Medical Engineering and Physics 38 (2016) 1185–1194

Contents lists available at ScienceDirect

Medical Engineering and Physics

journal homepage: www.elsevier.com/locate/medengphy

Feedback control of electrical stimulation electrode arrays

C.T. Freeman*, K. Yang, J. Tudor, M. Kutlu

Electronics and Computer Science, Faculty of Physical Science and Engineering, University of Southampton, Southampton SO17 1BJ, UK

ARTICLE INFO

Article history:
Received 22 October 2015
Revised 3 June 2016
Accepted 4 July 2016

Keywords:
Electrode arrays
Electrical stimulation
Feedback control
Assistive technology

ABSTRACT

Electrical stimulation electrode arrays are an emerging technology that enables muscles to be artificially contracted through the activation of their associated motor neurons. A principal application of electrical stimulation is to assist human motion for orthotic or therapeutic purposes. This paper develops a framework for the design of model-based electrode array feedback controllers that balance joint angle tracking performance with the degree of disturbance and modeling mismatch that can exist in the true underlying biomechanical system. This framework is used to develop a simplified control design procedure that is suitable for application in a clinical setting. Experimental results evaluate the feasibility of the control design approach through tests on ten participants using both fabric and polycarbonate electrode arrays.

1. Introduction

There is a pressing need for novel technologies to support recovery of arm function following neurological conditions such as stroke and multiple sclerosis. Electrical stimulation (ES) uses electric impulses to artificially activate nerve cells causing muscle contraction, and has become an area of intense engineering and clinical research over the last few years [1–3]. By directly activating weak or paralyzed muscles, ES is able to drive neuroplastic cortical changes to enable recovery. ES is supported by a growing body of clinical evidence [4–6], and is increasingly combined with mechanical support, taking the form of either passive orthoses or active robots. These devices help support the affected limb using various training modalities, and therefore help reduce muscle fatigue or provide functionality that ES cannot (e.g. to assist with forearm supination or help stabilize the scapula).

The recent emergence of transcutaneous electrode arrays has potential to improve selectivity, automate placement, and reduce fatigue and discomfort compared with single pad ES electrodes [7,8]. The freedom they embed to adjust the size and shape of the electrode means they can isolate smaller muscle groups, and thereby enable the user to perform a variety of functional tasks including walking [9,10], and hand/wrist motion [8,11].

A major aim of current ES electrode array research is to produce a flexible, breathable, and light weight device that patients can use at home to support independent living. Manufacturing processes capable of realizing this form of wearable technology have recently been demonstrated: screen printing of bespoke polymer based pastes has been successfully used to produce a flexible and breathable fabric electrode array [12], with an example shown in Fig. 1. Screen printing is a straightforward and cost effective fabrication method which facilitates significant design freedom in terms of pattern geometries [13,14]. This technique has overcome limitations of alternative fabrication techniques: embroidery requires expensive high quality custom made silver sputtered yarns [15], and weaving and knitting constrains the array design layout to follow the physical location of the yarns [16–18] and has a lack of homogeneity in electrical properties.

However, lack of precise, clinically feasible methods with which to control the ES applied to the large number of electrode array elements remains a substantial challenge. Existing control strategies are open-loop and use time-consuming element selection procedures, which limits accuracy and usability. For example, in the report by Heller et al. [9], array elements are stimulated sequentially to locate the best single site for drop foot, obtaining similar performance to that produced manually by a clinician. Each array element is also tested in turn in Schill et al. [19], using simple criteria to assess the quality of wrist stabilization, with tests performed on tetraplegic spinal cord injury patients. Other implementations such as Keller et al. [20] also operate in a similar way to a clinician manually repositioning a single electrode. In the work by Popović and Popović [21] array electrodes are selected to minimize a cost function based on joint angle data produced during individual activation, and in the work by Malesěvić et al. [22] the same form of data is used to train an artificial neural network. There is therefore a clear need for model-based feedback control

* Corresponding author. Fax: +44 2380593709.
E-mail addresses: cf@ecs.soton.ac.uk (C.T. Freeman), ky2@ecs.soton.ac.uk (K. Yang), mjt@ecs.soton.ac.uk (J. Tudor), mck1e12@ecs.soton.ac.uk (M. Kutlu).

http://dx.doi.org/10.1016/j.medengphy.2016.07.002
1350-4533/

Fig. 1. Screen printed fabric electrode array.

designs to improve accuracy, ideally embedding mechanisms to reduce the model identification time through selection of a reduced input search space. Control design in such a framework implicitly rests on a compromise between tracking accuracy of the nominal system and its robustness to model uncertainty. In order to reach this trade-off in a systematic manner, it is vital to employ a principled design procedure based on underlying theoretical performance and robustness results. To address this problem, the goal of this paper is two-fold:

1. We develop a comprehensive framework in which to design controllers to assist motion using ES electrode arrays. For the first time this establishes precise bounds on the level of modeling error that can be tolerated (e.g. due to muscle fatigue) and facilitates design of controllers that transparently balance tracking performance with robustness to such uncertainty and simplifications that enable clinically feasible identification methods to be employed.
2. We apply this framework to experimentally evaluate the performance of fabric electrode arrays with ten participants, and in particular compare achievable tracking accuracy with that of the leading alternative (arrays printed on polycarbonate with a hydrogel layer).

This paper exploits general robustness analysis developed by Freeman [23, Chapter 8] for ES control of the upper limb, but specifies them to array based linear feedback control. The significant simplification this unlocks enables more transparent results to be developed, which in turn lead to new control design procedures. This paper also contains far broader evaluation results, as well as a comparative study between two types of electrodes. The contents are organised as follows: Section 2 describes the model of the electrode array stimulated system, Section 3 develops a design framework for robust feedback controllers, and Section 4 presents a suitable model identification procedure. Experimental results are given in Section 5 and conclusions in Section 6.

2. Modeling of a single ES electrode array

Let signal $\boldsymbol{u} \in \mathcal{L}_2^n[0, T]$ contain the ES signals applied to each of the n elements of the array over time interval $t \in [0, T]$. The stimulation which then causes contraction of the ith muscle can be assumed to be a linear combination of those array elements within spatial range, and is therefore modeled by component

$$z_i(t) = \sum_{j=1}^{n} a_{i,j} u_j(t), \quad i = 1, \dots, m, \quad t \in [0, T], \tag{1}$$

within signal $\boldsymbol{z} \in \mathcal{L}_2^m[0, T]$, where $a_{i,j} \in \mathbb{R}_+$ is the contribution of the jth array element. If the ith muscle acts about a single joint with angle $\phi_k(t)$, then the Hill type model states that the resulting moment is

$$\tau_{k,i}\big(z_i(t), \phi_k(t), \dot{\phi}_k(t)\big) = h_i(z_i(t), t) \times \tilde{F}_{M,k,i}\big(\phi_k(t), \dot{\phi}_k(t)\big) \tag{2}$$

where $h_i(z_i(t),\ t)$ is a Hammerstein structure comprising static non-linearity, $h_{IRC,\ i}(z_i(t))$, representing the isometric recruitment curve, cascaded with stable linear activation dynamics, $H_{LAD,\ i}$, represented by state-space triple $\{M_{A,\ i},\ M_{B,\ i},\ M_{C,\ i}\}$. Bounded term $\tilde{F}_{M,k,i}(\cdot)$ captures the effect of joint angle and angular velocity on the moment generated. As multiple muscles and/or tendons may each span any subset of joints, the general expression for the total moment generated about the kth joint can be represented by

$$\tau_k\big(z(t), \boldsymbol{\phi}(t), \dot{\boldsymbol{\phi}}(t)\big) = \sum_{i=1}^{m} \Big\{ d_{k,i}(\phi_k) \times \tau_{k,i}\big(z_i(t), \phi_k(t), \dot{\phi}_k(t)\big)\Big\},$$
$$k = 1, \dots p. \tag{3}$$

Here $d_{k,i}(\phi_k) = \frac{\partial E_i(\phi_k)}{\partial \phi_k}$ is the moment arm of the ith muscle with respect to the kth joint, where continuous function E is the associated excursion [24]. Resulting moment $\boldsymbol{\tau} \in \mathcal{L}_2^p[0, T]$ actuates the joints of the inter-connected anthropomorphic and mechanical/robotic support structure, with associated joint angle signal $\boldsymbol{\phi} \in \mathcal{L}_2^p[0, T]$. As demonstrated by Freeman [23, Chapter 2], this structure can be represented by the rigid body dynamic system

$$\boldsymbol{B}(\boldsymbol{\phi}(t))\ddot{\boldsymbol{\phi}}(t) + \boldsymbol{C}(\boldsymbol{\phi}(t), \dot{\boldsymbol{\phi}}(t))\dot{\boldsymbol{\phi}}(t) + \boldsymbol{F}(\boldsymbol{\phi}(t), \dot{\boldsymbol{\phi}}(t)) + \boldsymbol{G}(\boldsymbol{\phi}(t))$$
$$+ \boldsymbol{K}(\boldsymbol{\phi}(t)) = \boldsymbol{\tau}(\boldsymbol{z}(t), \boldsymbol{\phi}(t), \dot{\boldsymbol{\phi}}(t)) \tag{4}$$

where $\boldsymbol{B}(\boldsymbol{\phi}(t))$ and $\boldsymbol{C}(\boldsymbol{\phi}(t), \dot{\boldsymbol{\phi}}(t))$ are respectively the $p \times p$ inertial and Coriolis matrices of the amalgamated anthropomorphic and mechanical/robotic support structure, and $\boldsymbol{G}(\boldsymbol{\phi}(t))$ is the $p \times 1$ combined gravity vector. The $p \times 1$ term $\boldsymbol{K}(\boldsymbol{\phi}(t))$ is the assistive moment produced by the mechanical passive/robotic support (see [23, Chapter 2] for explicit forms in both exoskeletal and end-effector cases). Finally, $\boldsymbol{F}(\boldsymbol{\phi}(t), \dot{\boldsymbol{\phi}}(t))$ is the $p \times 1$ vector representing joint stiffness, damping and friction effects, which for simplicity will be assumed to take the form

$$\boldsymbol{F}(\boldsymbol{\phi}(t), \dot{\boldsymbol{\phi}}(t)) = [F_{e,1}(\phi_1(t)) + F_{v,1}(\dot{\phi}_1(t)), \dots, F_{e,p}(\phi_p(t))$$
$$+ F_{v,p}(\dot{\phi}_p(t))]^\top. \tag{5}$$

Expansions in the form (5) can be made to incorporate more complex phenomena, e.g. those involving coupled position and velocity, or the addition of a varying set-point [25,26].

2.1. Operator description

The relationship between ES and joint angle defined by (1)–(5) can be expressed equivalently as

$$M : \mathcal{L}_2^n[0, T] \to \mathcal{L}_2^p[0, T] : \boldsymbol{u} \mapsto \boldsymbol{\phi} : \boldsymbol{\phi} = H_{RB} F_m(\boldsymbol{\phi}, \dot{\boldsymbol{\phi}}) H_{LAD} h_{IRC}(A\boldsymbol{u}), \tag{6}$$

with elements defined by the operators

$$A : \mathcal{L}_2^n[0, T] \to \mathcal{L}_2^m[0, T] : \boldsymbol{u} \mapsto \boldsymbol{z} : \boldsymbol{z}(t) = \begin{bmatrix} a_{1,1} & \cdots & a_{1,n} \\ \vdots & \ddots & \vdots \\ a_{m,1} & \cdots & a_{m,n} \end{bmatrix} \boldsymbol{u}(t),$$

$$\boldsymbol{h}_{IRC} : \mathcal{L}_2^m[0, T] \to \mathcal{L}_2^m[0, T] : \boldsymbol{z} \mapsto \boldsymbol{v} : \boldsymbol{v}(t)$$
$$= [h_{IRC,1}(\boldsymbol{z}_1(t)), \dots, h_{IRC,m}(\boldsymbol{z}_m(t))]^\top, \tag{7}$$