

Development of a Muscle Spindle Model

M. Mileusnic, G.E. Loeb, I.E. Brown

Alfred E. Mann Institute for Biomedical Engineering (<http://ami.usc.edu>)
Dept. of Biomedical Engineering, University of Southern California
Los Angeles, CA 90089 USA
mileusni@usc.edu, Gloeb@usc.edu, ianbrown@usc.edu

Abstract

We are constructing a physiologically realistic model of the muscle spindle to assist in the analysis of natural sensorimotor control and to design biomimetic systems for functional electrical stimulation (FES). Our model is composed of mathematical elements that correspond closely to the anatomical components of spindles. The resulting nonlinear model is reasonably accurate in predicting published records of spindle activity during a variety of ramp, triangular and sinusoidal stretches applied under various fusimotor conditions.

1. Introduction

The main objective of the study is to develop control systems that are applicable for restoration of reach and grasp tasks of paralyzed arm and hand muscles by FES. The motivation for the study is the development of modular, injectable devices (BIONs) that can be used both to stimulate paralyzed muscles and to provide artificial proprioceptive information for sensory feedback. We believe that the design of FES controllers should be informed by the sensory regulation of natural movement. Therefore, we are developing models of biological sensors, such as Golgi tendon organs, Renshaw cells and, most importantly, muscle spindles.

The muscle spindle must sense and accurately encode length and velocity over a very wide range of kinematic conditions, despite the relatively restricted dynamic range of firing rates for action potentials [7]. It does this by shifting the relative importance of and sensitivity to length and velocity by means of specialized fusimotor efferents (primarily gamma motoneurons) that the CNS controls separately from the alpha motoneurons controlling muscle force.

There are three main types of intrafusal muscle fibers within a typical muscle spindle: bag 1, bag 2 and chain fibers. The bag 1 fiber receives dynamic fusimotor control and is primarily responsible for velocity sensitivity of the spindle. The bag 2 fiber and chain fibers are innervated by the static fusimotor control and contribute mainly to length sensitivity. The spindle is innervated by two afferents, the primary (Ia) and the secondary afferent (II). The primary afferent detects motion at the equatorial region of all three intrafusal fibers (representing both length and velocity information), while the secondary afferent is located more eccentrically on only the bag 2 and chain fibers (sensitive primarily to length information).

1.1. Model Structure

The muscle spindle model consists of two intrafusal fiber models, bag 1 and combined bag 2 plus chain, reflecting their common fusimotor drive (Figure 1a).

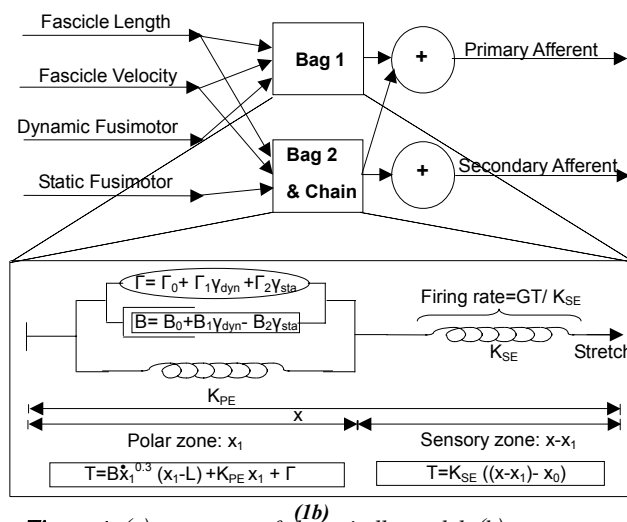


Figure 1. (a): structure of the spindle model; (b) structure of the intrafusal fiber model.

The three inputs to each intrafusal fiber model are the fascicle length, the velocity and the relevant fusimotor input. The spindle model computes two outputs similar to the biological sensors, i.e. primary and secondary afferent. Each intrafusal fiber is modeled with the same structure shown in Figure 1b, which is a modified version of the lumped linear spindle model suggested by McMahon [1].

The intrafusal fiber model is divided into a polar region and a central sensory region. The sensory region is modeled as a pure elastic element (K_{SE}), whose strain is linearly related to afferent firing rate. The tension within the sensory region is defined as:

$$T = K_{SE} ((x - x_1) - x_0) \quad (1)$$

Where x is fascicle length, x_1 polar region length and x_0 unloaded sensory region length.

The polar region is modeled as a spring (K_{PE}) with a parallel contractile element. The contractile element consists of the active force generator and the damping element. The tension within the polar region is defined as:

$$T = CBx_1^{0.3} (x_1 - L) + K_{PE}x_1 + \Gamma \quad (2)$$

The active-state force generator, Γ , is defined as the summation of a constant term (Γ_0), γ_{static} term ($\Gamma_1\gamma_{static}$) and $\gamma_{dynamic}$ term ($\Gamma_2\gamma_{dynamic}$), while damping term (B) is defined as the weighted sum of passive damping coefficient (B_0), $\gamma_{dynamic}$ term ($B_1\gamma_{dynamic}$) and γ_{static} term ($-B_2\gamma_{static}$). C is a constant describing experimentally observed effects of velocity on force, which are asymmetrical for lengthening and shortening. The model incorporates a length dependence of force production that assumes that the intrafusal fiber is operating on the ascending limb of a force-length relationship ($x_1 - L$) [9]. Importantly, the model incorporates the nonlinear velocity property (power 0.3) that has been described empirically [2].

For each intrafusal fiber model, the equations for tension within polar and sensory regions are combined into simple nonlinear first order differential equation representing net mechanical state. Afterwards, the primary afferent output is obtained by summing the outputs of bag 1 and bag 2 plus chain intrafusal fiber models, while secondary afferent output is obtained only from the bag 2 plus chain intrafusal fiber models.

2. Results

The anatomical and mathematical structure represented in Figure 1 was embodied as a set of nested blocks in the Simulink® modeling environment. The free coefficients were initially adjusted manually and in some cases optimized by using Levenberg-Marquardt method. The database included the wide range of spindle afferent activity reported in the experimental literature, which includes ramp, triangular and sinusoidal stretches applied during different fusimotor states.

The model's ability to reproduce primary afferent activity during ramp stretches is shown in Figure 2. The experimental data are from Crowe and Matthews [1], who employed three different velocities and three different fusimotor inputs (no input, $\gamma_{static}=70$ pps, $\gamma_{dynamic}=70$ pps). The responses of the same model to 8 mm triangular stretches employed Lennerstrand and Thoden's data [4,5] and are shown in Figure 3. Figure 3a deals with primary afferent firing at three different velocities, while Figure 3b is a firing record at the same stretching velocity but under different dynamic fusimotor inputs.

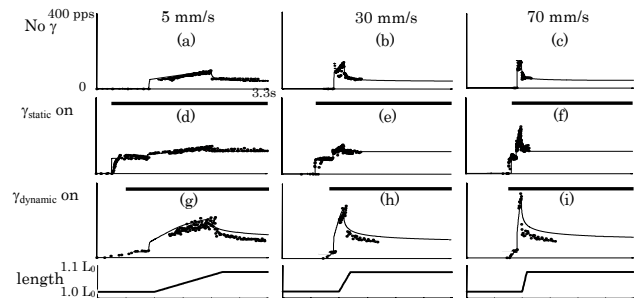


Figure 2. Model's primary afferent output vs. Crowe and Matthews data. a,d,g:5mm/sec; b,e,h:30mm/s; c,f,i:70mm/s; a,b,c:no fusimotor input; d,e,f: static fusim. input=70pps; g,h,i:dynamic fusim. input=70pps.

The final test of the model's performance is shown in Figure 4 and deals with sinusoidal stretches at different frequencies of stretch and various fusimotor drives [3]. Both during the dynamic fusimotor stimulation (Figure 4b) and without stimulation (Figure 4a), the model was reasonably accurate in predicting the primary afferent output. Since most of the current model's parameters have been manually adjusted (in order to provide insights into the function of the intrafusal components), further improvements are expected from automatic optimization of all parameters in parallel.

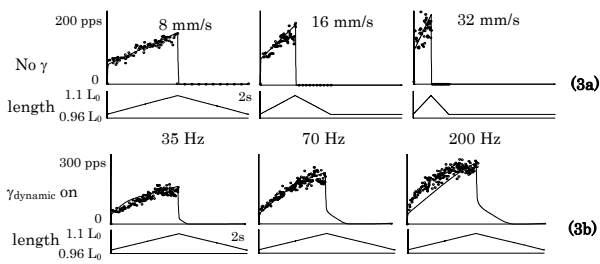


Figure 3. model's prediction of primary afferent during triangular stretch and Lennerstrand's data. 3a: three velocities of stretch without fusimotor stimulation; 3b: one velocity in presence of fusimotor input.

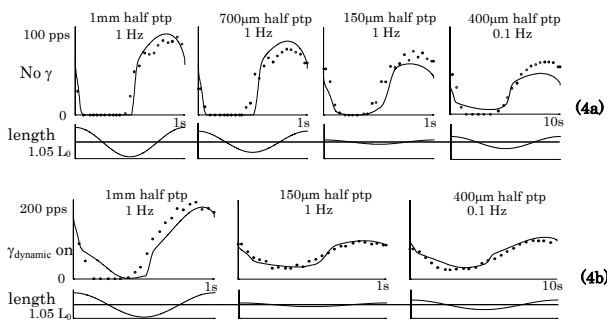


Figure 4. model's performance during sinusoidal stretches compared to data by Hulliger. 4a: no fusimotor input; 4b: dynamic fusimotor=87Hz.

3. Conclusion

The muscle spindle model is still under development and additional parameter optimization will be needed. As seen from current results, the primary afferent output is producing very good results for a wide range of kinematic and fusimotor states, and we hope the same will hold true for the secondary afferent modeling (for which less experimental data are available). The next step will be to complete models of the other biological sensors (Golgi tendon organ, Renshaw cells), as well as prosthetic sensors under development. Various combinations of these sensors will be used in models of sensorimotor regulators for control of complete musculoskeletal systems in order to understand which combinations of feedback information represent necessary and sufficient conditions for effective control.

The artificial sensors now being incorporated into the BION2 devices include a BIONic muscle spindle [6]. While one implant emits an RF signal, nearby implants will detect and quantify the strength of the RF signal, which varies with the distance and angular orientation between the emitter and detector. By using two implants located at different positions along the length of the same

muscle, the fascicle length can be sensed directly. The relative sliding motion between various muscles and a bone can be sensed by one implant in each muscle and one affixed to bone. This feedback information can then be used to infer limb posture and to develop an artificial reflexive control system similar to the biological systems that use spindle afferent feedback. Furthermore, by controlling the gain of the RF sensor, something akin to fusimotor control can be achieved in BION2 implants, whose dynamic range and resolution will be limited by considerations of data rate and resolution not unlike those encountered in neural signaling.

References

- [1] A. Crowe, P.B.C. Matthews, "The effects of stimulation of static and dynamic fusimotor fibers on the response to stretching of the primary endings of muscle spindles", *J. Physiol.*, Vol. 174, pp. 109-131, 1964.
- [2] J.C. Houk, W.Z. Rymer, P.E. Crago, "Dependence of Dynamic Response of Spindle Receptors on Muscle Length and Velocity", *J. Neurophysiol*, Vol. 46, pp. 143-166, 1981.
- [3] M. Hulliger, P.B.C. Matthews, J. Noth, "Static and dynamic fusimotor action on the response of Ia fibres to low frequency sinusoidal stretching of widely ranging amplitude", *J. Physiol.*, Vol. 267, pp. 811-838, 1977.
- [4] G. Lennerstrand, "Position and velocity sensitivity of muscle spindles in the cat. I. Primary and secondary endings deprived of fusimotor activation. *Acta. Physiol. Scand.*, Vol. 73, pp. 281-299, 1968.
- [5] G. Lennerstrand, U. Thoden, "Position and Velocity Sensitivity of Muscle Spindle in the Cat. II. Dynamic Fusimotor Single-Fibre Activation of Primary Endings", *Acta Physiol. Scand*, Vol. 74, pp. 16-29, 1968.
- [6] G.E. Loeb, R.A. Peck, W.H. Moore, K. Hood, "BION system for distributed neural prosthetic interfaces", *Medical Engineering and Physics*, Vol. 23, pp. 9-18, 2001.
- [7] G.E. Loeb, W.B. Marks, "Optimal control principles for sensory transducers", *Proceedings of the International Symposium: The Muscle Spindle*, pp. 409-415, 1985.
- [8] T.A. McMahon, *Muscles, reflexes, and locomotion*, Princeton University Press, 1984.
- [9] S.H. Scott, I.E. Brown, G.E. Loeb, "Mechanics of feline soleus: I. Effect of fascicle length and velocity on force output", *Journal of Muscle Research and Cell Motility*, Vol. 17, pp. 207-219, 1996.