Modeling Open-Loop Stability of a Human Arm Driven by a Functional Electrical Stimulation Neuroprosthesis

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***Abstract*— Functional electrical stimulation (FES) can be used to restore movement control following paralysis. For complex multijoint systems, it is becoming increasingly apparent that closed-loop controllers are needed. Designing a closed-loop control system is easiest when the open-loop system is stable. In this study we developed a computational model to assess the open-loop stability of FES-control systems. We used the model to examine the open-loop stability of the human arm throughout its reachable workspace. For each simulated position of the hand we examined the stability of the arm, assuming that a minimal pattern of muscle activation was used to support the arm against gravity. Only muscles available to an existing FES user were considered. We found that with this reduced muscle set, the stability of the arm was severely compromised. We also demonstrated that muscle co-contraction can be an effective method to improve the stability for many postures.**

I. INTRODUCTION

Functional Electrical Stimulation (FES) is a promising technology for activating muscles and restoring lost functions to patients with spinal cord injuries (SCI). The long-term goal of this study is to develop an FES-control strategy to restore reaching ability to people with paralyzed arms. For complex tasks requiring multiple muscles and joints, feedback control is likely to be necessary, but there are many challenges associated with implementing an FES feedback controller. These include the low stimulation rates typical for FES systems and the delays that can lead to feedback instabilities [1]. In control theory it is well known that stabilizing an unstable system generally requires high-frequency feedback, which is yet to be feasible using FES. Hence, the task of implementing an FES feedback controller can be simplified by understanding the open-loop stability of the arm and preferably staying within the operating regime where the arm is open-loop stable.

The stability of the arm at a fixed posture can be assessed by characterizing its mechanical impedance [2]. Stiffness, the static component of impedance, is most relevant to postural tasks. It quantifies the static restoring forces or torques in response to externally imposed displacements. The arm is open-loop stable if its stiffness results in forces and torques that oppose external displacements, as its viscosity will help damp out oscillations and restore to its original posture [3].

Musculoskeletal models are useful for exploring the feasibility of FES control systems. It was shown that musculoskeletal models of the upper extremity can be adapted to simulate the conditions of individuals with SCI at various levels, and the performance of different designs of neuroprosthesis in restoring arm functions were evaluated to guide the selection of appropriate designs for human subjects [4][5]. While these studies have been useful, they have not assessed the open-loop stability of the arm. Recently it was shown that the stiffness of an intact limb can be predicted by a musculoskeletal model incorporating the short-range stiffness (SRS) property of muscles [6][7], but such an approach has not yet been used to assess the stability of FES controllers.

The purpose of this study was to examine the open-loop stability of an FES controller for restoring reach. The investigation was accomplished by adapting an existing musculoskeletal model to reflect a current FES user. This adaptation involved reducing the number of muscles that could be activated, and also restricting the force that those muscles could generate. Importantly, the simulations incorporated muscle models that could predict stiffness over the full range of muscle forces. This resulting model was used to simulate postures throughout the reachable workspace assuming that a minimal set of muscle activations could be used to support the arm against gravity. These results were compared to those from a simulated unimpaired subject for which all muscles could be activated to their full capacity. Our primary hypothesis was that the selected set of minimal muscle activations would be sufficient to stabilize the simulated unimpaired arm throughout the workspace, but insufficient to guarantee stability for the simulated FES arm. Such a result would imply that more complex control is required for FES reaching as well as help identify the regions of the workspace where that control is most necessary.

II. METHODS

A. 3-D musculoskeletal model

The musculoskeletal model of the upper extremity developed by Holzbaur et al. [8], and now implemented in the OpenSim environment [9], was used in this study. The model incorporates kinematic representations for the shoulder and elbow joints, and includes 37 muscle segments. Our simulations considered five degrees-of-freedom (DOF): three at the shoulder, and two at the elbow. Among the 37 muscle segments, 15 segments were chosen to represent

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the muscles that can be activated by an implanted FES neuroprosthesis [10] (Subject 1). In addition, we scaled the maximum isometric force of the FES-activated muscles to 50% of their nominal values that appear in the original arm model [8], to capture the reduced force generating capability of muscles artificially activated using FES [5]. The inertial parameters were taken from Winter [11].

B. Solving the inverse kinematics

The endpoint of the arm was defined to be along the axis of pronation/supination at a distance from the elbow corresponding to the location of the knuckles. This corresponded approximately to the 5^{th} *metacarpophalangeal joint*. The endpoint was described in Cartesian coordinates (x, y, and z), whereas there were five DOFs for the joint space coordinate system. The mismatch in dimensions creates a 2-D null space in which the joint angles can vary arbitrarily while the same endpoint position is achieved. To reduce the dimension of the problem we posed one more constraint such that the palmar surface of the hand is always vertical. This resulted in a 1-D null space that could be parameterized by the angle of shoulder elevation. For a specified endpoint location, the posture was selected to result in the minimal amount of gravitational potential energy along this null space. Similar methodologies have been reported by De Sapio et al. [12].

C. Computing the muscle forces

An optimization algorithm was used to estimate the muscle forces required to support the arm against gravity. The torques that the active muscles needed to generate, τ_{muscle}, were defined as

$$\tau_{\text{muscle}} = G(q) \tag{1}$$

where q is the vector of joint angles, and G is the vector of gravitational torques. τ_{muscle} is the product of muscle moment arms and active muscle-fiber forces

$$\tau_{\text{muscle}} = R^{\mathrm{T}} F^{\text{m}}_{\text{active}} \tag{2}$$

where R is a 37×5 matrix of muscle moment arms, and $F^{\text{m}}_{\text{active}}$ is the vector of active muscle-fiber force. The active force for each muscle was limited to the maximum achievable muscle forces, F^{0}_{active}, at the simulated length

$$F^{\text{m}}_{\text{active}} = \alpha F^{0}_{\text{active}} \tag{3}$$

where α is the vector of muscle activations $\in [0, 1]$.

Because our simulations included more muscles than the number of joints, optimization was used to resolve the redundancy. A cost function minimizing the sum of squared activations was used, as suggested by Anderson [13].

D. Modeling arm stiffness

Once the active muscle force was specified, the short-range stiffness (SRS) for each muscle was then calculated as described by [14]. This calculation assumes the SRS for the entire muscle-tendon unit can be described by the series connection of an elastic tendon with stiffness K_{t} in series with a muscle having a force-dependent stiffness K_{m}. The net stiffness for each muscle-tendon unit is then given by

$$K_{\text{SRS}} = \frac{K_{\text{m}} K_{\text{t}}}{(K_{\text{m}} + K_{\text{t}})}. \tag{4}$$

The stiffness of the contracting muscle is dependent on the force within that muscle, $F^{\text{m}}_{\text{active}}$, as follows

$$K_{\text{m}} = \frac{\gamma F^{\text{m}}_{\text{active}}}{l^{\text{m}}_{0}} \tag{5}$$

where l^{m}_{0} is the optimal muscle length, and γ is a dimensionless scaling constant ($\gamma = 23.4$) used for all muscles [14]. The tendon stiffness was defined by the slope of the generic, dimensionless force-strain curve [15], and then scaled for each individual muscle-tendon unit.

The muscle stiffnesses were transformed into joint coordinates, as described by [16]

$$K_{\text{J}} = R^{\mathrm{T}} K_{\text{SRS}} R + \frac{\partial R^{\mathrm{T}}}{\partial q} F^{\text{m}}_{\text{active}} - \frac{\partial G}{\partial q}. \tag{6}$$

The first term in the right hand side is the transformation of the muscle SRS from muscle level to joint level. The second term is the equivalent stiffness resulting from the change of muscle moment arms to the change of joint angles. The third term is the equivalent stiffness reflecting how the gravitational torques change with joint angles. Passive joint properties were excluded in the model.

In this study, the joint stiffness alone provides a sufficient evaluation to the stability of the arm, as the viscous properties during the maintenance of posture are dissipative [3]. The eigenvalues of the joint stiffness matrix were used to determine the stability of the arm such that:

$$q^{0} \text{ is unstable if } \exists \text{ eig}\left(K_{\text{J}}(q^{0})\right) > 0. \tag{7}$$

The constraint that the palmar surface of the hand being in the vertical plane results in the forearm center of mass being above the pronation/supination axis of the forearm. This leads to an unstable situation in the absence of muscle co-contraction. Therefore, our stability analysis considered only the projection of the joint stiffness matrix that excludes elbow supination. In practice, passive joint properties (excluded in our model) or small amounts of muscle co-contraction could be used to stabilize this DOF, as illustrated in our results for the other DOFs below.

E. Simulated experiments

We used our model to characterize the arm's mechanical stability in a region of the reachable workspace commonly used in activities of daily living. This region was a 50×50×50-cm cube with its center at the shoulder height, roughly 35 cm in front and 5 cm to the right of the shoulder. The intersection of this region and the reachable workspace was evenly sampled at 5-cm increments, resulting in 671 target locations. The MATLAB–OpenSim API [17] was used to acquire the simulated muscle parameters from the OpenSim at each target location. Fig. 1 shows a horizontal slice of the simulation volume.

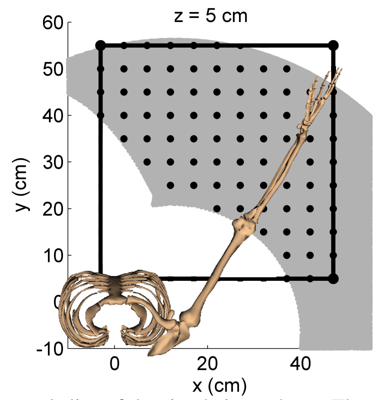

Fig. 1. A horizontal slice of the simulation volume. The coordinate system is defined as follows: the origin at the deepest point of *Incisura Jugularis*, x axis pointing lateral, y axis pointing anterior, and z axis pointing superior. The slice is taken at the plane of $z = 5$ cm. The gray region indicates the reachable workspace for our model. The black box is the defined region of interest and the black dots are the sampled endpoint locations.

Both FES and unimpaired models were simulated. Both models considered five DOFs and had identical inertial parameters. The unimpaired model included all 37 muscle segments at full strength, whereas the FES model considered only 15 segments at 50% strength, as described above. To separate the influence of the number of muscles from the strength of those muscles, we also simulated the unimpaired model at 50% strength and the FES model at full strength. Finally, we explored the use of co-contraction to stabilize the FES arm by using the original FES model (15 segments, 50% strength), but maximizing rather than minimizing the total activation at each target location.

III. Results

Our simulations indicated that the intrinsic stiffness of the muscles available to our FES subject was not sufficient to stabilize the arm in many regions of the workspace. Results are from the simulations employing a set of minimal muscle activations (Fig. 2). Muscle activations suitable for supporting the arm against gravity could be found for 516 of the 671 (76.9%) target locations. Only 23.4% of the target locations were found to be stable.

In contrast to the simulated FES subject, a suitable set of muscle activations could be found for all 671 target locations for the unimpaired simulations (Fig. 3). 81.4% of the target locations were stable using a minimal set of muscle activations. When the strength of the unimpaired simulations was reduced to 50%, 99.2% of the target locations still could be reached, and 80.6% of the target locations were stable. This suggests that the difference in the FES and unimpaired simulations did not result from the change in muscle strength. This conclusion was supported by our simulation in which the FES muscle set had a strength identical to that of the same muscles in the unimpaired simulations. In that case, a suitable set of muscle activations could be found for 77.5% of the target locations, and only 23.7% of the target locations were stable.

Co-contraction is known to be an effective method for stabilizing a limb [18], and that was found to be the case for our FES simulations. When a maximal set of muscle activations was chosen, thereby allowing for co-contraction, the percentage of stable arm postures increased from 23.4% to 61.6%. These results suggest that some degree of co-contraction may greatly improve the open-loop stability of an FES limb, though at the obvious expense of increased fatigue.

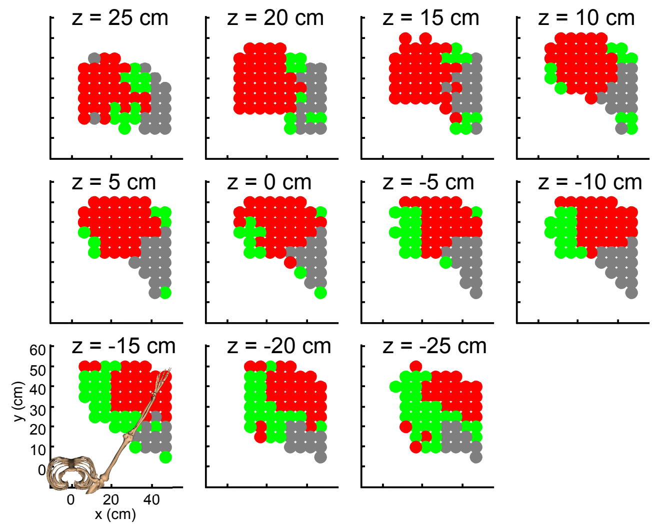

Fig. 2. Stability of the simulated FES arm for various endpoint locations. Green dots correspond to stable postures, red dots to unstable postures, and grey dots to postures for which no feasible muscle activations were found.

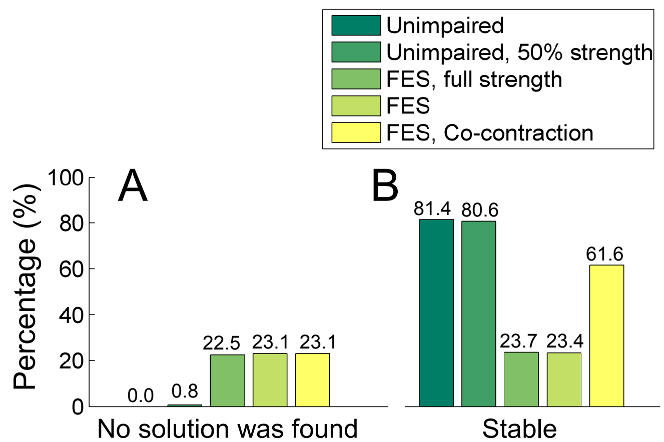

Fig. 3. Summary results for all simulated models. *(A)* The percentage of postures for which a set of muscle activations suitable for opposing gravity could not be found. *(B)* The percentage of tested postures that were stable.

IV. Discussion and Conclusions

The goal of this study was to evaluate the static open-loop stability of an arm controlled by FES for postures throughout the reachable workspace. All work was completed in simulation, using a realistic musculoskeletal model incorporating a scalable representation of muscle stiffness. The set of muscles available for FES were matched to an existing subject with implantable stimulators. Our results demonstrated that the reduced muscle set available to this particular subject severely compromises the open-loop stability of the arm. Even at postures where available set of muscles is sufficient to support the arm against gravity, stability is often not guaranteed when the gravitational forces are opposed by muscle activation patterns optimized to minimize the net activation across all muscles. In contrast, this same method for selecting muscle activations in a model of an unimpaired arm resulted in stable arm postures throughout the majority of the tested workspace. Stability of the simulated FES arm was increased dramatically by allowing for co-contraction, resulting in stable postures at more than twice as many locations within the workspace. These results suggest the need to consider both force capabilities and stability when designing FES controllers.

While our results provide novel insight to the design of multijoint FES controllers, there are a number of limitations to our study that must be considered. First, our model parameters were taken from the generic OpenSim model, and not matched to our specific FES subject other than to limit the available muscle set and the strength of those muscles. Previous model-based estimates of limb stiffness have been shown to be robust with respect to parameter variations [6][7], but we have yet to conduct sensitivity analyses for this study. Also, we considered only a portion of the reachable workspace, selected to reflect a volume relevant to typical tasks of daily living and also to avoid the extremes of joint motion. The full workspace for each subject should ultimately be considered, though an increase in the tested workspace is unlikely to affect our main conclusions.

There may be many ways to improve the open-loop stability of an FES arm. Our results demonstrated how this can be accomplished using co-contraction. Those findings are in agreement with the reported use of co-contraction to stabilize unimpaired movements [18], and the use of FES assisted co-contraction during isometric force generation [19]. A practical system would need to balance the benefits of co-contraction with the corresponding increase in fatigue, yet this may prove to be a simple strategy for expanding the regions where an FES-controlled arm is open-loop stable. An alternative approach would be to select arm postures that optimize stability, rather than those that simply minimize the required energy. This is an approach used by unimpaired subjects [20]. It is also one that has been shown to hold promise for FES [21], though that possibility has not yet been tested using the more realistic musculoskeletal models employed in this study.

In this study, the arm stiffness was simply used as an indicator to the stability. Given that the reduced muscle set available for FES reduces the regions of the workspace where the intrinsic muscle properties are sufficient to guarantee stability, it would be reasonable to consider incorporating limb stiffness or stability as part of the control objective. This would be especially helpful when controlling for interactions with the environment, since many typical activities can destabilize limb postures [16][22]. The strategy for directly controlling the stiffness of an arm has recently been introduced to the robotics community and gives rise to the design and control of variable stiffness actuators [23][24]. We are unaware, however, of similar attempts using FES. With the computational model that can provide realistic estimations of the arm's stiffness under FES control, we can make stiffness the direct control objective, and thus have the potential to restore even more functionalities using FES neuroprosthesis.

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