

EMG AND MUSCLE FORCE: AN INTRODUCTION *

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This paper is intended as an introduction to those methods of processing and presentation of the electromyogram (EMG) that give information about the muscle force or work in human movement. The physiological origin of the EMG is treated briefly, followed by a discussion of EMG recording and preprocessing: electrodes, preamplifier, rectifier, smoothing filter. Special attention is given to the prevention and suppression of interferences. The relation between EMG, muscle length and muscle force is explained on the basis of a three-component Hill muscle model. With this background, some possibilities on how to obtain quantitative information about muscle force, work and energy consumption in various categories of movement are summed up and discussed. Some possible applications in human movement studies are suggested.

Introduction

Electromyography (EMG) is becoming more and more a standard tool in the study of human movement. Many researchers, nevertheless, seem to feel some reserve towards this technique. This is quite understandable; at first sight the EMG signal seems erratic, badly reproducible and bearing little direct relation to the muscle force. It is therefore most of all used as a timing signal of the muscle activation; to determine whether the muscle is 'on' or 'off', or – at most – whether the activity is 'small' or 'large'. The main purpose of this paper is to show that this approach does not do justice to the quantitative information that can be gained from the EMG. In fact there is a very close relationship

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between EMG and muscle force. That this relationship is somewhat complicated, is due to the complexities of the muscle contraction mechanics and not to deficiencies in the EMG.

This paper is intended as an introduction to those methods of processing and presentation of the EMG that give quantitative information about the mechanical output of the muscle: force and work. It does not pretend the completeness of a review; this purpose is much better served by the recent paper of Perry and Bekey (1981) and the earlier review of Bouisset (1973). However, relevant papers since 1981 are included. There are more reviews that are of interest. As general introductions to EMG the book chapter by Grieve (1975) and the book of Basmajian et al. (1975) are recommended. Related review papers are the discussion on mathematical modelling of the EMG signal by De Luca (1979) and the extensive review on the neural control of muscular activity by Freund (1983). Indispensable for anyone working in the field of EMG is the ISEK report on units, terms and standards (Winter et al. 1980).

After a brief account of the origin of the EMG, a description is given of the way in which it may be picked up, amplified and preprocessed. A few technical points are treated in some detail; they are rather trivial from an engineer's point of view and therefore seldom mentioned in papers. For this reason they may easily be overlooked by investigators from other disciplines. The next section deals with the relation between EMG and muscle force. It can be seen as a summary and a generalization of the work of van den Berg and the present author (Hof and van den Berg 1981a, b, c, d). The final section sums up and discusses some possibilities on how to obtain quantitative information from the EMG about muscle force, work and energy consumption in various categories of movement. It concludes with some suggestions for applications.

The EMG

Introduction

The EMG is the signal that can be recorded by electrodes from an active muscle. Branches of a nerve fibre each activate the motor endplate of a muscle fibre (fig. 1). This induces two depolarization waves which travel at a speed of 3–6 m/sec to either end of the muscle fibre. The tissue between and around the active fibre is electrically

conductive. Electrical signals related to the fibre depolarization can therefore be recorded by electrodes at some distance from the fibre: the EMG. The depolarization wave is also led into the interior of the fibre by very fine transverse tubuli where it releases calcium ions which initiate the mechanical contraction process. The latter process is much slower than the electrical depolarization, which lasts only 2 msec. After a single depolarization the maximum force is not reached before 20–150 msec and the gradual decline of the force, due to reabsorption of the released calcium ions, takes at least twice as long. The mechanical processes as such are not accompanied by electrical phenomena.

The course of events in a muscle contraction may to a certain extent be compared to that in an automobile engine. Here too, the electrical impulses cause the 'ignition' in each cylinder. When the electrical signals from the cylinders are recorded, it can be deduced whether the engine is active and – after more computation – what are the number of revolutions per minute. The power developed by the engine cannot be found from these ignition signals alone, however. Additional information about the mechanical load and about the properties of the engine is needed for this.

The motor unit

A skeletal muscle is composed of muscle fibres, groups of which are innervated by a single alpha-motor neuron. The motor neuron, which is

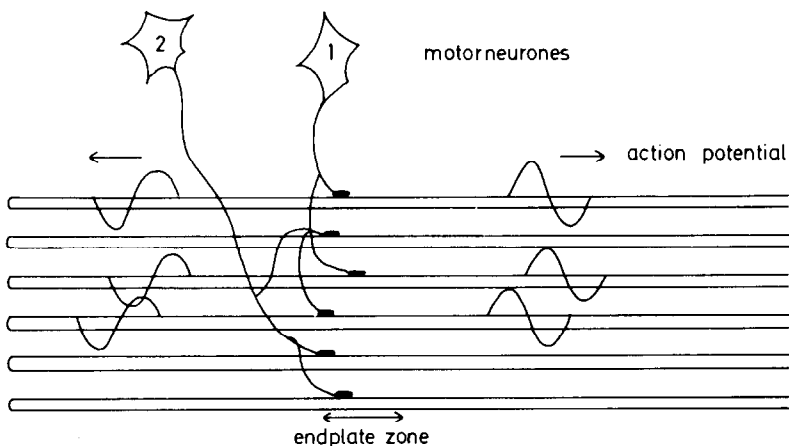


Fig. 1. Schematic diagram of two motorunits. Unit no. 1 has just fired.

situated in the spinal cord, has a long axon which goes, combined with other axons in a nerve, to the muscle. There each axon splits up into a number of branches each of which ends on a motor endplate on a muscle fibre. The ensemble of a motor neuron and the muscle fibres it innervates is called the motor unit. As the firing of a motor neuron leads to a contraction of all muscle fibres of the unit, the motor unit can be called the basic functional unit in muscular contraction.

The muscle fibres of one motor unit are scattered over a considerable area of the muscle cross section, intermingled with the fibres of some hundred other units. The motor units may have very different sizes, from a few fibres up to several thousands in large muscles. The mechanical and chemical properties are different among motor units. The small units are usually of the 'slow unfatiguable' type. The larger units can be 'slow unfatiguable', 'fast unfatiguable' or 'fast fatiguable'. The composition varies between muscles, postural muscles are predominantly slow, while muscles mainly involved in short and snappy contractions contain as a rule more fast fibres. There are differences between individuals as well, making some people apparently more fit for sprinting and others more for long-term activities.

There are two ways of controlling muscle force: changing the number of active motor units ('recruitment') and changing the firing frequency of the active units ('rate coding'). When a subject increases the force of a muscle the motor units are recruited one by one in order of magnitude ('size principle'). After recruitment the firing rate of each motor unit increases steadily with the exerted force. As a rule the units fire asynchronously. This course of events is reflected in the EMG (fig. 2). At very slight contraction (fig. 2a) only a few motor units are firing. At higher levels of activity so many units are active that at any instant the action potentials of several motor units overlap and a noise-like signal results, fig. 2b. This effect is analogous to what happens when one listens to the applause of an audience, at various levels of participation ('recruitment') and enthusiasm ('firing rate').

Electrodes

The main properties to be demanded from electrodes in quantitative EMG are a proper degree of selectivity and freedom from interferences, especially motion artefacts. The required amount of selectivity implies a compromise: the electrodes should record from the largest possible

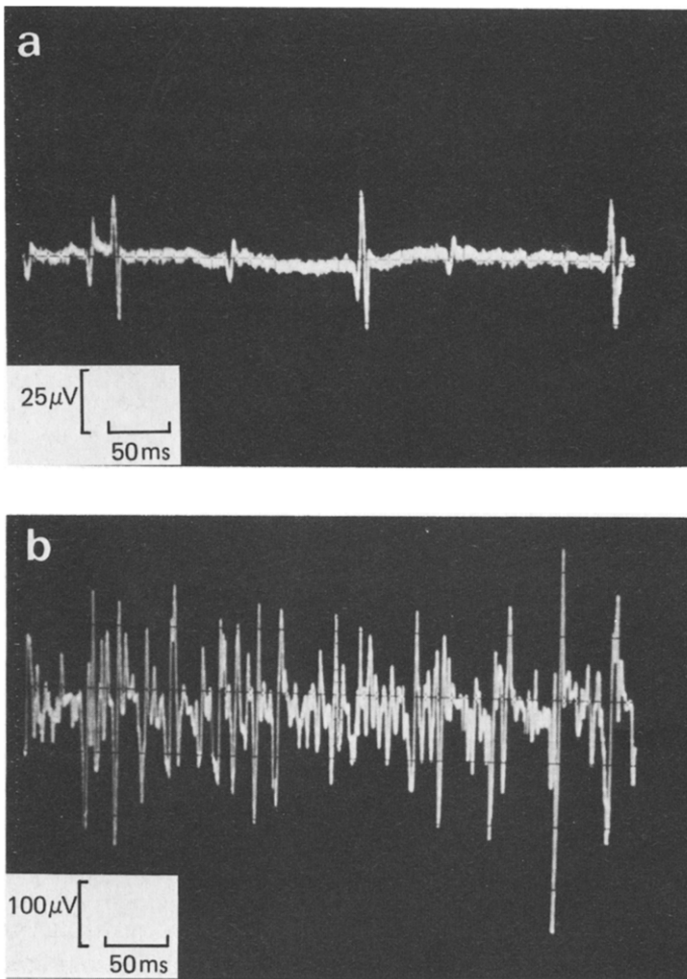


Fig. 2. EMGs of the human soleus, bipolar surface electrodes.

(a) Very slight contraction, separate motor unit potentials are visible.

(b) Strong contraction, showing an 'interference pattern'. The individual motor unit potentials are no longer discernible. Note the different voltage scales.

representative sample of the motor units of the muscle under study, but as little as possible from neighbouring muscles ('crosstalk').

The question of selectivity and the closely related problem of localization – where to put the electrodes – deserve a very careful consideration in every EMG experiment. In this context these points will not be discussed in depth. (For further information, refer to the extensive

discussion on selectivity by Grieve (1975) and to the treatise on methodological problems in muscle localization by Rozendal and Meijer (1982).)

It may seem a paradox, but crosstalk is less of a problem when the EMG is quantified than when it is only used to determine whether the activation is on or off. For instance, if it is wished to measure the EMG voltage E_A of muscle A in the presence of a crosstalk signal E_B from muscle B. Because E_A and E_B are uncorrelated stochastic signals (i.e., noise-like and from independent sources) the intensity of the recorded signal E is not just the sum of E_A and E_B , but

$$E = (E_A^2 + E_B^2)^{1/2} \quad (1)$$

As long as muscle A is inactive and $E_A = 0$, the crosstalk signal is disturbingly present: $E = E_B$, but as soon as E_A gets stronger than E_B the contribution of E_B gets lost. (In the same way as the voice of a boring uncle B gets lost when the cocktail party becomes more uproarious.) So, when E_A is twice E_B , $E = (1.25 E_A^2)^{1/2} = 1.12 E_A$. At still higher E_A , E and E_A are practically equal. The result of crosstalk is thus a 'noisy baseline', which at times may suggest a slight activity while there is none, but which does not seriously affect the quantitative interpretation at higher force levels. The possibility of crosstalk should be considered seriously, nevertheless, when choosing the electrode location; the ratio wanted/unwanted signal is what matters, not so much the signal strength itself.

For present purposes, to use the EMG for obtaining quantitative information about the muscle force, bipolar surface electrodes are as a rule the best choice. Experimental data on their selectivity and recommendations for positioning on a variety of muscles have been given by Zipp (1982a, b). The electrodes should be lightweight and should have a good skin adhesion. The metal electrode proper should lie recessed and be made of silver-silverchloride. Finally, the leads should be shielded, flexible and run parallel to the skin. For large muscles, disposable pediatric size ECG electrodes [1], which have an effective diameter (the skin area covered by electrode paste) of about 10 mm are very satisfactory. For small muscles, smaller electrodes may be preferable, e.g. the

[1] *Con. Med.* 303-3003, 933 or 934.

smallest non-disposable types, which stick with double-sided adhesive rings [2]. These electrodes fulfil the rather strict requirements regarding freedom of motion artefacts. In recordings of 'raw' EMG, namely, interference is easily noticed and can be interpreted as such, but when the EMG has to be processed in some way the high peaks of the motion artefacts are very disturbing and it is difficult to get rid of them afterwards. One of the recommended countermeasures is high-pass filtering (Halbertsma and de Boer 1981), but it remains necessary to combat the evil at the root, by a good electrode.

For muscles that are not accessible from the skin surface, for very small muscles and when regional differences within one muscle are to be studied, recording with fine-wire electrodes may be necessary. These electrodes consist of one (monopolar) or two (bipolar) very fine, isolated wires, 50 μm diameter, hooked and bare at the tips. They are inserted into the muscle with an injection needle, which is withdrawn afterwards. They are very selective, sometimes too much so, recording only from muscle fibres within some 0.3 mm from the bare ends of the wires (Gath and Stålberg 1979). This may include a sizeable number of motor units, for that matter, because of the scattered positions of the motor unit fibres. Drawbacks are motion artefacts, changes in sensitivity, wire fractures due to movements within the muscle and above all their invasiveness, which makes them less acceptable to subjects. Checking their location in non-superficial muscles, moreover, is very difficult (Rozendal and Meijer 1982). Needle electrodes, finally, have all the drawbacks of wire electrodes to an even higher degree, which makes them plainly unsuitable for EMG that is to be processed.

The amplifier

The elements of the EMG preprocessing are given in fig. 3 and their required characteristics in table 1. It should be noted that these specifications are guidelines for the types of EMG processing as discussed in this paper. A practical elaboration has been given by Halbertsma and de Boer (1981). Other methods – spectral analysis, single fibre recordings – and special circumstances – high interference, telemetry – will make other demands. Not really a link in the preprocessing chain, but very practical, is monitoring the preamplified EMG by means of a

[2] e.g. Beckmann Mini.

Table 1
EMG preprocessing.

Component	Property	Minimum specification	Remarks
Electrodes	Impedance	< 30 kΩ at 50 Hz	Reduce by skin cleaning, abrasion
Preamplifier	Input impedance	> 10 MΩ at 50 Hz	Electrode cables up to 2 m
		> 100 MΩ at 50 Hz	Very short cables (< 20 cm)
	CMRR	> 10.000 (= 80 dB)	
Band-pass filter	Noise	< 5μV _{eff}	100 kΩ source resistance
	Lower limiting frequency	30 Hz	Reduces electrode artifacts
	Upper limiting frequency	600 Hz Up to 10 kHz	Reduces noise, surface electrodes Wire electrodes
Rectifier	Full-wave, range	1:1000	
Low-pass filter	Time constant	10–25 msec	Dynamic, fast/slow muscles
		Up to 1 sec	Static
(Whole unit)	Total gain	0–20.000	Continuously variable

loudspeaker. Movement artefacts and other sources of interference are easily recognized when listening.

The specifications of the input preamplifier deserve some comment. A very important requirement for the first amplifier stage is a sufficient suppression of mains interference ('hum'), which manifests itself as a 50/60 Hz common mode voltage on both input electrodes. In a preamplifier with a high Common Mode Rejection Ratio (CMRR) such a common mode signal is highly suppressed compared to the differential EMG signal. The overall CMRR, however, is degraded when both electrode impedances Z_1 and Z_2 are not perfectly equal at 50/60 Hz, due to a voltage division with the preamplifier input impedance Z_i (Huhta and Webster 1973). This degradation amounts to

$$\text{CMRR (inbalance)} = Z_i / (Z_1 - Z_2) \tag{2}$$

Differences $Z_1 - Z_2$ of 5 – 10 kΩ can hardly be avoided with surface electrodes. In order to achieve a CMRR over 10.000 a very high input

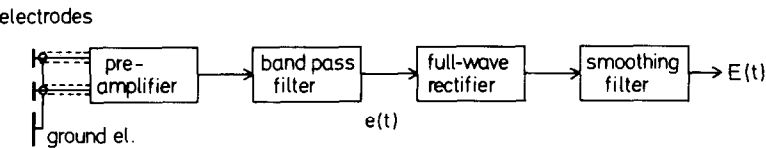


Fig. 3. Block diagram of the EMG pre-processing. See text.

impedance (more than 100 M Ω) is therefore desirable, but this makes only sense if the preamplifier can be placed within a few cm of the electrodes. A shielded electrode cable of 1 metre, namely, has a capacitance of about 100 pF and therefore an impedance of only 30 M Ω at 50/60 Hz. In such a case it is of little use to have an input impedance over 10 M Ω , and one has to be satisfied with an effective CMRR of about 1000. This story may explain why a bad electrode contact manifests itself first of all by an increased hum and why long electrode cables can introduce hum, even if they are perfectly shielded.

Rectifying and smoothing

The third and fourth stages of the preprocessing scheme as proposed in fig. 3 are full-wave rectifying and low-pass filtering ('smoothing'). Many investigators use this approach, but not all (see Perry and Bekey (1981) and the ISEK report (Winter et al. 1980)). A justification, based on a description of the EMG signal, (cf. fig. 2) is therefore needed.

$$e(t) = I(t) \cdot \underline{n}(t) \quad (3)$$

Here $\underline{n}(t)$ is a stationary stochastic process with zero mean and unit variance. In more common terms this means that $\underline{n}(t)$ is a noise signal with unit intensity and with the characteristics of the EMG signal at constant contraction. $I(t)$ is, in this formulation, the time-varying EMG intensity (in mV r.m.s.). The validity of the representation (3) is based on the fact that the characteristics of $\underline{n}(t)$, like the normalized amplitude distribution or the power spectrum, are not (or hardly) dependent on the EMG intensity $I(t)$ or on the magnitude of the muscle force (Shweddyk et al. 1977; Hogan and Mann 1980b). Because of this and because of the experimental evidence to be discussed below, it will be assumed in what follows that the intensity $I(t)$ of the EMG contains all necessary information to determine the muscle force. The only reservation to be made is that no fatigue occurs.

The restriction to the intensity of the EMG only is a consequence of the subject matter of this paper: the relation between EMG and force. It does not imply that the EMG power spectrum and related variables, like correlation functions and zero-crossing counts, do not contain interesting information. Only the extraction of the average motor unit firing frequency (Blinowska et al. 1979, van Boxtel and Schomaker

1983) and the effects of fatigue, which deserve a review of their own, are here mentioned. All that can be done on the latter subject is to give some openings to the vast literature: Bigland-Ritchie et al. 1980; Porter and Whelan 1981; and Petrofsky et al. 1982.

The simplest way to obtain an estimate of $I(t)$ is to rectify the EMG and to smooth it subsequently by means of a low-pass filter. $E(t)$ is then obtained, which is the smoothed version of $I(t)$. Some smoothing is necessary because of the random character of $e(t)$ which means that $E(t)$ contains random fluctuations as well. Expressed as a fraction ϵ of $E(t)$ itself this amounts to (Kadefors 1973):

$$\epsilon = 1/(2BT)^{1/2} \quad (\text{r.m.s}) \quad (4)$$

where T is the effective averaging time of the smoothing filter and B is the statistically equivalent bandwidth of the unprocessed EMG, which is usually between 100 and 300 Hz for surface electrodes.

It can be seen that the smoothing filter must satisfy two conflicting requirements: it should smooth $I(t)$ as little as possible and, on the other hand, it should have a long effective averaging time T , to minimize the random fluctuations in $E(t)$. For a given T a so-called 'averaging' filter with a time constant τ_1 just equal to T gives the fastest response (Kreifeldt 1971; Garland et al. 1972), but there may be valid reasons to choose some other type of filter. The simple first-order RC filter that is often used, is far from optimal, however.

For studies of EMG in fast movements the smoothing time constant should be short and the random error must be accepted. For the human calf muscles, which have twitch contraction times around 100 msec, an averaging filter with $\tau_1 = 25$ msec is used giving rise to (with $B = 280$ Hz) an ϵ of 27%. When the contractions only change slowly in strength, as in many isometric experiments, the filter time constant can be taken much longer, up to one second (with ϵ around 5%).

A reduction in the random fluctuations below the limit of eq. (4) seems possible but requires considerable effort: B can be enlarged somewhat by intricate filtering techniques or the signals of several pairs of electrodes can be pooled (Hogan and Mann 1980a, b; Harba and Lynn 1981). When the EMG patterns are repeatable (but how to determine that?) the rectified EMGs can be averaged over some hundred contractions to obtain reasonably low ϵ values, with hardly any pre-smoothing (Wadman et al. 1980; Erkelens and Bosman 1983).

A remark should be made about terminology. The preprocessed EMG $E(t)$ is known under several names: mean rectified EMG, smoothed rectified EMG (s.r.e.), low-pass filtered rectified EMG or linear envelope EMG, all of which are permissible, provided that the filter characteristics are specified. Often the word 'rectified' is omitted in the first three terms, which is in fact not correct. The root mean square (r.m.s.) EMG is not exactly the same, it is obtained by subsequently squaring $e(t)$, smoothing and taking the square root, but as a rule it is closely proportional to the mean rectified EMG. The very current term 'integrated EMG' (IEMG) for the mean rectified EMG is confusing and incorrect. It should be reserved for the integral $\int E dt$, which is something quite different (ISEK report, Winter et al. 1980).

EMG and muscle force

Isometric quasi-static contractions

When studying the mechanism of EMG generation it is quite evident that the EMG intensity should increase with the muscle force. The orderly mechanism of motor unit recruitment and rate increase suggests, in addition, that the EMG-force relation should be a fixed one. In the simplest case, when the muscle is constant in length (isometric) and when the force only changes slowly (quasi-static), it can be shown, experimentally, that *the mean rectified EMG is linearly proportional to the muscle contractile force*:

$$F = g_1 E + \text{const.} \quad (\text{isometric, quasi-static}) \quad (5)$$

with g_1 a constant gain factor. It should be stressed that (5) is an experimental relationship, which has been confirmed very often since Bayer and Flechtenmacher (1950) and Lippold (1952), but about which there is still controversy (cf. Hof and van den Berg 1977; and Perry and Bekey 1981). A theoretical explanation of this simple relationship (5) from known data on motor unit sizes, recruitment and rate coding schemes, etc. has only recently been attempted (Ray and Guha 1983) and is surprisingly complicated. In the following, the linear relationship (5) will be assumed to be valid, although any other one-to-one relationship would not introduce fundamental problems. A discussion will follow in the section 'Applications B'.

To determine g_1 it is obviously necessary that the muscle force be measurable. This is not always easy. The best way is to use a dynamometer, fixed rigidly to the limb, which can measure the moment of the force that the muscle exerts around a joint. Often it will be sufficient to use the EMG-joint moment relation thus obtained. When the muscle force is to be calculated, the musculo-skeletal geometry should be known. When no measuring device is available – not even a set of weights – or when calculation of the absolute muscle force is not possible, the smoothed rectified EMG can be expressed as a fraction of the EMG at a short maximum voluntary contraction (MVC), i.e., a contraction a motivated subject can hold for a few seconds. It will be clear that this is not a very rigid standard. For several muscles, e.g. the small finger muscles, the MVC seems to be quite reproducible. Our own experience with leg muscles is, however, that a slight discomfort in the limb fixation can prevent the subject – quite unaware – from exerting his real maximum force. Another complication is that there can be more muscle groups in a kinematic chain, and only one group forms the weakest link. When the load is held in the hand, for example, are the elbow flexors then determining the MVC, or the wrist flexors, or the shoulder muscles? This depends very much on the mechanical set-up. It may be remarked that in dynamical contractions the EMG can be higher than the level corresponding to 100% isometric MVC.

Dynamic contractions

The remarkably simple result (5) of the preceding paragraph holds only for a muscle that remains isometric, constant in length. *In the general case* with non-isometric contractions, however, *the EMG alone is not sufficient to predict muscle force but the time course of the muscle length should be known as well*. Contractions will be considered here in the widest sense, in which the time course of the muscle length and of the activation may have an arbitrary form. In the section to follow some special cases will be covered. In what follows, mathematical formulas could not be completely avoided, because they are at times the only way to express things briefly and unambiguously. Readers not familiar with this practice may be advised to skip them on first reading and to try to follow the argument by inspecting the figures.

The basic idea of EMG-to-force-processing is (a) that it be assumed that the preprocessed EMG $E(t)$ is an adequate measure of muscle

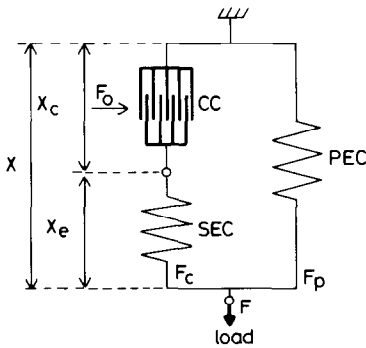


Fig. 4. Schematic diagram of the Hill muscle model. For symbols see Appendix.

activation; (b) that the muscle length is determined in some way; and (c) that the muscle force from these two signals be calculated on the basis of a physiological model of the muscle. In the author's laboratory it has been shown that this approach results in a reasonably accurate prediction of the muscle force for the human triceps surae (Hof and van den Berg 1981b, c, d). This in turn indicates that the assumption (a) is valid and that the muscle model (c) as used is suitable for the purpose. A short description of this model will be given. Details, discussion and references can be found in Hof and van den Berg (1981a).

The Hill muscle model (fig. 4) consists of three components. The parallel elastic component (PEC) represents the elasticity of the passive muscle and the ligaments. The behaviour of the active muscle is determined by the contractile component (CC) and the series elastic component (SEC), which are connected in series. These model components should not too readily be interpreted as muscle structures. The tendon and the aponeuroses seem, indeed, to share the greater part of the SEC extension, but the muscle crossbridges are, also, somewhat compliant. The lengths of the elements are expressed as normalized lengths $x = l/l_0$, in which l is the instantaneous muscle length and l_0 the length at a standardized position. Length changes of limb muscles can be calculated from joint angle changes [3]. From the arrangement

[3] In order to obtain more generally applicable formulae and figures, the calf muscle data of Hof and van den Berg (1981a-d) are in the following expressed in forces and lengths: $F = M/d$ and $x = d/l_0(\pi/2 - \phi) + 1$, with l_0 = shank length = 40 cm and d = lever arm of the calf muscles with respect to the ankle = 5 cm.

of the components in the diagram of fig. 4 it can be seen that

$$x = x_c + x_e \quad (6a)$$

$$F = F_c + F_p, \quad (6b)$$

further, that the PEC length is equal to the total muscle length x , and that the forces in CC and SEC are equal ($= F_c$). A different arrangement, with PEC parallel to CC, is also in use, but the differences with the author's model are in practice very small.

The *parallel elastic component* represents the properties of muscle, tendons and joint ligaments when the muscle is not active and is characterized by a load-extension $F_p(x)$ -diagram (fig. 5a). This relation can be modelled by an exponential function

$$F_p(x) = F_{p0}e^{p(x-1)} \quad (7)$$

or, where necessary, by a sum of more exponentials (Yoon and Mansour 1982), in which case $F_p(x)$ can become negative for a certain range of x . Usually the passive force will have a significant value only near the longest and shortest muscle lengths allowed in vivo.

Force-length relation

In an active muscle which is kept isometric, the muscle force depends on the muscle length, in fact on the length of the CC because the effect is due to a changing degree of overlap of the thick and thin filaments. Fig. 5b gives an example. This relation can be expressed as

$$F_c = f(x_c)F_0 \quad (x_c \text{ constant}) \quad (8)$$

where the function $f(x_c)$ has a value between 0 and 1. This function can be approximated by a broken line (as in fig. 5b), but also by a parabola (Woittiez et al. 1983) or by a Gaussian curve (Hatze 1981). The variable F_0 has a value equal to the isometric force at the maximum of the force-length relation. F_0 will be called the *active state* of the muscle and will be returned to later.

Force-velocity relation

As soon as the CC is no longer isometric, the muscle force is

influenced by the muscle's shortening speed $v_c = -dx_c/dt$. This is described by the Hill equation, which, in the form used here, is in fact a force-length-velocity relation:

$$F_c = F_0 \frac{f(x_c) - nv_c/b}{1 + v_c/b} \quad (9)$$

with $F_c \leq (1 + c)F_0f(x_c)$.

The latter inequality sets an upper bound to F_c at negative (i.e., lengthening) velocities. In fig. 5c the Hill eq. (9) is shown for some values of $f(x_c)$. It can be seen that the force decreases with increasing rate of shortening, below the isometric value, which is determined by the force-length relation. The rate of decrease with velocity depends on the parameter b , while parameter n determines the concave form of the curves. When the muscle is lengthened the force is some 10–20% (depending on the parameter c) higher than when isometric.

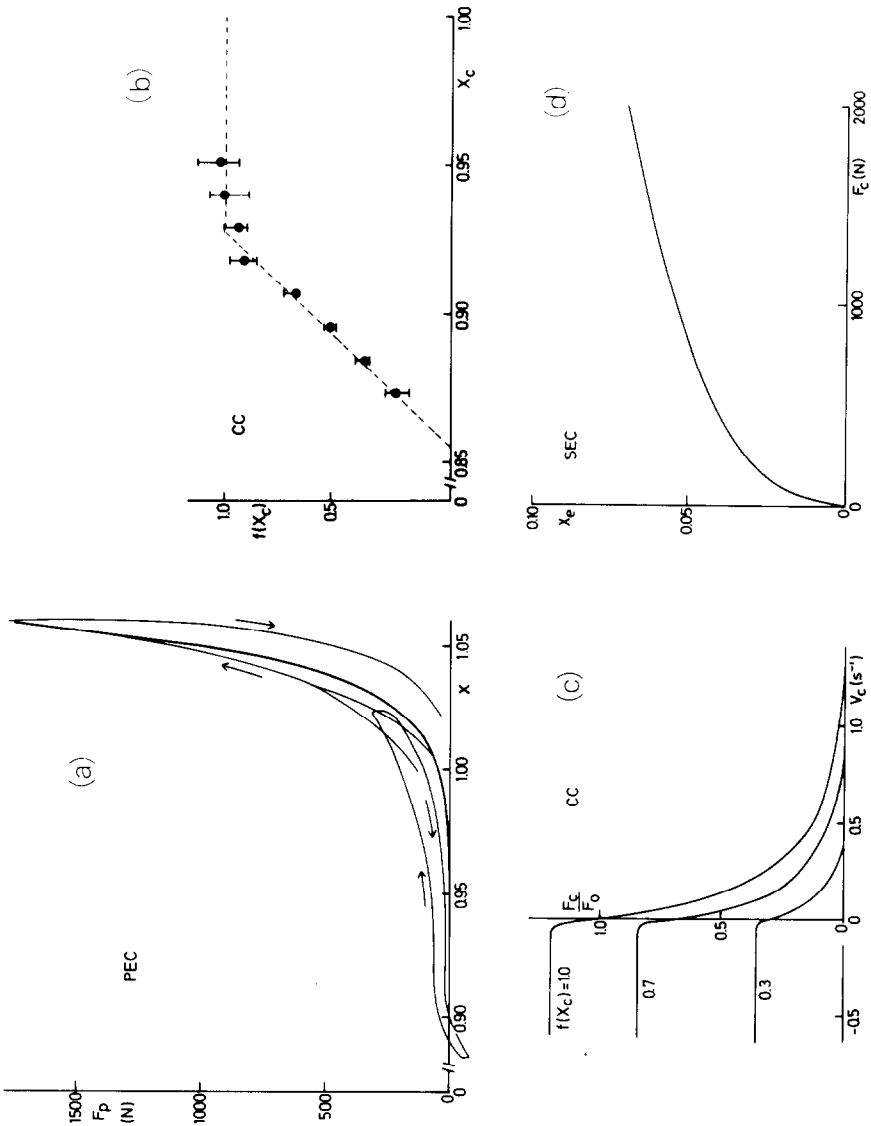
The *series elastic component* has a load-extension characteristic similar to the PEC, but now the SEC length x_e is the dependent variable, fig. 5d. In the same way as the PEC, and in fact as most soft biological tissues, the SEC is initially compliant and becomes stiffer at higher forces. A knitted stocking is an everyday example of the same phenomenon. A useful model is a combination of a logarithmic and a linear function:

$$x_e = \frac{1}{\alpha} \ln \left(\frac{F_c + F_s}{F_s} \right) + \frac{F_c}{k}, \quad (10)$$

in which α , k and F_s are parameters; this function has been drawn in fig. 5d. It should be noted that even in a so-called isometric contraction (with x constant) the CC may shorten very considerably, while stretching the SEC. E.g. at $F_c = 2000$ N, x_e amounts to 7% of the muscle length and the CC may shorten from 1.00 to 0.93, cf. figs. 5d and 5b.

The *active state* F_0 in the Hill model is assumed not to be influenced by the mechanical events: force, length or velocity. This assumption, together with the experimental finding (5) and with (8) suggests for situations where F_0 and the smoothed EMG E can be considered stationary, that there is a linear relation between both quantities

$$F_0 = gE \quad (\text{quasi-static}) \quad (11)$$



This relation, with F_0 obtained from (9), was already confirmed in 1954 by Bigland and Lippold in experiments with isotonic (F_c constant) contractions.

In general, however, F_0 may change rapidly in time and a more complicated processing than the simple proportionality of (11) is necessary to obtain the active state $F_0(t)$ from the smoothed rectified EMG $E(t)$. The temporal characteristics and the summation properties of the active state, as they are known in physiology, should here be taken into account as much as possible.

Fig. 6 shows how this can be done. As long as $E(t)$ rises, $F_0(t)$ is made equal to $gE(t)$, as in (11). When a relative maximum of $gE(t)$ occurs, this maximum is held for a time τ_2 and followed by an exponential decay with time constant τ_3 , fig. 6a. This time course of $F_0(t)$ goes on until $gE(t)$, or the plateau resulting from a later peak of $gE(t)$, becomes larger than $F_0(t)$. In that case $F_0(t)$ follows the rise of $gE(t)$ again, see fig. 6b.

The time course of F_c and F is quite different from that of F_0 , fig. 6a gives the isometric twitch as an example. This is mainly due to the CC-SEC interaction: at the onset of a contraction the SEC will be stretched first, the CC therefore has to shorten initially at a high speed and at a low force (fig. 5c). In a twitch the turning point in this process, where the shortening speed v_c is zero and $F_c = f(x_c)$ F_0 is maximal, is only reached when F_0 is already declining again (fig. 6a). After this the SEC recoils and the CC is stretched (v_c is negative). In non-isometric contractions the course of things is much more complicated: at any instant the CC length x_c is determined by the difference between the muscle length x , which is measured, and the SEC stretch x_e , which is determined by the CC force F_c (eq. 6a). The influence of the changing

Fig. 5. Components of the Hill muscle model. Curves are characteristic for the human calf muscles. For orientation: in a 70 kg subject walking at moderate speed the peak force can be 2500 N and x is between 0.95 and 1.02.

- (a) Parallel elastic component, passive force F_p as a function of whole muscle length x . Thin lines: measured data, note hysteresis. Thick line: exponential model (7) with $F_{p0} = 50$ N and $p = 60$.
- (b) Contractile component, force-length relation $f(x_c)$. Points are measured data, the dashed line gives the approximation. It may be expected that $f(x_c)$ will decrease above $x_c = 1.00$ (Gordon et al. 1966).
- (c) Contractile component, Hill relation: CC force F_c as a function of CC shortening speed v_c at $f(x_c) = 1.0, 0.7$ and 0.3 according to eq. (9). Parameter values $b = 0.15$ s⁻¹, $c = 0.2$, $n = 0.12$.
- (d) Series elastic component. SEC extension x_e as a function of the applied force F_c . Parameter values: $\alpha = 80$, $k = 1.6 \cdot 10^5$ N, $F_s = 20$ N.

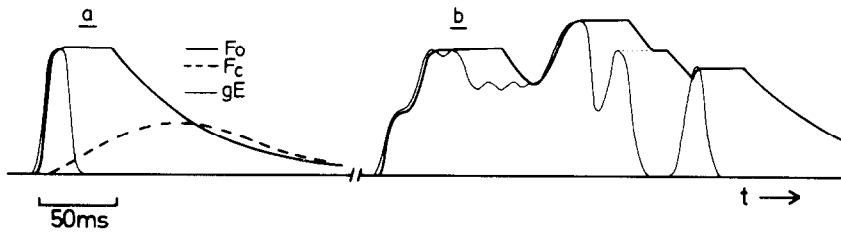


Fig. 6. Active state F_0 as processed from the smoothed rectified EMG $E(t)$, see text.

Parameters: $\tau_2 = 30$ msec, $\tau_3 = 60$ msec.

- (a) One short burst of EMG activity gives a $F_0(t)$ response resembling that of a single twitch. The contractile force response for an isometric contraction has been added: F_c .
- (b) A more complicated EMG pattern. Note the effects of the peak holding. The 'silent period' in the EMG, e.g., results only in a gradual and delayed decrease of F_0 .

SEC stretch on CC speed, force and contractile efficiency should certainly not be underestimated. In human walking, for example, it often occurs that the CC shortens while the muscle as a whole is stretched. When this happens, work done on the muscle is stored as elastic energy in the SEC (Hof et al. 1983a).

Applications

Assuming the EMG-force relation as sketched above to be valid, several relations between EMG and mechanical variables that have been proposed in literature can be evaluated. These relations apply to the force (A–D), the mechanical work (E) and the energy consumption (F). Following these, some methods of presentation of the EMG will be discussed and consideration given to what they can say about muscle activity and function (G–K).

(A) *The complete model* gives the possibility to calculate the force of a single muscle from the EMG and the muscle length for arbitrary contractions. Several things are needed for this. A representative and artefact-free EMG should be recorded. The muscle length is to be determined, e.g., from limb angle measurements by a geometrical conversion, for which anatomical data on the muscle leverage are needed (cf. Frigo and Pedotti 1978). A computational facility is necessary to solve the set of (differential) eqs. (6–10) and to perform the

active state processing that together make up the muscle model. In our (1981a) set-up an electronic analogue processor was used, this has several advantages, but, today, an implementation on a digital computer would be more practical for most users. Finally the parameters of the muscle model are to be determined. Some of these can be considered constant for one muscle for various subjects, but others need to be assessed for every subject, the gain factor g even changes for every new electrode placement. Every experimental session is therefore to be preceded by a calibration procedure, in which the model output is compared to the muscle force, measured by a dynamometer, for a standardized set of contractions. For calf muscle experiments of the author, to give an example, the calibration takes about two hours, including the application of the electrodes and the necessary calculations. The EMG to force processing itself, on the other hand, is done in real time. This is in contrast with kinetic analysis, where the preparation time may be shorter, but where the digitizing and the computing can usually not be done in real-time.

In addition to previous results (Hof and van den Berg 1981b, c, d; Hof et al. 1983), fig. 7 may give an idea of the possibilities of EMG to force processing. The force of the calf muscles has been determined from EMG processing (drawn line 4a) as well as by sagittal kinetic analysis from the total ankle moment (dotted line 4b) in walking and stepping (Hof et al. 1983b). The comparison gives an idea of the accuracy of the EMG to force processing method in walking: while the peak force was about 2000 N, the root mean square error was 340 N for this step, 460 N average over 5 subjects with 15 steps each. Inspection of the various signals in this figure gives an idea of the complexities of the EMG-dynamic force relation. It can be seen that the force (4) goes on for some 160 msec after the cessation of the EMGs (1 and 2). The magnitude and the form of the smoothed rectified EMG (3) and the force (4) are quite different. The courses of the CC length x_c and the total muscle length x differ considerably too (7 and 8, respectively). This holds even more so for CC and whole muscle shortening speed v_c and v (6 and 10). Taken together, recordings like these give an interesting view on the way muscles function in natural activities.

A simplified muscle model has been used by Olney and Winter (1982a, b) for the ankle plantarflexors (calf muscles) and dorsiflexors (tibialis anterior) as well as for the knee flexors and extensors. The force-length and force-velocity relations were linear and the active state

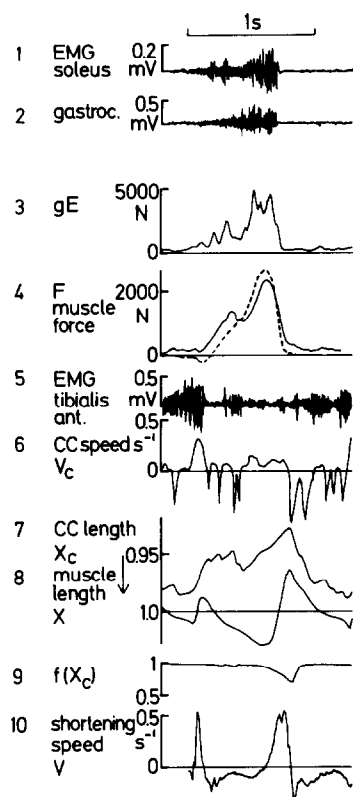


Fig. 7. EMG to force processing for the human calf muscles in walking. Subject's weight 71 kg, speed 1.5 m/sec, steplength 0.88 m, trial 48111.

From top to bottom: EMGs of *M. soleus* and *gastrocnemius* (1, 2). *gE*: smoothed rectified EMG (weighted sum of soleus and gastrocnemius 3). *F*: muscle force from EMG processor (drawn line, 4a) and from the total ankle moment obtained by sagittal, kinetic analysis by means of a force plate and a SELSPOT system (dotted line, 4b). The part drawn boldly can be compared to the calf muscle force because in that time interval the antagonist muscles were estimated to contribute little to the total moment. EMG of *M. tibialis anterior*, antagonist of calf muscles (5). v_c : Shortening speed of CC (6). x_c : Relative length of CC (7). x : Total muscle length, derived from ankle angle (8). $f(x_c)$: Value of force-length relation (9). v : Shortening speed of the whole muscle (10).

and SEC-CC interaction were lumped together in a linear second order low-pass filter (cf. paragraph C below). The reported error for the ankle, 142 N average over 8 steps, is lower than that obtained by the author. This may be because their force curves are smoother, due to the different processing method, and because the walking contractions of

which they report the error were the same as those on which the parameters were fitted.

(B) *Quasi-static isometric contractions.* The course of the argument so far is that, inspired by the experimental finding of a linear EMG-force relation (eq. 5) and by the results on dynamic contractions of Bigland and Lippold (1954), a linear relation between rectified EMG and active state in the quasi-static case was assumed: eq. (11). Combining eqs. (6b), (8) and (11) results indeed in a linear relation between rectified EMG E and muscle force F for this case (with $v_c = 0$):

$$F = F_c + F_p = f(x_c)gE + F_p(x) \quad (\text{quasi-static, isometric}) \quad (12)$$

This formula can give some insight in the experimental problems involved. The gain factor g is found to be dependent on the subject and on the electrode placement, changing 30% and more even in the same subject. It therefore has to be determined before each experiment by means of some force measurement set-up. The passive force $F_p(x)$ has to be taken into account, although it is relatively small in the midrange of the joint. Then the muscle length can drastically influence the F/E ratio because of the force-length relation $f(x_c)$. It must be taken into account here that many muscles cross more than one joint. The length of biceps brachii, for example, is not only changed by elbow rotation but also by the shoulder position and by the degree of supination of the lower arm (Maton and Bouisset 1977).

In isometric contractions the muscle length x is constant but the CC-length x_c is not: the CC shortens at the expense of a stretch of the SEC when the force increases. As soon as the CC length reaches the range where the force-length relation ($f(x_c)$, fig. 5b) decreases, the EMG force relation will become non-linear, concave upward, because more EMG is needed for the same force increment (Moritani and De Vries 1978, Hof and van den Berg 1981c).

The last problem to be mentioned here is that as a rule more than one muscle contributes to the total moment around a joint, which is in fact what is measured by an external measuring device. Agonists can share unequal and varying proportions of the total moment (Hof and van den Berg 1977) and even antagonists can become active (Pruim et al. 1978).

The complications outlined here may explain part of the non-linear

EMG-force relations reported in the literature. There remain, however, trustworthy accounts of non-linearities which cannot be explained this way, in muscles with inhomogeneous (fast/slow) motor unit populations (Bigland-Ritchie et al. 1980). As discussed at the outset of this section, this would introduce complications for the dynamic EMG processing, but no fundamental ones.

The difficulties, as discussed here, have rightly not prevented the static isometric EMG to become more and more a standard tool in biomechanics. As an example, the work of Andersson et al. (1980) on the loading of the spinal muscles in weight-bearing tasks might be mentioned.

(C) Dynamic isometric contractions. For isometric contractions the Hill model can be approximated by a linear first-order transfer function (Gottlieb and Agarwal 1971; Calvert and Chapman 1977). When the EMG-active state conversion is also approximated by a first-order transfer function, the muscle force in dynamic isometric contractions can be predicted with a fair degree of success (Coggshall and Bekey 1970; Gottlieb and Agarwal 1971; Crosby 1978). The main imperfection of the method, as reported, is that a descending force drops too early. This suggests that the prediction might be improved if the non-linear EMG to active state conversion as described here (fig. 6) would be employed.

(D) Fixed movement pattern. The Hill equation (9) together with the EMG-active state relation suggests a simple property for dynamic non-isometric contractions with a fixed movement pattern, i.e. where the muscle length $x(t)$, and thus the speed $v(t)$ as well, is always the same function of time. Suppose that, in such a case, an EMG activity pattern $E(t)$ gives rise to a force $F(t)$. If now the EMG keeps the same form but is multiplied by some factor (thus equals $aE(t)$) then the force will have an identical course as before except that it is also multiplied by the same factor a : $aF(t)$. Such a relation has been confirmed experimentally for subjects bicycling at constant speed (Bigland-Ritchie and Woods 1974).

A closer analysis reveals that some conditions have to be met for this EMG-force relation to hold: (a) The passive force F_p should be comparatively small, or be accounted for. Then the Hill eq. (9) is a function of the CC length $x_c(t)$ and CC speed $v_c(t)$, instead of $x(t)$ and $v(t)$,

respectively. This forces the assumptions (b) that the SEC has a logarithmic characteristic (which is approximately true, eq. 10) and (c) that the non-linearity due to the interaction of the SEC stretch with the force-length relation does not interfere (see B).

This principle has been used by Journée et al. (1983) in their studies of pathological tremors, but in other experiments on periodic movements it is also relevant (see F, below).

(E) *Muscle work.* Bouisset and co-workers have advocated a proportional relation between integrated EMG and muscle work (e.g. Bouisset and Goubel 1973; Cnockaert 1978)

$$W/l_0 = \int_{t_1}^{t_2} Fv dt \approx \int_{t_1}^{t_2} g_2 E(t) dt. \quad (\text{Bouisset}). \quad (13)$$

Such a relation is not fully consistent with the EMG-force relation as sketched above. From the Hill eq. (9) the CC work per unit time $F_c v_c$ can be calculated as a function of v_c for a constant F_0 , thus at a constant level of EMG (fig. 8) [4]. It appears that $F_c v_c$ is dependent on v_c , thus (13) cannot hold in general, only for a certain range of CC speeds is it approximately valid. So, in isometric contractions ($v = 0$) the work is equal to zero but the EMG is certainly not. Even less does (13) hold for eccentric contractions, where W is negative. In fact, as pointed out by Bouisset himself (1981), a proportional EMG-work relation is only valid under special experimental circumstances, e.g., when light inertial loads are moved, and it is influenced by the effects of elastic energy storage (Cnockaert 1978).

A linear relation between a sum of muscle work and integrated force (impulse) has been proposed by Gans and Noordergraaf (1975):

$$k_1 \int_{t_1}^{t_2} Fv dt + k_2 \int_{t_1}^{t_2} F dt \approx \int_{t_1}^{t_2} E dt. \quad (\text{GN}) \quad (14)$$

[4] $F_c v_c$ and Fv are not the same, it is true, but their definite integrals are equal

$$\int_{t_1}^{t_2} F_c v_c dt = \int_{t_1}^{t_2} Fv dt,$$

if they encompass a complete contraction, with $F(t_1) = 0$ and $F(t_2) = 0$, so that no elastic energy is present in the SEC at $t = t_1$ and $t = t_2$.

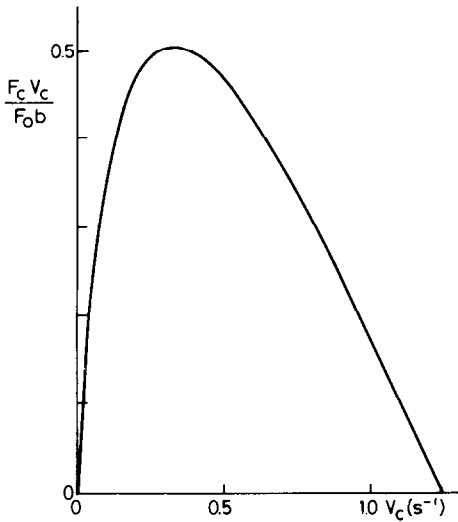


Fig. 8. Muscle CC work per unit time $F_c v_c$, at a constant level of EMG E and active state F_0 , as a function of CC shortening speed v_c . Parameters $b = 0.15 \text{ sec}^{-1}$, $n = 0.12$, as for human calf muscles. Force-length relation $f(x_c) = 1$.

It can be inferred from inspection of the graphs of $F_c v_c$ (fig. 8) and F_c (fig. 5c) as a function of v_c that – with a clever choice of the ratio k_1/k_2 – eq. (14) will give a reasonable fit for v_c between zero (thus including isometric contractions) and the value where $F_c v_c$ is maximal. Eq. (14) will hold only approximately, again, and not for negative v_c , but for positive v_c it will be more accurate than (13). The difference between v and v_c should be considered here as well.

(F) *Metabolic energy consumption.* Several authors have reported a linear relation between mean rectified EMG and muscle metabolic energy consumption P_m , measured as the oxygen consumption $\dot{V}_{O_2}$ (Henriksson and Bonde-Petersen 1974; Bigland-Ritchie and Woods 1974, 1976; Petrofsky 1979; Patla et al. 1982). At first sight this is not to be expected. According to Hill (1938) the net metabolic energy consumption of a contracting muscle per unit time amounts to

$$P_m(\text{net}) = nF_0 b l_0 + nF_0 v_c l_0 + F_c v_c l_0 \quad (15)$$

in which the terms at the right hand denote the maintenance heat, the

shortening heat and the mechanical work (see E), respectively. $P_m(\text{net})$ is the work and heat production in the muscle during the contraction. Other metabolic processes, such as the production of ATP from glycogen, consume about an equal quantity of work; the total energy consumption P_m (gross), which can be measured indirectly by the oxygen consumption $\dot{V}_{O_2}$ (with 3 ml O_2/min about equal to 1W), is therefore about twice $P_m(\text{net})$. Fig. 9 shows that P_m varies strongly as a function of v_c .

The apparent contradiction is not real, however; the reported experiments all dealt with cycling on a bicycle ergometer and special care was taken to keep the pedalling rate constant. In such a case the “fixed movement pattern” (D) applies, and the force F , the mechanical power terms Fv and Fv_c as well as the energy consumption P_m are linearly related to the rectified EMG (each integrated over several cycles). In fact there is a distinct speed dependency, especially when pedalling at positive and negative (backward) speeds is compared (Bigland-Ritchie and Woods 1976). The linear relation between EMG and $\dot{V}_{O_2}$ for uphill (10%) walking at various speeds – but not for running – as found by Delhez et al. (1969) thus remains an intriguing result and not easy to explain.

In analogy with the Gans-Noordergraaf relation (14) a relation between EMG, force and energy consumption can be derived from the Hill eq. (9) and from eq. (15)

$$bl_0 F_c + P_m(\text{net}) = (1 + n) bl_0 F_0 = g_3 E \quad (16)$$

which can be expected to hold from isometric contractions up to the maximum shortening speed, maybe it is even not too far off for negative speed (cf. fig. 9). In fact Hill (1938) derived his force-velocity relation from an experimental force-energy relation like (16).

(G) EMG plus muscle length. A few studies on dynamic movements present the EMG (raw or smoothed rectified) together with the muscle length, obtained from cine analysis or otherwise. Although such an approach does not result in quantitative data on the muscle force, it is remarkable how interesting conclusions can be drawn by merely combining these two sources. Examples are from animal studies: Goslow et al. (1981); Weys and Van der Wielen-Drent (1982); and from human gait studies: Van der Straaten et al. (1975); Thorstensson et al. (1982).

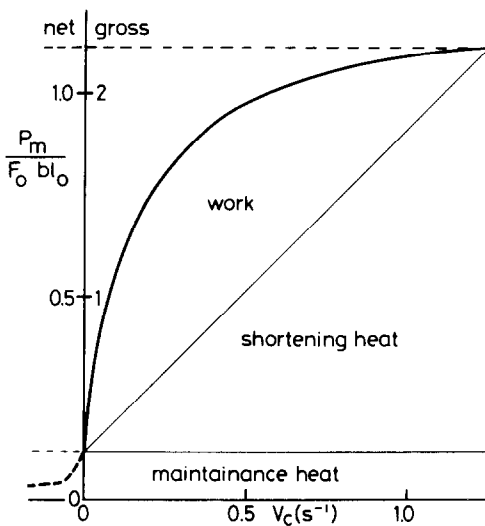


Fig. 9. Muscle energy consumption per unit time P_m , at a constant level of EMG E and active state F_0 , as a function of CC shortening speed v_c . Both net and gross values have been given for P_m . Parameters as in fig. 8. The course of P_m for negative v_c is not precisely known, it is reported to become even lower than isometric (Hill 1960). Dotted line at $P_m = (1 + n) bl_0 F_0$ represents eq. (16).

Of course the differences in timing between EMG and force and the modulation of the force by the CC length changes should be kept in mind when evaluating this kind of data.

A class on their own are the studies on human gait in which the joint moments, obtained from kinetic analysis, are combined with EMG and muscle length data: Cappozzo et al. (1976); Pedotti (1977) and the valuable book of Inman et al. (1981).

(H) Amplitude probability density function. In ergonomic research it can be necessary to monitor muscular activity over a long time: up to several hours. A method to compress the EMG data from such long records is to determine the amplitude probability density function (APDF, Hagberg 1979). Such a diagram gives, as a function of the smoothed (50–100 msec) rectified EMG level, the percentage of the total time that the EMG was below that level. It thus runs from 0%, for signal strengths below the noise level, to 100% for the highest EMG level encountered in the session. When the EMGs are calibrated, e.g., to 100% MVC, this gives (together with other relevant data) an idea of the

level of exertion of the muscle under study, (Jonsson et al. 1981). This calibration is only valid for isometric contractions, of course. For non-isometric contractions an interpretation on the basis of eq. (16), which relates the EMG to a combination of muscle force and energy consumption, might be helpful, but the relation, if any, with perceived exertion and fatigue remains to be explored.

(K) EMG only. For sake of completeness, the more or less classical form of EMG presentation will be mentioned: records of the EMG, together with a verbal description of the movement or posture and an estimate of the degree of effort. The EMG can be represented in several ways: unprocessed or 'raw', smoothed rectified or integrated in some form. The smoothed rectified EMG has several advantages above the raw signal: it is easier to read its level from a display, and the frequency range of the recording instrument needs only to be about 40–50 Hz, so that pen recorders can be employed. Estimating the amplitude of a raw EMG is rather difficult, only a few levels can be discerned with some certainty, and the frequency spectrum goes up to some 300–400 Hz, which can only be represented faithfully by an oscilloscope, an ink-jet or an UV-recorder. These considerations are even more relevant when the processing and storage of the data are done by a computer (Halbertsma and de Boer 1981).

In type K studies the EMG is largely used as an indicator whether a muscle is active or not. From the above it will be clear that even this seemingly simple question has its pitfalls. Apart from complications due to crosstalk or mechanical artefacts (e.g., van den Berg and Spoor 1957) the EMG can serve as an indicator of the neural activation, with an accuracy of the order of the duration of a single motorunit action potential (10–20 msec). When, from the EMG, is to be inferred whether the muscle develops force or not, the complexities of the EMG-force relation come into play. In the human calf muscles the force begins to rise soon after the start of the EMG and is halfway up after 50–100 msec. The force decay is even slower, it is not completed before 160–200 msec after cessation of the EMG (cf. figs. 6a and 7). These 'electro-mechanical delays' will be shorter for faster muscles, but they are very relevant in interpreting EMGs of rapid movements (Ralston et al. 1976).

Discussion

It will be obvious that not all the categories A–K, as summed up above, more or less in order of decreasing complexity, are fully comparable. Those denoted with G, H and K are a few of the many possible ways of reporting EMG findings as such. Conclusions from these data about the neural control or the rough course of the muscle force can only be inferred in an indirect way. The advantage of these methods, especially K (EMG + verbal description), is their simplicity: all that is needed is a set of amplifiers and a multichannel recorder. Virtually all older EMG research of human movement (for which the book of Basmajian (1974) remains the indispensable guide) has been reported with method K, and this method will remain useful when the EMGs of many muscles are to be collected simultaneously. It is hoped that the reader is convinced that quantitative data, in the form of rectified and preferably calibrated EMG can add much to the value of the results. This holds even more so if data on the muscle length can be added (method G), but this implies considerably more effort, especially when cine recordings are to be processed.

In the opinion of the author the relations between EMG and work (E) and energy consumption (F) are not applicable without due consideration. They cannot be isolated from the context of the physiological experiments from which they have been obtained. In fact the EMG serves here mainly as an indicator of the degree of activity of a motorunit population, e.g., when positive and negative work had to be compared (Bigland-Ritchie and Woods 1976), or when the relative degree of involvement of the various muscles is the object of research (Henriksson and Bonde-Petersen 1974). Eq. (16), a relation between force, energy consumption and EMG, might hold quite generally. It is not so easy to think of a situation in which two of these variables are available and the third one needs to be assessed. It may, nevertheless, be useful for interpreting data on unprocessed EMG in dynamic contractions.

The categories A–D cover relations between EMG and force without (A) or with (B, C, D) restrictions in the kind of movement. The relation described in D, with a fixed pattern of movement, may find – and has already found – application in rather specific experimental situations. It will always remain necessary to ascertain that the movement pattern (the time course of the muscle length) and the sharing of the load

between the muscles does not change with variation in load.

Most of the papers concerned with the static-isometric EMG-force relation have been concerned with proving or disproving the linearity of this relationship. It is the author's opinion that the time has come to apply this relation more widely for solving biomechanical questions. When the inherent pitfalls are sensibly avoided, EMG methods have the potential to provide a solution for the indeterminacy problem of kinesiology: how to discern the separate muscle forces that together make up the total joint moment. The point is, that when only mechanical calculations are used for such a problem, there are always more unknowns (muscle forces) than there are equations. The usual way out of this problem is to assume some optimality criterion, e.g., equal muscle forces or equal muscle stresses, but the results so obtained can never have the demonstrative force of an experiment, often they are physiologically impossible (cf. Hatze 1980). Even within the restriction to quasi-static (postural) situations, quantified EMG can provide useful results.

As far as is known, the possibility of EMG-force processing for dynamic isometric contractions (C) has hardly ever been used in biomechanical work, maybe because the method is not generally known, maybe for lack of suitable problems. In fact it is quite simple to apply. The necessary equipment is a sufficiently fast force transducer (e.g., with strain gauges) and a variable filter, either second-order linear, or the non-linear active state converter (Hof and van den Berg 1981a) followed by a linear first-order filter. Three parameters, a gain factor and two time constants, are to be estimated. As a matter of course, it should be ascertained that the muscle shortening during the action is sufficiently small and slow to be considered isometric.

Using a muscle model (A) provides the most general method to estimate the muscle force and work from the EMG. As is in the nature of things, it is also the most complex method to implement and to use. To evaluate the possibilities of the method, it must be kept in mind that it can be looked upon from different viewpoints. It can be seen as a competitor to the usual kinetic methods, that determine the muscle forces and joint moments from external force measurements with Newton's laws. From this viewpoint EMG processing may sometimes be of advantage, sometimes not. The greatest advantage, although unexploited as yet, is the possibility to solve the indeterminacy problem. The other viewpoint is to consider it part of a research program

(outlined among others by Hatze 1981) in which, on the basis of both physiological and biomechanical knowledge, what is studied is not so much *what* the muscles do, but above all *how* they perform their tasks in natural movements.

Appendix

Nomenclature

a	An arbitrary positive constant;	
APDF	Amplitude probability density function (Hagberg 1979);	
b	Parameter of the Hill equation, eq. (9) and fig. 5c, determines behaviour at positive speed;	(sec ⁻¹)
B	Statistical equivalent band width of the unprocessed EMG;	(Hz)
c	Parameter of the Hill equation, eq. (9) and fig. 5c, determines behaviour at negative speed;	
CC	Contractile component;	
CMRR	Common mode rejection ratio, the suppression of signals common to both inputs (mostly mains interference) in proportion to the amplification of the differential signal (i.e. the EMG);	
d	Lever arm of the calf muscles with respect to the ankle axis, see footnote [3];	(m)
$e(t)$	The unprocessed EMG;	(μ V r.m.s.)
$E(t)$	The amplified, rectified and smoothed EMG;	(V)
E_A, E_B	Ditto for muscles A and B;	(V)
F	Total muscle force;	(N)
F_c	Force exerted by CC;	(N)
F_0	Active state = force exerted by CC when $v_c = 0$ and $f(x_c) = 1$;	(N)
F_p	Force of PEC;	(N)
F_{p0}	Parameter of PEC, eq. (7) and fig. 5a;	(N)
F_s	Parameter of SEC, eq. (10) and fig. 5d;	(N)
$f(x_c)$	Force-length relation of CC, eq. (8) and fig. 5b (value between 0 and 1);	
g	Proportionality constant between rectified EMG E and active state F_0 , eq. (11);	(N/V)
g_1	Proportionality constant between rectified EMG E and isometric muscle force, eq. (5), $g_1 = f(x_c) g$, eq. (12);	(N/V)
g_2	Proportionality constant between integrated rectified EMG E and muscle work W , according to Bouisset, eq. (13);	(J/Vsec)
g_3	Constant in eq. (16), $g_3 = (1 + n)bl_0g$;	(W/V)
$I(t)$	Time varying intensity of the unprocessed EMG, eq. (3);	(μ V)
k	Linear elasticity parameter of the SEC, eq. (10) and fig. 5d;	(N)
k_1, k_2	Constants in the relation proposed by Gans and Noordergraaf, eq. (14);	
l	Muscle length;	(m)
l_0	Muscle length at the reference position;	(m)
M	Moment of the muscle force(s) with respect to the joint axis;	(Nm)
n	Parameter of the Hill eq. (9), determines its convex shape;	
$\underline{n}(t)$	Stationary stochastic process with zero mean and unit variance;	
p	Parameter of PEC, eq. (7) and fig. 5a;	

PEC	Parallel elastic component;	
$P_m(\text{net})$	Production rate of heat and work during a muscle contraction;	(W)
$P_m(\text{gross})$	Ditto, with the heat production of the associated chemical processes included, $P_m(\text{gross}) = 2 P_m(\text{net})$;	(W)
r.m.s.	Root mean square;	
SEC	Series elastic component;	
t	time;	(sec)
T	Effective averaging time of a smoothing filter;	(sec)
v	Muscle shortening speed, $v = -dx/dt$;	(sec^{-1})
v_c	CC shortening speed, $v_c = -dx_c/dt$;	(sec^{-1})
$\dot{V}_{O_2}$	Rate of oxygen consumption;	(ml/min)
W	Work done by a muscle between time t_1 and t_2 , see eq. (13);	(J)
x	Normalized total muscle length $= l/l_0$;	
x_c	Normalized CC length;	
x_e	Normalized SEC length, eq. (10);	
Z_i	Input impedance of EMG preamplifier (at 50/60 Hz);	(Ω)
Z_1, Z_2	Impedances of differential electrode pair (50/60 Hz);	(Ω)
α	Logarithmic SEC elasticity parameter, eq. (10) and fig. 5d;	
ϵ	Coefficient of variation of the fluctuations in $E(t)$;	
τ_1	Time constant of the EMG smoothing filter;	(sec)
τ_2	Plateau duration of active state;	(sec)
τ_3	Time constant exponential decay of active state;	(sec)
ϕ	Angle of the ankle joint, $\phi = \pi/2$ ($= 90^\circ$) in upright standing, footnote [3].	(rad)

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