In Proceedings of the 7th Annual Conference of the International Functional
Electric Stimulation Society, IFESS (2002)

A Biomechanical Model of the Partially Paralyzed Human Arm

Rahman Davoodi, Ian E. Brown, Emanuel Todorov, and Gerald E. Loeb
Department of Biomedical Engineering and A.E. Mann Institute for Biomedical Engineering
University of Southern California, Los Angeles, CA 90089 USA davoodi@usc.edu

Abstract

A biomechanical model of a partially paralyzed human arm has been developed to aid in analysis of FES controllers for reaching in quadriplegia. The model represents an average adult arm and is based on data from cadaver measurements reported in the literature. Particular emphasis has been placed on the accuracy of parameters that are important for control, including the moment arms for muscles that act on multiple degrees of freedom and the range of sarcomere lengths over which each muscle operates. Sharing and upgrading the model are facilitated by the use of popular modeling and simulation environments. The model is currently being used to design FES controllers for point-to-point reaching and arm posture maintenance in quadriplegia.

1. Introduction

Biomechanical models of the extremities provide convenient and safe environment for design and evaluation of increasingly complex FES control systems such as for the restoration of reaching in quadriplegia. Here we report on the development of a computer model of the human arm that can simulate the behavior of a typical partially paralyzed arm under arbitrary muscle excitations and external forces.

Improvements in human arm modeling have paralleled the improvements in modeling software, computational power, and availability of more extensive biomechanical data. However, previously published models are missing some important features that are required for functionally realistic arm models. For example, model parameters such as the skeletal size and muscle moment arms are usually taken from measurements made from different size specimens; moment arms of the muscles (that may span multiple joints) are often correct only about a single joint and/or only for a single arm posture; and the muscles may not operate in the correct region of the sarcomere force-length curve. Further, because of the use of proprietary modeling environments, modification or upgrade of these models are often difficult.

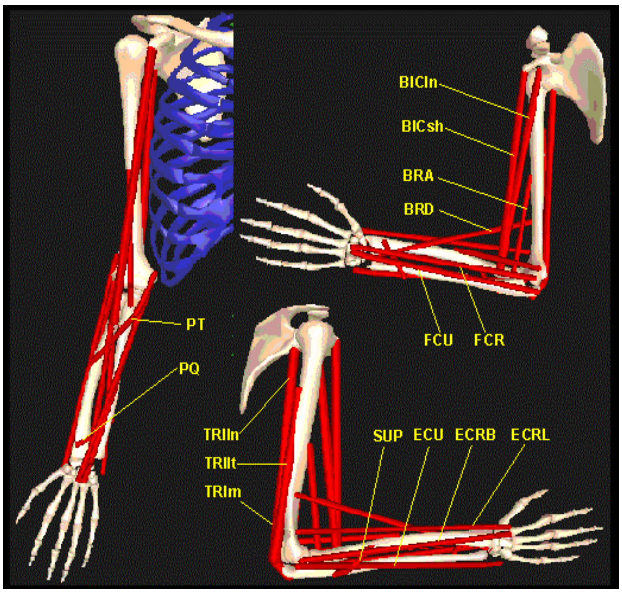

Figure 1. Musculoskeletal model of the arm in SIMM.

2. Methods

The model simulates a clinically relevant case seen often in quadriplegia in which the shoulder remains largely under voluntary control but the lost natural control of the other arm joints must be restored through FES. The model has five segments including the clavicle, humerus, ulna, radius and hand and nine rotational degrees of freedom in five joints including 2 in the sternoclavicular joint, 3 in the glenohumeral joint, 1 in the elbow joint, 1 in the forearm joint, and 2 in the wrist joint that are involved in 3D reaching tasks. To model voluntary control of the shoulder joint complex, this boundary of the modeled system is constrained to follow the shoulder joint trajectories recorded off-line or captured in real-time from subjects performing normal daily-life reaching functions. In FES applications, these voluntary shoulder movements will be used as command signals. Fifteen electrically stimulated muscles actuate elbow, forearm, and wrist joints (Fig. 1).

The sizes of the body segments reported in the literature usually represent a particular specimen with its own peculiarities that may not be representative of the average. Furthermore, no single source contains enough data to complete most musculoskeletal models including the one here. Therefore, we decided to model an average adult arm by averaging the values of the bone sizes, rotation axes, and parameters such as the elbow carrying-angle from the ten cadaver specimens reported by Murray [1]. These data are among the most complete measurements performed and include the elbow musculoskeletal geometry, moment arms and muscle architectural parameters of ten human arm specimens. The data missing from these source were scaled from other literature sources to complete the model. In the absence of a validated scaling procedure for the anthropometric data, we used a simple, common-sense scaling procedure based on the length of the overlapping skeletal segments. To complete the missing data in the more extensive data set (Murray's data), data for other joints from other sources were scaled by the size ratio of a common parameter reported in both data sets.

The use of previously published regression equations to estimate inertial parameters resulted in unrealistic negative values for some moments of inertia. Therefore we used Hanavan's geometric models [2] to estimate the inertial parameters. The simple geometric models of the body segments with uniform densities were built according to anthropometric data specifying the segment's length and circumferences at both ends. The geometric models and the experimental segment density data were used to calculate the segments mass and moments of inertia.

To complete the anatomical model, the muscles were attached to the bony landmarks by specifying their coordinates in each bone's reference frame. Adjustments to the size and configuration of cylindrical wrapping surfaces (around which a muscle is constrained to move) and slight adjustments to the muscle attachment points (within the range of the anatomically described attachment region) were used to reproduce the average experimental moment arms as a function of joint angle. For muscles spanning more than one joint, their moment arms about the primary joint, where the muscle is a prime mover, were matched first. Then, additional parameters such as the muscle insertion points were modified to match the moment arms about the remaining joints as well as possible. The average muscle moment arms about the elbow joint were taken from Murray [1]. Because the moment arm data for the forearm and wrist joints were not reported by Murray, they were taken from multiple sources (e.g. [3]). None of these sources report the size of the arm specimens that would allow scaling of the moment arm data. Therefore, the model moment arms were designed to lie in the middle of the range of data from these sources.

A realistic muscle model must produce realistic muscle force and apply it correctly to the skeletal system. The latter can be achieved by accurate muscle moment arms. For realistic muscle force production, however, one needs to make sure that the muscle fascicles operate in the correct region of the force length curve over the anatomical range of the joints that it crosses. The lengths of the muscle fibers and tendons were adjusted to match the experimental operating ranges of the muscles.

The architectural parameters of the muscles were fed to Virtual Muscle™ [4] to build muscle force production models as Simulink blocks. The musculoskeletal model was implemented in SIMM (Musculographics Inc., USA) and converted to a Simulink block by MMS [5]. The Simulink blocks modeling the musculoskeletal system and muscle force production were connected to complete the forward dynamic model of the human arm that runs in the Simulink simulation environment.

3. Results

For faithful representation of the muscles' actions on the skeleton, muscle moment arms must be correct for all the joints spanned by the muscle and for all arm configurations. As an example, modeled moment arms of the biceps muscle about elbow and forearm joints are compared with experimental data in Fig. 2. The modeled and experimentally estimated force-length curves of the muscles are superimposed on a normalized sarcomere force-length curve in Fig. 3. A good agreement shows that the muscle force production capacity at any given arm posture is modeled correctly.

4. Summary and Conclusions

We have tried to model correctly the muscle moment arms about all of the joints spanned by the muscles. The lack of adequate experimental moment arm data however is a limiting factor. Most cadaver measurements focus on the moment arms about the joint where the muscle is a primary mover. There is a clear need for additional moment arm data about all the joints spanned by each muscle and in different arm configurations. The biomechanical data collected from a single cadaver specimen should also be as complete as possible because there is no validated methodology for scaling the biomechanical data among specimens. We have tried to alleviate this problem by basing our model on the most extensive set of data and using a simple procedure to scale the missing data from other sources.

Our selection of popular and readily available modeling and simulation environments enables the model to be shared with other researchers or upgraded later to include newer sets of biomechanical data. We are currently using the model to simulate a partially paralyzed arm where the shoulder joint is forced to move according to normal reaching trajectories and the FES controller activates the paralyzed distal muscles to restore a synergistic reaching movement. We are also developing a real-time version of the model that will be controlled by the shoulder movements of a normal subject in a virtual reality environment where we can compare the actual arm movements with the simulated FES trajectories.

Acknowledgement

Funded by the A.E. Mann Institute for Biomedical Engineering.

References

[1] Murray, W. M., "The functional capacity of the elbow muscles" Ph.D. thesis, Northwestern University, 1997.

[2] Hanavan, E. P., "A mathematical model of the human body." AMRL-TR-64-102, Wright Patterson Air Force Base, Ohio, 1964.

[3] Loren, G. J., Shoemaker, S. D., Burkholder, T. J., Jacobson, M. D., Friden, J., and Lieber, R. L., "Human wrist motors: biomechanical design and application to tendon transfers," *J Biomech.*, vol. 29, no. 3, 331-342, 1996.

[4] Cheng, E. J., Brown, I. E., and Loeb, G. E., "Virtual muscle: a computational approach to understanding the effects of muscle properties on motor control," *J.Neurosci.Methods*, vol. 101, no. 2, 117-130, 2000.

[5] Davoodi, R. and Loeb, G. E. Conversion Of SIMM To SIMULINK For Faster Development Of Musculoskeletal Models. IFESS 2001, 282-284.

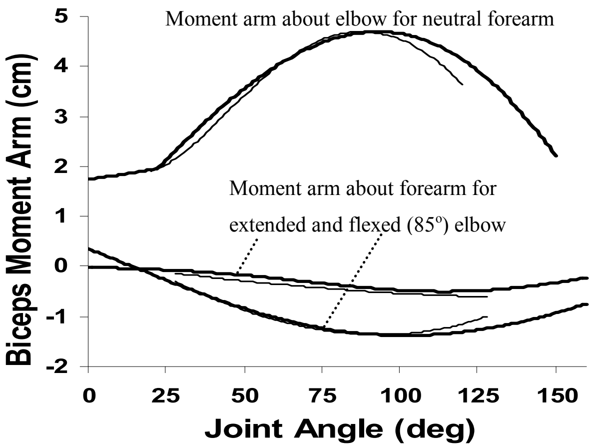

Figure 2. Model (thick line) and experimental (thin line) moment arms of Biceps muscle about the elbow and forearm joints.

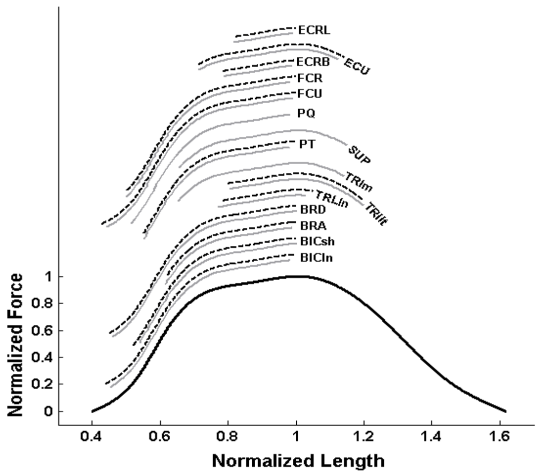

Figure 3. Experimental (dashed lines) and modeled (solid-gray lines) operating ranges of the muscle fibers superimposed on a normalized sarcomere force-length curve (solid black line).