Relationship Between Force and Integrated EMG Activity During Voluntary Isometric Anisotonic Contraction*

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Summary. The surface electromyogram (EMG) was recorded for the extensor carpi radiales of normal human subjects during voluntary isometric anisotonic contractions as well as the force generated by the muscle. The relationship between force and instantaneous integrated EMG could not be fitted by a single linear function or a parabola.

The muscle was then considered as a system (a) presenting a maximum force and (b) in which the EMG activity represented the input and the contraction force the output, that is $F = f(iEMG)$ and not, as is usually done, $iEMG = f(F)$. In addition the summed $iEMG$ from the beginning of the contraction was used since the actual force depends on all preceding events. These findings led to testing for a non-linear and asymptotic function relating F to $iEMG$. A double exponential function:

$$F = a\left[1 - \exp\left(-b\Sigma\Delta iEMG\right)\right] + d\left[1 - \exp\left(-c\Sigma\Delta iEMG\right)\right]$$

could account for the two mechanisms by which the force of a voluntary contraction is increased.

Key words: Electromyogram – Isometric contraction – Muscle tension

Introduction

It is well known since Adrian and Bronk (1929) that the force of a voluntary contraction is a function of two mechanisms acting together: recruitment, when the number of active motor units can be increased, and rate firing, when the rate of discharge in the motor units can be increased. Although much literature exists on the relationship between the electromyogram (EMG) and force (F)

(Bouisset 1973), the results of these experiments are not quite in agreement. They are only mathematical representations and none appears to be linked to the physiological behaviour of muscular contraction. As for what is called integrated EMG (iEMG), this relationship has been shown to be either linear (Lippold 1952; Inman et al. 1952; de Vries 1968; Stephens and Taylor 1972; Moritani and de Vries 1978; Lind and Petrofsky 1979) or non-linear without precision (Bottomley et al. 1963; Vredenbregt and Rau 1973), quadratic (Zuniga and Simons 1969; Pertuzon 1972; Maton 1973; Komi and Viitasalo 1976) or even linear then exponential (Kuroda et al. 1970). These discrepancies can probably be explained by many factors and particularly by experimental procedure. First, the contraction type used has been either isometric or anisometric and either isotonic or anisotonic, the curves then being drawn either from sampled points or continuously. Secondly, there has been some confusion between smoothed and integrated (or averaged) EMG, and in this last case between instantaneous integration and integration measured during an isometric contraction and calculated for unit time. A third parameter, which is most important to consider, is the final level of force exerted, and it must be emphasized that in many studies this final level was only about 50% of the maximum force. Finally, fatigue has not always been clearly taken into consideration.

Although surface electrical activity of a muscle represents a very complex summation of the elementary intramuscular events, a linear relationship has been demonstrated to exist between them (Maton et al. 1969). It is thus generally accepted that the surface EMG provides an effective indirect measurement of the force exerted by a muscle, provided that all phenomena of fatigue have been avoided.

It then seems necessary to re-examine the relationship between F and iEMG in regard to one important factor, the actual increase of force as reflected by increased EMG activity which involves two parameters: the preceding level of force and the rate of increase in EMG activity, both a result of elementary events in the muscle fiber, release and pumping of calcium. In order to obtain a quantitative relationship it is imperative to use an isometric, that is without shortening of the muscle, and an anisotonic, that is during a progressive increase of force, contraction. Under the conditions of this study, a link will be sought between EMG and the physiological characteristics of muscle contraction.

Methods

One hundred and thirty isometric anisotonic tests were carried out on the extensor carpi radiales (ECR) of five normal subjects.

Each subject was tested sitting with his right forearm fastened to a splint in a horizontal plane and the elbow at 90°. The forearm was in a fixed semi-prone position, and the hand was in a splint connected to a mechanical system which rotated on a vertical axis concentric to that of the wrist. The force of extension was applied to a bar at the level of the dorsal surface of the metacarpals. The torque thus induced was transmitted to a strain gauge through a disk. Holes in the disk allowed the hand to be fixed in any position of flexion or extension (Fig. 1), at intervals of 10°. The origin $\alpha = 0°$ was chosen to correspond to the angle at which the dorsal surface of the hand was in the same plane as that of the dorsal surface of the forearm. Deviations from this position were considered to be
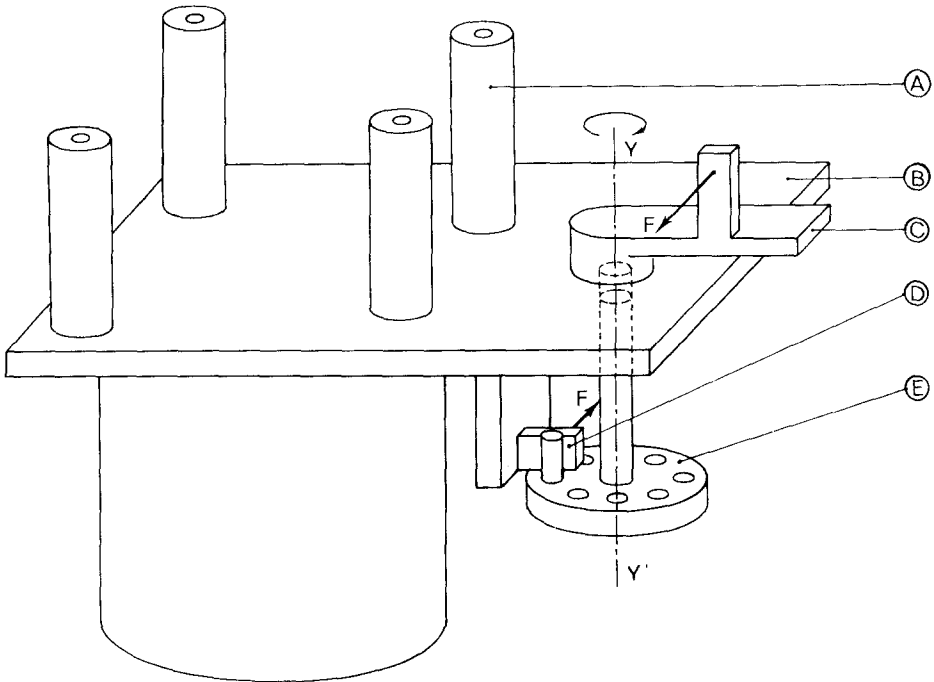
Fig. 1. Experimental assembly. *A* Plugs used to fix the forearm which is supported on *B*. *C* Holder of the hand, rotating around the Y−Y' axis. *D* strain gauge. *E* Disk with holes at 10 deg intervals

negative for extension and positive for flexion. The osteofibrous channel of the ECR tendons is sufficiently rigid to act as a sort of pulley so that the direction of the tendons stays fairly constant in relation to the metacarpals during flexion or extension of the wrist (Metral et al. 1974). Accordingly, it was assumed that tension transmitted to the strain gauge was always directly proportional to the muscular force and that the muscle length was directly proportional to the angle of flexion or extension of the wrist.

An oscilloscope was used to display a linear ramp target with a slope which represented a force rate, that is an increase in tension (kg) per second. The ramp target could be varied in latency, amplitude and slope. The object was for the subject to follow this ramp target as linearly as possible as guided by the force signal derived from the strain gauge and displayed on the oscilloscope through a second channel. The resulting force rates (v) were classified into three categories: 1.5−2.5, 2.5−3.5, and 3.5−5.5 kg/s (corresponding to 9, 6, and 3 s in order to reach the maximum voluntary contraction). A 3−5 min rest was provided between tests in order to avoid fatigue (Scherrer and Monod 1950). The EMG was recorded using two silver skin electrodes (5 mm diameter) placed 2 cm apart along the longitudinal axis of the ECR in the middle of the muscle belly. The amplified EMG was full wave rectified and integrated (operational amplifier circuit of 50 ms, time constant). The integrator gave a pulse every 200 ms whose amplitude represented the integrated EMG during the preceding 200 ms: ΔiEMG. Force, ΔiEMG and EMG were stored on a FM tape recorder and displayed by an inkwriter recorder (Mingograph). For each $(\Delta i$EMG$)_n$ pulse the simultaneous force value, F_n, was measured as shown in Fig. 2. For each F_n the sum of the n previous ΔiEMG was calculated, i.e., $\Sigma_{i=0}^{n} \Delta i$EMG. The iEMG unit used here was the value of a 400 Hz sine signal of 1 mv amplitude integrated during 200 ms.

Regression equations were calculated by the least squares method and the mean square error about the regression was calculated for each experiment. Since the electrical activity measured was that of two muscles [ECR brevis being more active than ECR longus during pure extension (Tournay
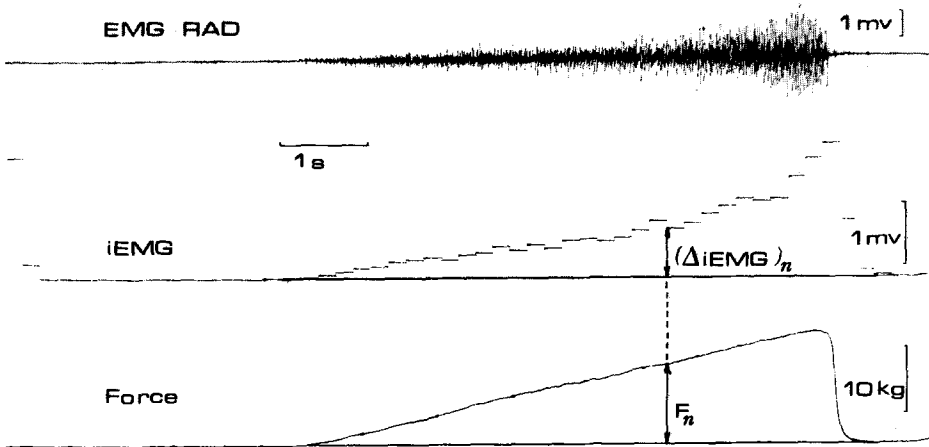
Fig. 2. Upper trace: the surface EMG of the ECR. Middle trace: integrated EMG. Each 200 ms step corresponds to the integral of the surface EMG during the previous 200 ms: ΔiEMG. Lower trace: force signal from the strain-gauge. Each $(\Delta i\text{EMG})_n$ value was linked to the corresponding Force F_n. Note the linearity of the ramp force which was a prerequisite for experimental validity

and Paillard 1953)], and that *ECR*, extensor carpi ulnaris and extensor digitorum work synchronously during extension of the wrist (Basmajian 1974), it was necessary to verify the results on another muscle. The above apparatus was slightly modified in order to test the biceps brachii (BB). Fifteen experiments on five subjects were made on that muscle allowing comparison of the results to those of many previous works.

Results

Instantaneous iEMG − Force Relationship

There appeared to be two aspects to the relationship between ΔiEMG and F (Fig. 3). The first was the existence of a remarkable linearity (correlation coefficient > 0.99) observed between 0 and about 50% of the maximum force. It is important to note that at these force levels there was very good amplification right from the beginning of the recording. At forces greater than 50% maximal, the slope was steeper and the dispersion of measurements was greater, in particular because near the maximum force, that is for the last two or three points, the increase of ΔiEMG was even more rapid. Nevertheless, for purposes of analysis a straight line was drawn which fitted well with the regression line (correlation coefficient > 0.90).

   It is important to note that when the full range of voluntary contraction was used it was not possible to fit the data with a single function. The slopes in corresponding phases of the bilinear *i*EMG curve were identical for a given subject regardless of the force rate imposed ($p < 0.001$). However, there were large differences found in the *i*EMG slopes between subjects. In all cases, the beginning of the second slope corresponded to about the same force for a given subject but was somewhat different among subjects, ranging from 50−70% of
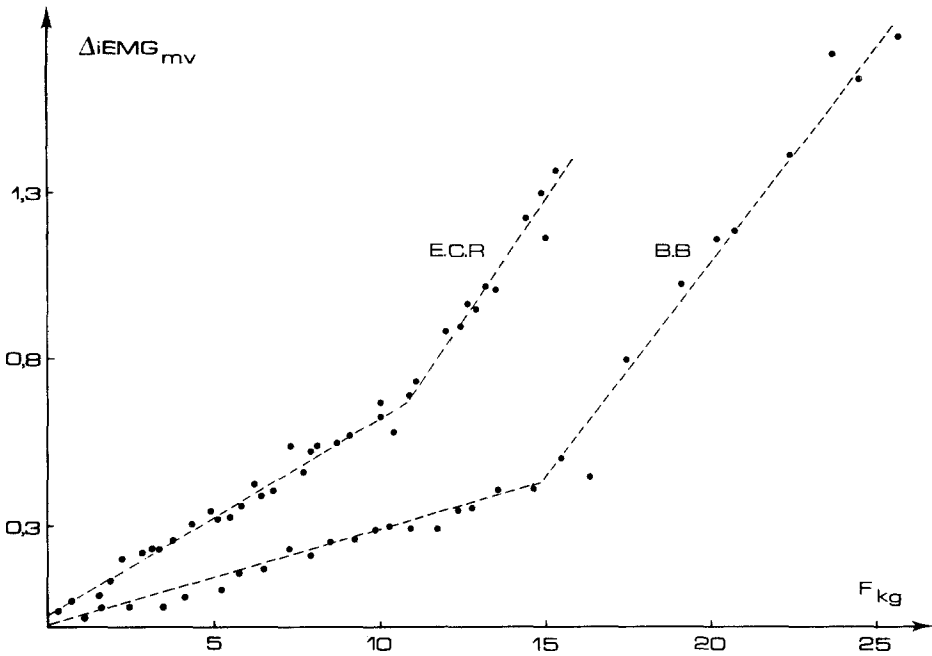
Fig. 3. Relationship between force and elementary integrated EMG every 200 ms (ΔiEMG). The change in slopes, at about 50% of maximum force, was evident for the ECR. A similar result was obtained for the biceps brachii (BB)

the maximum voluntary contraction. In fact these results depended not only on the integration technique which permitted measurement of electrical activity closely corresponding to the intensity of force, but also on the accuracy by which the ramp target was followed. When the subject had difficulty following the target, there were fluctuations of the force signal which corresponded to even greater variations of ΔiEMG. In such cases it was possible to find a good parabolic fit for the entire curve, often with a correlation coefficient > 0.98. Nevertheless, the first points of these curves were quite linearly disposed and a simple parabola could not have been the correct function.

Finally, in some experiments the subject was told at the end of the ramp force to maintain during two or three seconds a force near its maximum. What is most interesting is that ΔiEMG remained constant during this short period indicating that there was no fatigue during the three to nine seconds of the contraction.

$\Sigma\Delta iEMG$ − Force Relationship

In these experimental conditions it was thus impossible to find between F and ΔiEMG either a linear or a parabolic relationship but rather a function consisting of two successive segments of different slopes. The relationship between F and iEMG was then re-examined according to four general
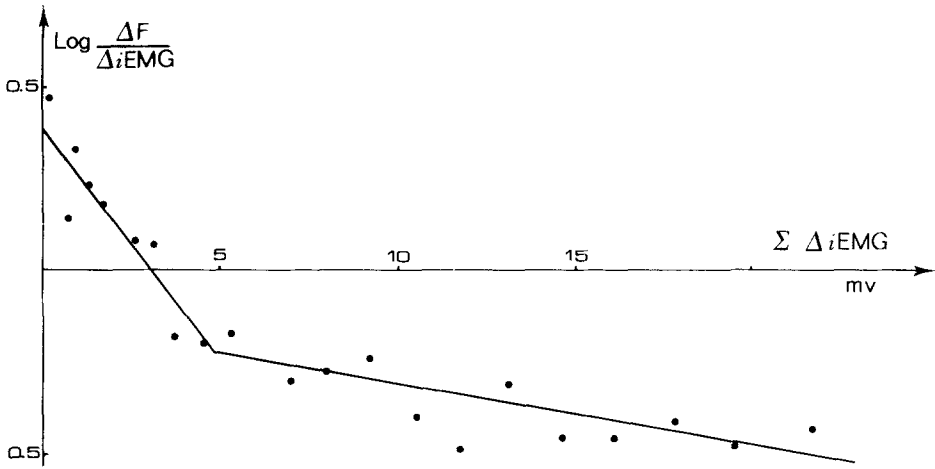
Fig. 4. The logarithm of the ratio $\Delta F/\Delta iEMG$ did not vary linearly with the total integrated EMG. Measurements yielded two approximately successive straight lines. This suggests that the relationship between F and $\Sigma\Delta iEMG$ is more complex than a single exponential. Note that Fig. 4 cannot be compared with Fig. 3. In fact, the slope of the lines of Fig. 3 would be obtained by $\dfrac{d(\Delta iEMG)}{dF}$

conditions: (1) even though electrical activity precedes F, the tension produced must be considered as the output of the neuromuscular system with $iEMG$ representing the electrical input to the muscles, that is, $F = f(iEMG)$ which appears evident, although all previous treatments of this relationship have been as $iEMG = f(F)$; (2) when $iEMG = 0$, F must equal 0, which was not the case in many of the proposed parabolic relationships; (3) since each level of force is related to the previous level, if the mechanisms of force augmentation are to be reflected in a F-EMG relationship, the summed $iEMG$ from the beginning of the contraction must be used; (4) the relationship between summed $iEMG$ and force has to be consistant with the familiar concept of maximal muscular tension. Knowing that in animals the relationship between the tension developed by a motor unit and the frequency of stimulation of the motoneuron appears to be asymptotic (see Discussion), a hypothesis is that the general relationship between F and $iEMG$ is also asymptotic for the entire muscle. The upward inflexion of the $\Delta iEMG/F$ curve mentioned above is an argument in favor of this possibility since it implies that near the maximal force $iEMG$ increases faster than the force.

Under these conditions the simplest relationship between F and $iEMG$ should be an asymptotic function such as $F = a\,[1 - \exp(-\,b\Sigma\Delta iEMG)]$. If the increment is considered to be identical with the differential, then:

$$\frac{dF}{d\Sigma\Delta iEMG} = \frac{\Delta F}{\Delta iEMG} \quad \text{and} \quad \log\frac{\Delta F}{\Delta iEMG} = -\,b\Sigma\Delta iEMG + \log ab$$

where F and $\Delta iEMG$ are the effective experimental data. In all cases studied the experimental plot of log $\Delta F/\Delta iEMG$ (Fig. 4) showed that it was not one simple line but that the data corresponded to two straight lines.
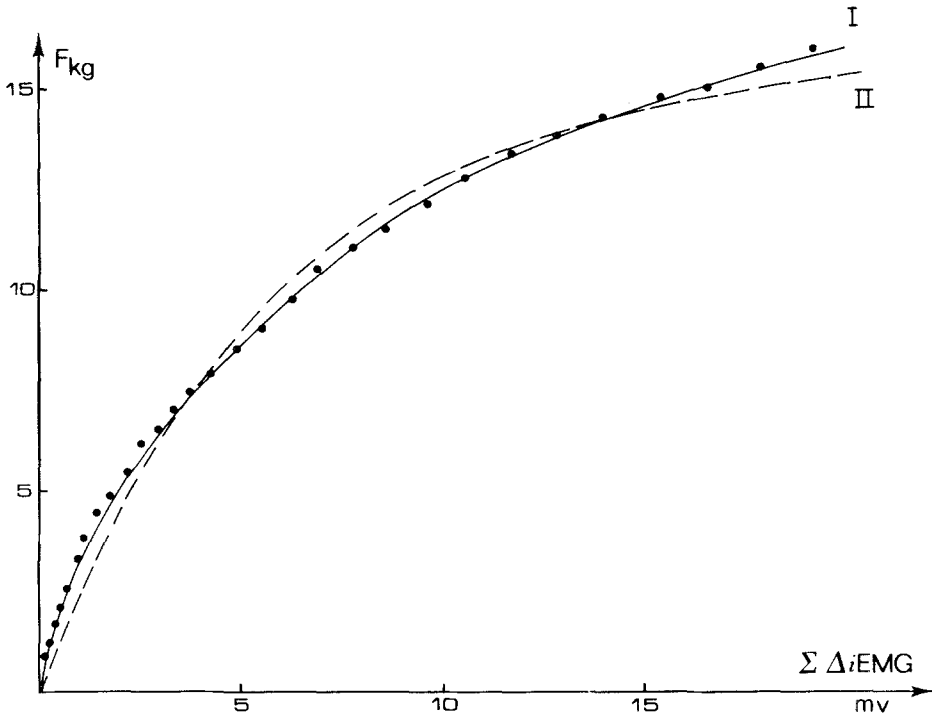
Fig. 5. Plot of F against $\Sigma\Delta i$EMG. Note the remarkable fit between experimental data (●) and the regression curve (continuous line I): mean square error: 0.016. For comparison, the regression curve for a single exponential was drawn (dotted line II): mean square error: 0.25. A systematic deviation occurred with the experimental measurements being first above then below the regression curve, confirming the results of Fig. 4

These results suggested that the general relationship between F and iEMG was the sum of two increasing exponentials such as:

$$F = a\left[1 - \exp\left(- b\Sigma\Delta i\text{EMG}\right)\right] + d\left[1 - \exp\left(- c\Sigma\Delta i\text{EMG}\right)\right]$$

where b and c had very different values. Figure 5 shows remarkable agreement between all the experimental measurements including the initial and final ones. The mean square error of the regression for this function was always lower than with a single exponential by one order of magnitude and on a level with the accuracy of measurement. This result was obtained for all 130 experiments on ECR and 15 on BB, independent of the angle and the force rate of contraction. It should be noted that the better the ramp force linearity, the better the statistical fit. It should also be emphasized that a simple comparison could not be made between this relation which involved $\Sigma\Delta i$EMG from the beginning of the contraction and the preceding ones which concerned only ΔiEMG.

Coefficient Variations with Angle and Speed

Another aim of this work was to try to find a physiological correspondance to each of the parameters of the proposed function. Thus the muscle length and the speed of contraction (v) were varied.
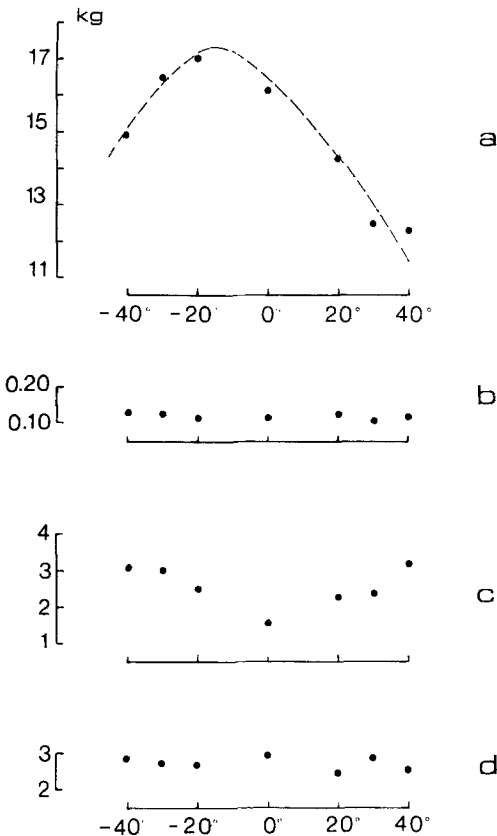
Fig. 6. Variations of each parameter (**a, b, c, d**) of the double exponential equation with muscle length (expressed in degrees of flexion or extension). For **a**, the maximum of the curve — fitted by eye — was near the resting length (10°). **b** and **d** were independent of the angle. **c** was minimum at about the resting length. Note that ordinates for **a** and **d** are in kg, for **b** and **c** in mV^{-1}

Variations of a. The relationship between a and the muscle length was a bell-shaped curve (Fig. 6) with a maximum corresponding to a wrist angle (α) between 0° and −20°. This value was very near the resting length of the muscle which had been proposed to be −10° (Metral et al. 1974) and thus corresponded to the maximum of the length-tension curve. a increased very slightly with the speed of contraction. The slope of the linear regression (Fig. 7a) was significantly different from zero ($p < 0.01$). There was a marked variation in a between subjects according to general strength as well as the variation for a at different angles and force rates for each individual. However, the results were quite reliable for each test at a given angle and speed for a given subject.

Variations of d. As shown in Fig. 6, d was fairly constant with variations in muscle length and varied only slightly from one subject to the other. It was also independent of the force rate: the slope of the regression line (Fig. 7b) was not significantly different from zero ($p < 0.001$).

Consequently, the sum $a + d$, which represents the asymptote value, corresponded to the theoretical maximum force and varied with the muscle length as a classical length-tension curve. This result was particularly important to consider because it was one of the necessary conditions for the function to be valid.

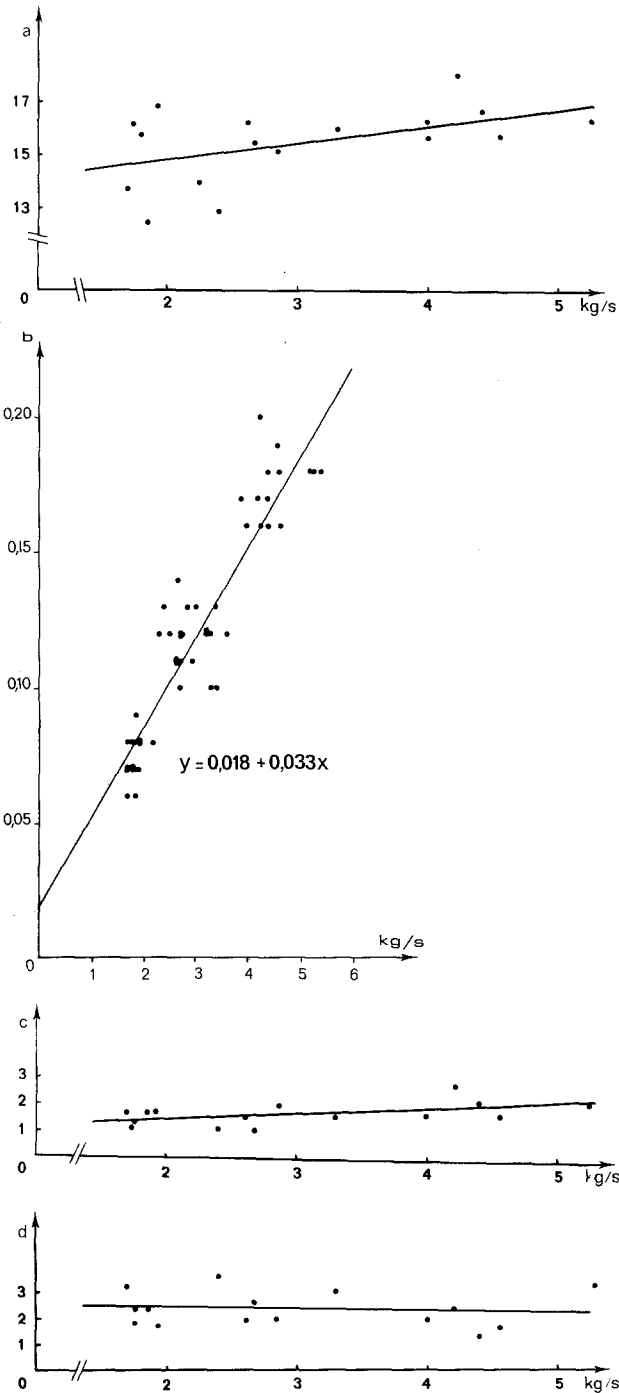**Fig. 7.** Variations of each parameter (**a, b, c, d**) of the double exponential equation with speed of contraction. Regression lines, calculated by least squares, had slopes significantly different from zero ($p < 0.01$) for **a, b,** and **c**. In contrast, that slope for **d** was not significantly different from zero ($p < 0.001$)

Variations of b. This parameter could be considered as independent of the muscle length at each force rate (Fig. 6). However, it varied linearly with the rates used here. The slope of the regression line (Fig. 7b) was highly significantly different from zero ($p < 0.001$). There were two more important findings. First, the slope of the linear relationship between b and v for all angles was identical for the three subjects for whom sufficient data was obtained (0.030; 0.030; 0.033). Secondly, for each force rate category b was independent not only of the angle but also of the subject.

Variations of c. As did b, which has the same mathematical function, c varied linearly with v, the slope of the regression line (Fig. 7c) being also significantly different from zero ($p < 0.01$). In regard to length, it appeared that c was minimum near the resting length (Fig. 6). This was observed for all five subjects to varying degrees, and accordingly c behaved very differently than b which was constant whatever the length. The value of c was about ten times higher than that of b which was also an important finding according to the results shown in Fig. 4.

Discussion

There are two aspects that need to be discussed regarding the present experiments on isometric contraction.

Force–ΔiEMG Relationship

Since Lippold (1952) and until very recently (Komi and Viitasalo 1976; Maton 1977; Moritani and de Vries 1978; Lind and Petrofsky 1979) the great majority of experiments regarding the mathematical relationship between the *i*EMG and the force produced in human muscle contraction have been made using isotonic contractions. This implies that the curves have been drawn from sampled points corresponding to distinct force levels and to an *i*EMG measured for each force level and calculated for unit time.

We believe that isometric isotonic contraction cannot be compared with our isometric anisotonic condition because: (1) there could have been a motor neuron adaptation (Gillies 1972; Tanji and Kato 1973a; Freund et al. 1975) which was not the case during a continuous linear growth of force; and (2) fatigue could have occurred where the EMG amplitude showed a progressive increase (Scherrer and Bourguignon 1959; de Vries 1968; Petrofsky and Lind 1978). These have been shown not to be the case in the present experiments.

The only experiments having used force isometric anisotonic conditions are, to our knowledge, those of Zuniga and Simons (1969); Vredenbregt and Rau (1973); and Stephens and Taylor (1972). The good fit with the parabola shown by the first two papers might well be explained by dispersion of the data due to irregular contraction. That the pursuit task has to be perfectly linear in order to

observe a bilinear relationship has been emphasized. For example, the figures of Zuniga and Simons (1969) show different segments of the force rate curve. Under such conditions it would be very difficult to observe the two lines reported here. On the other hand, it is particularly interesting to note that in the Fig. 10 of Stephens and Taylor (1972) the graph of force/smoothed rectified EMG for the first dorsal interosseous muscle is clearly made of two parts although they claimed that the relationship was always linear. The linearity observed in the present study between 0 and about 50% of the maximum force agrees with their findings and with Milner-Brown et al. (1973) who suggested that, in that range of force, the combination of recruitment and rate coding could give a linear relationship between force and iEMG. As for the curve representing forces above 50% maximal, the increase of the iEMG is more rapid, in particular near maximal force. This result has already been noted by Kuroda et al. (1970) in their "linear − plus − exponential" relationship. This increase of the iEMG with the force is perhaps to be compared to the results of Olson et al. (1968) for whom the EMG amplitude of a motor unit with a high threshold is greater than that of a unit with a low threshold. As a matter of fact, if the relationship proposed here between F and $\Sigma\Delta i$EMG is accepted, it is not possible to arrive at a linear relationship between F and ΔiEMG. In fact, one can demonstrate that when the second exponential is close to its maximum (that is when $\geqslant$ F max/2), the increase of ΔiEMG with F is hyperbolic. In summary, the most important result of this first part appears to be that there is a change in the slope of the F/ΔiEMG curve, occurring between 50 and 70% of F max.

Force −$\Sigma\Delta iEMG$ Relationship

This is the first time that iEMG summed from the beginning of the contraction has been used to find a relationship between EMG and force. The statistical fit between experimental data and theoretical curves is remarkable. The reliability of the results for a given subject and the homogeneity of the parameters for different subjects argue well for this model under the experimental conditions of this study.

A general relationship between F and iEMG cannot be explained easily on a theoretical basis, due to physical and physiological reasons already mentioned and recently reviewed by De Luca (1979). In particular it is well known that there are at least three different motor unit types. These have been characterized by their contraction speed, fatiguability and maximum force (Mac Phedran et al. 1965; Close 1967; Burke et al. 1973). Moreover, the size of a motor unit potential depends on fiber diameter, fiber number, and fiber dispersion for each motor unit (Olson et al. 1968). Nevertheless an attempt was made to give a physiological substratum to the general behaviour of the entire function. Tanji and Kato (1973b) have claimed that 46% of the motor units, for the abductor *digiti minimi,* are recruited for only 20% of the maximum force, and Grillner and Udo (1971) have shown that 90% of the cat soleus motor units were recruited at 50% of the maximum force. Milner-Brown et al. (1973b) as well as Freund et al. (1975) agreed with these findings for the human first dorsal interosseus muscle.

It has been shown that b is smaller than c by one order of magnitude. In fact, the second exponential is nearly asymptotic for about 50% of the maximum force. It is thus tempting to suppose that the two exponentials proposed are tied to the two mechanisms of force increment − recruitment and rate coding − each of them being alternatively preeminent according to the level of force, even if they grow together. Thus it can be supposed that the first exponential is linked to the contractile properties of the motor unit. Three arguments support this hypothesis. First, it is a non-linear function as has been shown for isometric force and stimulation frequency, not only for single motor units (Kernell 1966) but for entire muscles as well (Rack and Westbury 1969). Secondly, a varied with muscle length as did the length-tension curve obtained with isolated fibers (Gordon et al. 1964), whole muscle in animals (Aubert 1956; Rack and Westbury 1969), and whole muscle in man (Pertuzon 1971). Finally b being the same for a given force rate regardless of the subject seems a good argument for its close relationship with the general properties of the motor units.

The second exponential can be related to motor unit recruitment. It is also a non-linear term, and it has been recently shown that the relationship between the force and the number of motor neurons recruited is curvilinear (Clamann et al. 1974), nearly asymptotic (Grillner and Udo 1971) and even exponential-like (Milner-Brown et al. 1973a). Its being asymptotic fits very well with the fact that the motor neuron pool is finite. This is in accordance with the observation that d was independent not only of the muscle lenght but also of the speed of contraction. It is important to note that d has force dimensions and cannot be directly tied to a number of motor units of the muscle. Variations of c with v conform with the fact that the motor unit recruitment follows the force rate (Milner-Brown et al. 1973b). As for the variations of c with muscle length, it is comparable to the length-tension curve which implies that for a given muscular tension, increasing numbers of motor neurons must be recruited as the muscle progressively deviates from the resting length.

If one admits these hypotheses, they imply that the surface EMG can give an idea of global behaviour of the muscle. It is well known that the muscle fibers of a motor unit are randomly distributed within its territory. From histological (Brandstater and Lambert 1973) and electromyographic (Buchthal et al. 1959) evidence, it has been assumed by De Luca (1979) that any portion of a muscle may contain fibers belonging to as many as 20−50 motor units. Therefore myoelectric activity recorded by the surface electrodes could correspond to a relatively significant sample of the total motor units which in turn could give an idea of the mechanisms of force increment. That all the proposed hypothesis are in accordance with the known properties of muscle behaviour, and that the same results have been verified for two different groups of muscles, seem to be two arguments in favor of our assumption. However, this probably holds true only during a linear isometric anisotonic contraction and with the use of the iEMG summed since the beginning of the contraction.

References

Adrian ED, Bronk DW (1929) The discharge of impulse in motor nerve fibers, part II. J Physiol 67: 119–151

Aubert X (1956) Le couplage energétique de la contraction musculaire. Editions Arscia, Brussels

Basmajian JV (1974) Muscles alive. 3rd edn. Williams & Wilkins Co., Baltimore

Bottomley A, Kinnier Wilson AB, Nightingale A (1963) Muscle substitutes and myoelectric control. J Br Inst Radio Eng 26: 439–448

Bouisset S (1973) EMG and muscle force in normal motor activities. In: Desmedt JE (ed) New developments in electromyography and clinical neurophysiology, vol 1. Karger, Basel, pp 547–583

Brandstater ME, Lambert EH (1973) Motor unit anatomy. In: Desmedt JE (ed) New developments in electromyography and clinical neurophysiology, vol 1. Karger, Basel, pp 14–22

Buchthal F, Erminio F, Rosenfalk P (1959) Motor unit territory in different human muscles. Acta Physiol Scand 45: 72–87

Burke RE, Levine DN, Tsairis P, Zajac FE (1973) Physiological types and histochemical profiles in motor units of the cat gastrocnemius. J Physiol 234: 723–749

Clamann HP, Gillies JD, Skinner RD, Henneman E (1974) Quantitative measures of output of a motoneuron pool during monosynaptic reflexes. J Neurophysiol 37: 1328–1337

Close R (1967) Properties of motor units in fast and slow skeletal muscles of the rat. J Physiol 193: 45–55

De Luca CJ (1979) Physiology and mathematics of myoelectric signals. IEEE Trans Biomed Eng, BME-26: 313–325

De Vries HA (1968) "Efficiency of electrical activity" as a physiological measure of the functional state of muscle tissue. Am J Phys Med 47: 10–22

Freund HJ, Budingen HJ, Dietz V (1975) Activity of single motor units from human forearm muscles during voluntary isometric contractions. J Neurophysiol 38: 933–946

Gillies JD (1972) Motor unit discharge patterns during isometric contraction in man. J Physiol 223: 36P-37P

Gordon AM, Huxley AF, Julien FJ (1964) The length-tension diagram of single vertebrate striated muscle fibers. J Physiol 171: 28P-30P

Grillner S, Udo M (1971) Recruitment in the tonic stretch reflex. Acta Physiol Scand 81: 571–573

Inman VT, Ralston HJ, Saunders JB de CM, Feinstein B, Wright EW Jr (1952) Relation of human electromyogram to muscular tension. EEG Clin Neurophysiol 4: 187–194

Kernell D (1966) The repetitive discharge of motoneurones in muscular afferents and motor control. In: Granit R (ed) Nobel symposium. Almquist and Wiksell, Stockholm, pp 351–362

Komi PV, Viitasalo JHT (1976) Signal characteristics of EMG at different levels of muscle tension. Acta Physiol Scand 96: 267–376

Kuroda E, Klissouras V, Milsum JH (1970) Electrical and metabolic activities and fatigue in human isometric contraction. J Appl Physiol 29: 358–367

Lind AR, Petrofsky JS (1979) Amplitude of the surface electromyogram during fatiguing isometric contractions. Muscle Nerve 2: 257–264

Lippold OCJ (1952) The relation between integrated action potentials in a human muscle and its isometric tension. J Physiol 117: 492–499

Mac Phedran AM, Wuerker RB, Henneman E (1965) Properties of motor units in a homogenous red muscle (soleus) of the cat. J Neurophysiol 28: 71–84

Maton B (1973) Influence du temps d'intégration sur la relation entre l'EMG de surface intégré et la force au cours de la contraction isométrique, isotonique. Electromyogr Clin Neurophysiol 13: 307–318

Maton B (1977) Fréquence et recrutement des unités motrices du muscle biceps brachial au cours du travail statique chez l'homme normal. J Physiol (Paris) 73: 177–199

Maton B, Bouisset S, Métral S (1969) Vomparaison des activités électromyographiques globale et élémentaire au cours de la contraction volontaire. Electromyography 9: 311–323

Métral S, Lemaire C, Monod H (1974) Force length integrated EMG relationships for submaximal isometric contractions. In: The control of upper extremity of prothesis and orthesis. CC Thomas, Springfield, IL, pp 13–22

Milner-Brown HS, Stein RB (1975) The relation between the surface electromyogram and muscular force. J Physiol 246: 549–570

Milner-Brown HS, Stein RB, Yemm R (1973a). The orderly recruitment of human motor units during voluntary isometric contraction. J Physiol 230: 359–370

Milner-Brown HS, Stein RB, Yemm R (1973b) Changes in firing rate of human motor units during linearly changing voluntary contractions. J Physiol 230: 371–391

Moritani T, De Vries HA (1978) Reexamination of the relationship between the surface integrated electromyogram (IEMG) and force of isometric contraction. Am J Phys Med 57: 263–277

Olson CB, Carpenter DO, Henneman E (1968) Orderly recruitment of muscle action potentials. Arch Neurol 19: 591–597

Pertuzon E (1971) Relation force-longueur isométrique du muscle humain en contraction maximale. CR Soc Biol 165: 328

Pertuzon E (1972) La contraction musculaire dans le mouvement volontaire maximal. Thèse Doctorat d'Etat, Lille, p 210

Petrofsky JS, Lind AR (1978) Metabolic, cardiovascular, and respiratory factors in the development of fatique in lifting tasks. Am Physiol Soc 45: 64–68

Rack PMH, Westbury DR (1969) The effects of length and stimulus rate on tension in the isometric cat soleus muscle. J Physiol 204: 443–460

Scherrer J, Bourguignon A (1959) Changes in electromyogram produced by fatigue in man. Am J Phys Med 38: 170–180

Scherrer J, Monod H (1960) Le travail musculaire local et la fatigue chez l'homme. J Physiol (Paris) 52: 419–501

Stephens JA, Taylor A (1972) Fatigue and maintained voluntary muscle contraction in man. J Physiol 220: 1–18

Tanji J, Kato M (1973a) Firing rate of individual motor units in voluntary contraction of abductor digiti minimi muscle in man. Exp Neurol 40: 771–784

Tanji J, Kato M (1973b) Recruitment of motor units in voluntary contraction of a finger muscle in man. Exp Neurol 40: 750–770

Tournay A, Paillard J (1953) Electromyographie des muscles radiaux à l'état normal. Rev Neurol 89: 277–279

Vredenbregt J, Rau G (1973) Surface electromyography in relation to force, muscle length and endurance. In: Desmedt JE (ed) New developments in electromyography and clinical neurophysiology, vol 1. Karger, Basel, pp 607–622

Zuniga EN, Simons DG (1969) Nonlinear relationship between averaged EMG potential and muscle tension in normal subjects. Arch Phys Med Rehab 50: 613–620