1 Introduction

The contraction of a skeletal muscle involves the asynchronous activation of a population of muscle fibres. Each fibre is activated by the propagation of a wave of electrical depolarisation/repolarisation along the fibre membrane (the muscle fibre action potential). The spatial-temporal summation of these individual action potentials gives rise to a voltage measurable on the skin surface over the muscle. This voltage has become known as the electro-myogram (EMG). This electromyographic activity (EMG) provides a convenient although indirect measure of the neural input to the muscle and is used extensively in studies of neural mechanisms in motor control. The EMG will also be related to the active muscle force; if this relationship were known, it could be used to provide quantitative estimates of the forces arising from neural activity. This information would be of great value in studies of normal and pathological motor control, biomechanics, rehabilitation medicine and sports science.

There have been numerous attempts to characterise the EMG/force relationship. Most often the signal processing scheme shown in Fig. 1 has been used. The EMG signal is detected with bipolar electrodes, differentially amplified, high-pass filtered to eliminate movement artefacts, passed through a static nonlinearity (usually a full wave rectifier) and finally low-pass filtered.

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Biomechanics

Dynamic relationship between EMG and torque at the human ankle: variation with contraction level and modulation

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Abstract—Stochastic system identification techniques were used to determine the dynamic relationship between the electromyogram (EMG) and torque in the ankle muscles of normal human subjects. EMG and torque were recorded while subjects were asked to modulate ankle torque by tracking a computer-generated stochastic waveform. Nonparametric impulse response functions (IRFs) relating EMG to ankle torque were computed and parameterised by determining the parameters of the second-order system which provided the best least-squares fit. Two sets of experiments were carried out. In the first, the mean level of torque was varied from 5 per cent of the maximum voluntary contraction (MVC) to 10 per cent MVC while the depth of modulation was held constant at ±5 per cent of MVC. In the second series of experiments the mean torque was held constant at 25 per cent MVC while the depth of modulation was varied from ±2.5 per cent to ±25 per cent. The major findings were: (1) A second-order, low-pass filter provided a good quasilinear model of the EMG/torque dynamics under all conditions; (2) The model parameters depended only weakly on the mean level of torque; (3) In contrast, the model parameters depended strongly on the amplitude with which the contraction was modulated; the natural frequency increased significantly with the depth of modulation.

Keywords—Activation dynamics, Dynamics, Electromyogram, EMG, EMG-force dynamics, Human ankle, System identification, Tibialis anterior, Torque, Triceps surae

Fig. 1  EMG signal processing. The signal is detected using bipolar electrodes, differentially amplified, high-pass filtered to eliminate movement artefacts, passed through a static nonlinearity (usually a full wave rectifier) and finally low-pass filtered.

static nonlinearity (usually a full wave rectifier) and low-pass filtered (Coggshall and Bekey, 1970; Crosby, 1978; Gottlieb and Agarwal, 1971; Soechting and Roberts, 1975).

The static relationship between EMG and force has been studied extensively, but no consensus has been reached as yet regarding its form. In some studies the mean value was found to vary linearly with muscle force (Milner-Brown and Stein, 1975; Stephens and Taylor, 1973); in others it followed a parabolic relationship (Zuniga and Simons, 1969; Vredenbrgt and Rau, 1973). These contrasting results are likely to arise from differences in the experimental paradigms, the signal processing techniques and the muscles studied.
The dynamics of the EMG/force relationship have been studied using system identification methods (Bawa and Stein, 1976; Cogshall and Bekey, 1970; Crochetiere et al., 1967; Crosby, 1978; Gottlieb and Agarwal, 1971; Mannard and Stein, 1973; Soechting and Roberts, 1975). Most studies agree that these dynamics may be modelled as a second-order low-pass filter. However, estimates of the natural frequency and damping parameter vary widely and little attention has been paid to how well the models predict the observed behaviour or their sensitivity to parameter variation. Moreover, the effects of the properties of the neural activation on these dynamics have not been studied systematically despite animal experiments showing that there are significant changes with the number of active motor units and their firing rate (Mannard and Stein, 1973).

This paper describes a study in which subjects were trained to modulate ankle torque randomly about a constant mean level. Stochastic system identification techniques were used to determine the dynamic relationship between the EMG and ankle torque at different mean levels and modulation amplitudes. The major findings were:

(a) A second-order low-pass filter provided a good quasi-linear model of the EMG/force dynamics under all conditions.

(b) The model parameters depended only weakly on the mean level of torque.

(c) In contrast, the model parameters depended strongly on the amplitude with which the contraction was modulated; the natural frequency increased significantly with the depth of modulation.

2 Methods

Subjects lay supine on the experimental table with the left foot attached to the pedal of a servocontrolled actuator by an individually fabricated boot (Weiss et al., 1984). The left leg was fixed rigidly by means of straps at the pelvis and the thigh, and a clamping mechanism just above the knee.

2.1 EMGs

Bipolar EMGs were recorded from tibialis anterior (TA), triceps surae, (TS) using disposable surface electrodes (Hewlett Packard 14445A) with a diameter of 10 mm and an interelectrode distance of about 40 mm. Prior to applying the electrodes the skin was shaved and cleaned with ethanol to reduce skin impedance (TAM and Webster, 1977). The TA electrodes were placed just lateral to the femur about one-third of the distance from the knee to the lateral malleolus. The TS electrodes were placed about two-thirds of the distance from the head of the fibula to the heel.

The signals from the electrodes were connected to differential preamplifiers having a gain of 100, a common mode rejection ratio of 100 dB, an input impedance of 300 MΩ and a frequency response extending from 0 to 15 kHz. This raw EMG signal was then high-pass filtered (two-pole Butterworth filter, 10 Hz cutoff), full-wave rectified, and further amplified to a level suitable for sampling.

2.2 Paradigm

The maximum voluntary contraction (MVC) in dorsiflexion and plantarflexion was determined at the start of each experiment. These MVC values were used to normalise contraction levels and modulation levels between subjects.

Subjects were provided with an oscilloscope display of a tracking command and low-pass filtered ankle torque. They were instructed to track the command signal by modulating the activity of their ankle muscles. The command signal, generated by the computer, was a 2048 point pseudorandom binary sequence (PRBS) displayed repeatedly at 100 Hz. An auditory input whose frequency was proportional to the amplitude of the tracking signal was also provided to assist the subjects in tracking. With practice, subjects were able to perform the tracking task rather well as shown in Fig. 2, which illustrates the command signal, the torque and the EMG activity for a typical trial.

Two series of experiments were carried out. In the first, trials were obtained at six mean levels of contraction varying from 5 to 30 per cent of the MVC using a constant modulation amplitude of ±5 per cent MVC. In the second series, the mean level was held constant at 25 per cent MVC and the modulation amplitude was varied from ±2.5 to ±25 per cent of MVC. For each trial the subject was asked to start tracking the command signal and sampling was started once the subject was tracking adequately. A rest period of at least 2 min was enforced between trials to prevent fatigue. In each experiment, the trials at different mean levels or modulation amplitudes were carried out in random order.

2.3 Experimental control and data acquisition

Experimental control, data acquisition, and subsequent analysis were undertaken using Nexus (Hunter and Kearney, 1984), our language for physiological signals and systems analysis on a PDP-11/73+ minicomputer operating under RSX-11M +.

Ankle torque was measured with a very stiff torque transducer (50000 N m rad⁻¹) incorporated within the actuator. Torque and EMGs were filtered with eight-pole Bessel filters (cutoff 45 Hz) to reduce aliasing, and then