EMG–FORCE DYNAMICS IN HUMAN SKELETAL MUSCLE*

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Abstract—An on-line parameter tracking algorithm, implemented on an analogue computer, is used to obtain parameter values in an assumed mathematical relation between the full-wave rectified EMG and the force produced by the human triceps muscle in an isometric task. The relation between the actual force produced by human subjects and the computed force predicted by the model is discussed.

INTRODUCTION
When a muscle contracts, a complex electrical signal, in some cases with peak amplitudes of as much as several millivolts, may be recorded with electrodes placed on the skin overlying the contracting muscle. This complex voltage has been termed the myoelectric response, myoelectric signal, or EMG. The myoelectric signal results from the summation of motor unit potentials produced in muscle tissue underlying the electrodes.

In humans, it has been shown experimentally that there exist certain quantitative relationships between the myoelectric response of a muscle and the steady-state (isometric) force generated by that muscle. In one series of experiments, LIPPOLD (1952), the 'integrated electrical activity' was shown to be proportional to the force produced by that muscle. The 'integrated electrical activity' can be related to the standard deviation of the myoelectric signal \( \sigma_M \) in the following way. The 'integrated electrical activity' can be related to the standard deviation of the myoelectric signal \( \sigma_M \) in the following way. The 'integrated electrical activity' was determined as a function of the force produced by the gastrocnemius-soleus muscle group. To calculate the 'integrated electrical activity' of a muscle, the myoelectric signal was amplified with a capacitive coupled amplifier and recorded on chart paper. The absolute value of the area under the curve was measured with a planimeter. This is equivalent to integrating the output of a full-wave linear rectifier whose input is the myoelectric signal, as in Fig. 1. For sufficiently long integration times, the mean or expected value of the output is approximated by

\[
E [\left| V_M(t) \right|] \approx \frac{1}{T} \int_0^T \left| V_M(t) \right| dt
\]

where \( V_M(t) \) is the myoelectric signal, as recorded on the surface of the skin overlying the muscle.

It is known that the mean of the output of a full-wave linear rectifier is proportional to the standard deviation of the input, for inputs with Gaussian amplitude probability densities and zero means, DAVENPORT and Root (1958). A Gaussian assumption in this case is reasonable.

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since the myoelectric signal is the summation of a large number of individual motor unit potential waveforms and the ‘central limit theorem’ of probability theory can be invoked. Hence, the ‘integrated electrical activity’ can be considered proportional to $\sigma_M$, the standard deviation (R.M.S. value) of $V_M(t)$. Thus, in the aforementioned experiments, a linear relationship between $\sigma_M$ and force has been substantiated experimentally.

Other experiments in which $\sigma_M$ for the human biceps muscle was measured electronically during the course of an experiment also show a linear relationship between $\sigma_M$ and force, DeVries (1965).

The purpose of this paper is to describe the extension of the static EMG–force relationship to the dynamic case.

**EXPERIMENTAL PROCEDURE**

The particular muscle selected for this study was the triceps muscle of normal, healthy humans. The triceps was chosen for several reasons.

1. Ease of placement of myoelectric recording electrodes.
2. Ease of measurement of force produced at the wrist.
3. The triceps is the muscle primarily responsible for forearm extension (forearm flexion, for example, involves at least three muscles, the biceps, the brachialis, and the brachioradialis).

More specifically, the long head of the triceps was used as the source of the myoelectrical signal.

The myoelectrical signal was obtained using Beckman Biopotential Electrodes placed directly over the belly of the muscle and spaced one and one-half inches apart. The electrodes consist of a pellet of sintered silver–silver chloride surrounded by a plastic housing. A small quantity of conductive jelly was placed between the silver–silver chloride pellet and the skin. The skin of the subject was prepared by swabbing it with alcohol and puncturing the epidermis with a sterile disposable hypodermic needle at the point of application of the electrode. The electrical resistance of the skin–electrode interface was measured and found to be less than 5000 $\Omega$ in all instances. The electrodes were held in place by an adhesive collar fitting around the electrode.

The myoelectric signal was amplified by a capacitive-coupled amplifier, Dana Laboratories No. 3800. A differential input to the amplifier was used with a third electrode placed on the arm, but not over the muscle, which served as a common connection. The 120 dB common mode rejection ratio of the Dana amplifier effectively eliminated 60 Hz interference. The gain of the amplifier was 1000 with a bandwidth of 1 kHz.

The output of the amplifier was processed through a linear full-wave rectifier. The output of the rectifier is denoted by $V_r(t)$.

The downward force exerted by the subject was measured by multiplying the deflection of a very stiff mechanical system with a lever arrangement which displaced a slug in a linear differential transformer. The output of the force transducer produced a d.c. voltage proportional to the force exerted by the subject. The transducer force–voltage characteristics were measured and found to be linear over the range of operation.

**PARAMETER IDENTIFICATION**

It was assumed that the relationship between the rectified myoelectric activity and force can be characterized by a system described by a linear differential equation with constant coefficients. Alternatively, the system may be characterized by its transfer function, $H(s)$, by the use of Laplace transforms.

A second-order differential equation was used to approximate the system. Initial experiments with a third-order differential equation did not yield any improved results. If we denote the modelled force by $F_h(t)$, the second-order differential equation can be written as:

$$\frac{d^2 F_h(t)}{dt^2} + a_1 \frac{dF_h(t)}{dt} + a_2 F_h(t) = KV_r(t).$$

(2)