a direct affect on independence, reducing the social and financial burden of stroke [6].

Functional Electrical Stimulation (FES) can enable patients to practice meaningful, functional tasks and has been shown to improve range of movement, strength, and spasticity, while additionally having a positive effect on motor control [13]. Statistical evidence shows that benefits of FES are greatest when combined with maximum voluntary effort from the patient to assist their completion of the task [1, 11]. Unfortunately most upper extremity (UE) most of UE rehabilitation systems that have been clinically trialled employ open-loop or triggered control of one or two muscles [10]. This means they are unable to enforce the required connection between FES and voluntary intention, however to achieve precise movement requires advanced controllers. Iterative Learning Control (ILC) is one advanced control method that operates by comparing movement data from a previous attempt at a task to an idealised reference trajectory for the same task. Previous studies combining FES and ILC have demonstrated feasibility of using Pals Plus electrodes to deliver precisely controlled stimulation to the anterior deltoid, triceps and wrist extensors [8].

The system developed in this paper supports training of goal-oriented tasks through substantial innovations in the stimulation hardware, sensing equipment, control algorithms, and task display.

2 Methods

Stroke participants (N=4) undertook seventeen intervention sessions, each of one hour duration. As seen from Figure 1 participants sit on a perching stool in front of a touch table, and a SaeboMAS® arm support (Saebo Inc., Charlotte) is used to de-weight their upper extremity according to individual need and task during each session. A Kinect (Microsoft, Washington) and a PrimeSense (Apple Inc., California) are used to measure the position of joint centres within the shoulder, elbow and wrist. Data from these sensors are fed into the control algorithm hardware and software, which updates the FES control signals for each muscle group to provide just enough electrical stimulation. FES was applied to the anterior deltoid and triceps via single electrodes, and wrist/finger extensors via an electrode array to assist performance of functional tasks.

Functional tasks that are typically performed in everyday life were designed to offer a range of reaching challenges across the workspace. Figure 1 shows the four main images used on the touch table background; a default image, a bathroom sink, a coffee table, and a chopping board. Tasks comprised reaching, grasping and manipulating using real objects relevant to each image. There are five main tasks; closing a drawer, switching on a light switch, stabilising an object, button pressing and repositioning an object. These tasks include picking up a toothbrush, and moving the item to a different position on a representation of a sink displayed by the touch table.

The system software undertakes tracking of the participant’s movement in real-time, extraction of kinematic variables, and subsequent implementation of control schemes to adjust the FES in real-time. To accurately assist functional UE movement requires precise feedback control of appropriate joint angles. Kinect is employed to provide shoulder, elbow and wrist joint positions. However, it provides only a single joint position for the hand. Therefore, the GO-SAIL+ system also incorporates PrimeSense to collect the wrist position data and individual finger joint centre position data. The model employed in this research is a substantial development of UE models used in previous FES-based stroke rehabilitation research [8], and includes a comprehensive hand description. Joint angles $\phi_1, \ldots, \phi_5$ denote the orientation of the upper arm and forearm segments, with joint axes that are chosen to align with the motion elicited by FES. The procedure employed to define $\phi_1, \ldots, \phi_5$ is described in [5].

The FES hardware comprises Pals Plus electrodes, an electrode array, and electronic components to generate and route stimulation signals. The Pals Plus FES surface electrodes are positioned on the participant’s
anterior deltoid and triceps muscles, with placement following clinical guidelines [7]. Similarly the electrode array is placed on the participant’s forearm to actuate wrist and hand extensor muscles.

Figure 1: GO-SAIL+ system components.

2.1 Biomechanical Model

The reach and grasp tasks consist of repeated movements for the participant’s affected arm, with a rest period in between during which their arm is returned to a common starting position. The requirement to repeatedly perform a set of finite duration tasks enables ILC to be utilised to control the FES signals for each muscle group. Each ILC trial starts from a fixed initial arm position and the performance error from each trial is used to update the control parameters in an attempt to increase the accuracy of the subsequent attempt. The control signal is generated using kinematic joint information, in combination with a dynamic model of system described in [2], given by

\[
B(\Phi)\dddot{\Phi} + C(\Phi, \dot{\Phi})\ddot{\Phi} + F(\Phi, \dot{\Phi}) + G(\Phi) + K(\Phi) = g(u, \Phi, \dot{\Phi}) - J^T(\Phi)h
\]  

where \(B(\cdot)\) and \(C(\cdot)\) are 17-by-17 inertial and Coriolis matrices, \(F_h(\cdot)\) and \(G_h(\cdot)\) are friction and gravitational vectors, \(h_h\) is a vector of external force and torque comprising components from the spring support and interaction with objects, \(g(\cdot)\) is moment produced by muscles, and \(J(\cdot)\) is the system Jacobian. The vectors \(\Phi = [\phi_1, \ldots, \phi_{17}]^T\) shown in Figure 2 and \(u = [u_1, u_2, u_3, u_4]^T\) respectively denote joint angles and applied electrical stimulation, where \(u_1(t)\) and \(u_4(t)\) represent the electrical stimulation pulse-width applied to the anterior deltoid and triceps muscles and \(u_2(t)\) and \(u_3(t)\) represent the electrical stimulation pulse-width that is routed to elements within the electrode array. This model is next used by the FES control system to produce an input signal that results in accurate completion of each task.

Figure 2: Arm and support kinematic model.
2.2 Control System

Identification methods have been developed in previous research to establish parameters within a model of the form (1). However, the close proximity of muscles in the hand and wrist means it is not feasible to extend these to include the hand and wrist. Therefore, a simplified model structure is produced by assuming the inertial and Coriolis coupling between arm segments and the hand is negligible compared with the coupling due to spasticity and inherent stiffness of the muscular tendon structure. This enables the system to be decoupled, and the hand and wrist dynamics can then be identified using the experimental procedure described in [4].

\[ G \text{ feedback controller, } \]

The associated error is given by \( e(t) = [e_1(t), e_2(t), \ldots, e_n(t)]^T \). Feedback controller component \( C_{C_{w}} \) takes the form of an input-output linearising controller, in series with a linear feedback controller, \( K_{f}(s) \) that stabilises the resultant dynamics \( H(r)(s) \) to produce the closed-loop system \( \Phi_{W}(s) = K_{f}(s)K_{w}(s) = \Phi_{W}(r)(s) \). Using this control structure, voluntary movement by the participant can be treated as iteration-invariant disturbance and can be compensated for. A robust ILC scheme can deal with dynamic changes and model inaccuracies due to fatigue, spasticity and other physiological effects. The phase-lead ILC algorithm has the form

\[ v_{w,k+1} = v_{w,k} + L_{w}e_{w,k} \]

where learning operator \( L_{w} \) is designed using relationship \( \frac{v_{u,k+1}}{v_{w,k}} = G_{w}(s) \).

\[ \begin{bmatrix} \phi_{1}\left( s \right) \\ \phi_{2}\left( s \right) \\ \cdots \\ \phi_{n}\left( s \right) \end{bmatrix} = \Phi_{W}\left( s \right) \]

Figure 3: Block diagram of the ILC scheme

The control system is shown in Figure 3, where \( \Phi_{W}(s) = [\phi_{1}(s), \phi_{2}(s), \ldots, \phi_{n}(s)]^T \) contains the reference joint angles, which are partitioned as \( \Phi_{W}(s) = [\Phi_{W_{1}}(s), \Phi_{W_{2}}(s)]^T \), and \( \Phi_{W_{1}}(s) \) denotes the joint angles on trial \( k \).

The associated error is given by \( e_{w} = \Phi_{W} - \Phi_{w} \), and is partitioned as \( e_{w}(t) = [e_{w_{1}}(t), e_{w_{2}}(t)]^T \). Feedback controller component \( C_{C_{w}} \) takes the form of an input-output linearising controller, in series with a linear feedback controller, \( K_{f}(s) \) that stabilises the resultant dynamics \( H_{w}(s) \) to produce the closed-loop system \( G_{w}(s) := (I + H_{w}(s)K_{w}(s))^{-1}H_{w}(s)K_{w}(s) = \Phi_{w}(s) \) where \( H_{w}(s) = [Bw_{w}^{2} + (C_{w_{2}} + F_{w_{2}})s + (C_{w_{1}} + F_{w_{1}} + G_{w_{1}} + K_{w_{1}})]^{-1}\operatorname{diag}(h_{LAD,6}(s), h_{LAD,17}(s)) \) shown in Figure 4.

Having stabilised the wrist and hand joint dynamics, an ILC controller is required to accurately track of \( \Phi_{W} \). This takes the form

\[ v_{w,k+1} = v_{w,k} + L_{w}e_{w,k} \]

with operator \( L_{w} \) designed using relationship \( \frac{v_{u,k+1}}{v_{w,k}} = (I + H_{w}(s)K_{w}(s))^{-1}H_{w}(s) \) such that \( \lim \ e_{w,k} = 0 \) with \( \lim \ v_{w,k} = v_{w,k}^{*} \). As in the previous case a large range of designs is available [3]. In both cases, a robust ILC scheme can deal with dynamic changes and modelling inaccuracies due to fatigue, spasticity and other time-varying physiological effects.

\[ \begin{bmatrix} \phi_{1}\left( s \right) \\ \phi_{2}\left( s \right) \\ \cdots \\ \phi_{n}\left( s \right) \end{bmatrix} = \Phi_{W}\left( s \right) \]

Figure 4: Hand and wrist block diagram

For wrist and hand, linearising feedback controller \( C_{LAD} \), which produces \( C_{LAD} = K_{w}(s)(P(t)W)^T \) where

\[ P(t) = \begin{bmatrix} d_{1,2,3,4,5,6}(\mu_{1}) & \cdots & d_{1,2,3,4,5,6}(\mu_{24}) \\ \vdots & \ddots & \vdots \\ d_{1,2,3,4,5,6}(\mu_{1}) & \cdots & d_{1,2,3,4,5,6}(\mu_{24}) \end{bmatrix} \]

and the linear feedback controller \( K_{w}(s) \) is designed to stabilise the resulting linear closed-loop system

\[ G_{w}(s) := (I + H_{w}(s)K_{w}(s))^{-1}H_{w}(s)K_{w}(s) = \Phi_{w}(s) \] where \( H_{w}(s) = [Bw_{w}^{2} + (C_{w_{2}} + F_{w_{2}})s + (C_{w_{1}} + F_{w_{1}} + G_{w_{1}} + K_{w_{1}})]^{-1}\operatorname{diag}(h_{LAD,6}(s), h_{LAD,17}(s)) \) shown in Figure 4.
3 Results

Feasibility was established in preliminary tests with unimpaired participants (N=2) who provided no voluntary effort. These confirmed high levels of performance over a range of functional tasks. For the case of stroke participants, results showed that FES-assisted performance increased over the course of the intervention for a range of functional tasks. As seen from Table 1, statistically significant improvements were also observed in FES-unassisted tasks over the course of the intervention. In particular, range of movement (ROM) increased at the shoulder, elbow, wrist and index finger joints over a range of tasks; the high light switch demonstrated the most significant gain in shoulder flexion ROM, the contralateral reach in elbow extension ROM, the near reach in wrist extension ROM and the far reach in index finger extension ROM.

Table 1: Stroke participant clinical assessment data

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<th>FES-Unassisted Range of Movement</th>
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Figure 5: Participant 1’s tracking results for light switch task

4 Conclusions

Results confirm that FES, mediated by ILC, successfully assisted participants in completion of functional tasks, and training transferred to tangible changes in motor performance. The key findings were significant improvements in FES-unassisted performance with different metrics. In addition, participants reported that the system was usable, enjoyable and motivating, and importantly that the intervention was effective in reducing weakness, leading to changes in everyday activities at home. Finally, the feasibility of using low-cost, user-friendly sensing approaches and arm support mechanisms that can be used in conjunction with FES-assisted tasks were established and provide an important step towards the transference of such a rehabilitation system to a home-based system.

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References


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