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Evaluation of handgrip force from EMG measurements

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A series of experiments were performed in order to investigate whether estimates of handgrip force could be derived with fair accuracy from surface EMG levels recorded on the finger flexors of the forearm, taking into account the position of the wrist in the flexion–extension plane and in the ulnar–radial deviation plane. Handgrip forces (on a JAMAR dynamometer) and corresponding surface EMG levels (on the finger flexors of the forearm) were recorded for 20 subjects in 11 postures of the wrist and for 30% and 70% of the MVC in neutral posture. A mathematical empirical model was developed using multiple non-linear regression analysis. Although quite simple, it provides very reliable results, the correlation coefficient between predicted and observed forces being 0.895. Its use must, however, be restricted to work situations where: (a) the hand efforts are of the same type and involve the same muscles as those exerted on the dynamometer; (b) the hand is in neutral pro-supination; and (c) no voluntary effort is exerted by the wrist flexors except for maintaining the wrist posture.

Keywords: EMG, handforce, biomechanics

The carpal tunnel syndrome (CTS) is the consequence of the compression of the median nerve in the carpal tunnel, and consists of sensitive and motor disorders with (eventually) paralysis of the muscles innervated by this nerve. Many factors can cause this syndrome: organic or metabolic disorders (diabetes, hypothyroidism), particular physiological states (pregnancy, menopause), or occupational factors, such as extreme postures of the wrist, high grip forces and high repetitivity of movements (Armstrong and Chaffin, 1979; Silverstein *et al*, 1987).

The study of the contribution of these last occupational factors to the development of CTS requires their quantification at the workplace during work. Reliable techniques for measurement of the angles at the different joints during activities have become increasingly available in recent years (such as infrared video systems, and goniometers fixed near the axes of rotation of the joints). The evaluation of grip force, in contrast, still remains to be developed for practical use in the workplace. Three different methods have been investigated:

(1) direct evaluation using transducers placed on the hand (these transducers are at present still clumsy, and interfere with the work being performed);
(2) indirect evaluation of the grip force by the worker (Moneim, 1991);
(3) indirect evaluation from EMG recordings.

The present paper deals with this last procedure.

In order to be used routinely in field studies, the method of evaluating handgrip force must be simple, safe and confortable for the worker, and fairly accurate. It is therefore out of question in this context to use wire electrodes implanted in the muscles, as is done in the laboratory or for clinical research. One has to rely solely on surface electrodes attached to the forearm. This is, in theory, open to criticism, and its limitations will be discussed later in this paper. Nevertheless, this technique has a series of advantages (Basmajian and De Luca, 1985), and is frequently used in practice (Grieco *et al*, 1989; Moore *et al*, 1991; Sundelin and Hagberg, 1992; Westgaard, 1988). The development of small lightweight EMG recorders (Baskin *et al*, 1987), such as the one used in the present study, will contribute to the increasing use of this method by occupational physicians in the field.

The basic reference in the field of ergonomics for the procedure of deriving estimates of forces from EMG recordings is the paper by Armstrong *et al* (1979). This technique has since been used by several researchers (Armstrong and Chaffin, 1979; Silverstein *et al*, 1986; Loslever *et al*, 1992; Aptel *et al*, 1993). In the initial procedure, the force from the integrated EMG is estimated using curves drawn for several hand positions and for each worker individually.

The aim of the present study was to investigate whether a mathematical model could be developed that could be used simply in the field to estimate handgrip force with reasonable accuracy from the electrical activity recorded by means of surface electrodes attained to the skin over the flexor muscles of the forearm. Calibration would be done in one neutral position only for each individual. The model would also take into account the angles adopted simultaneously by the wrist in the two planes of movement: flexor–extension and ulnar–radial deviation.

Materials and method

The research included simultaneous measurements of grip forces and EMG for different angles of the wrist.

Forces were measured using a hydraulic dynamometer (JAMAR type PC 5030G1) with a scale of 0–90 kg, modified in order to allow digital reading of the force and the recording of the signal on an FM recorder. Calibration was performed before and checked after the series of experiments (no modification was observed). Following Mathiowetz *et al* (1984), it was decided to perform all the tests with a spacing of 4.5 cm between the two handles of the dynamometer. *Figure 1* shows the dynamometer and the position used for the hand. During each test, the subject was asked to increase the force progressively over 1 s until the desired value was reached, and to maintain this force continuously during 4–5 s before the effort was progressively released. Each effort of less than 50% of maximum voluntary contraction (MVC) was followed by a rest period of at least 60 s at the minimum, while a rest period of at least 2 min was allowed after any effort greater than 50% of MVC.

The preparation of the skin and the location of the different electrodes were performed according to the recommendations of Basmajian and De Luca (1985).

The EMG was recorded using silver/silver chloride monopolar surface electrodes (Medicotest N-00-S). These surface electrodes were positioned over the flexor carpi radialis lying superficially above the flexor digitorum superficialis. They were located exactly in the region halfway between the motor point and the proximal insertion. The anatomical localization of the flexor carpi radialis was done by feel as the subject put them in isometric tension. The ground electrode was located on the medial epicondylitis (*Figure 1*).

The EMG signal was recorded on an ambulatory MEGA ME 3000 (Mega Electronics Ltd, Kuopio, Finland), which computed the RMS value in µv and stored it in a digital memory at a sampling rate of 10 Hz (Baskin *et al*, 1987). The frequency bandwidth of this instrument reached from 20 to 600 Hz.

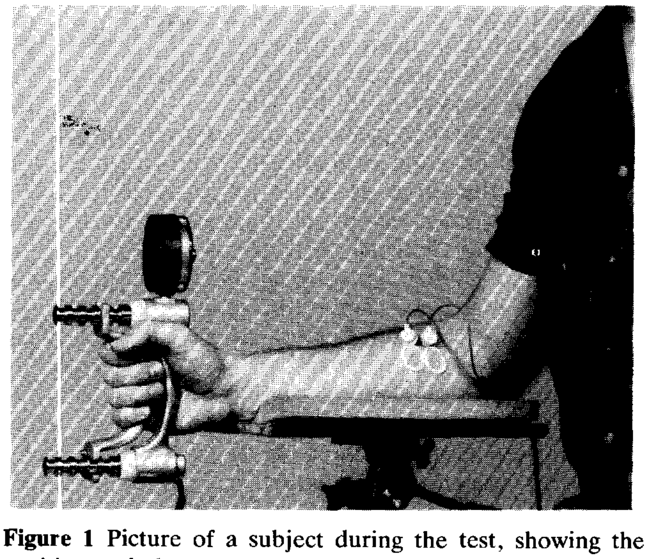

Figure 1 Picture of a subject during the test, showing the positions of the arm, forearm and wrist, as well as the dynamometer.

The angles were recorded using a Penny and Giles goniometer (Blackwood Ltd, UK), which consists of two strain gauges held along the wrist using two plastic pieces, one attached to the hand and the other to the forearm. The force and angle signals were recorded continuously on magnetic tape (FM cassette recorder TEAC HR-30G) for later treatment and analysis by computer. They were both sampled 80 times per second, then calibrated using previously determined calibration curves. Relative values of force and EMG were calculated on the basis of the maximum voluntary force and the corresponding integrated EMG observed in neutral posture of the wrist. Similarly, the relative values of angles were calculated using the maximum angle values recorded at the beginning of the experiment in full voluntary flexion, extension and deviations, the subject simply holding the dynamometer without exerting any force.

Measurements were made with the subject sitting, the forearm in a horizontal posture and in neutral supination, the arm between 10 and 20° in abduction and in external rotation of 0–15°, and therefore with the elbow angle at 90° (*Figure 1*). These postures were adopted in order to simulate the posture generally adopted at workplaces in industry, as it has been shown that the posture of the arm can influence the hand force (Mathiowetz *et al*, 1985).

The research began with a pilot study to test the reliability of the hardware used and to determine the type of relationship existing between the EMG and the forces exerted in three positions of the wrist. This pilot study was conducted with five male subjects from 28 to 47 years old, voluntarily enlisted, right-handed, in good health and without pathological past records at the level of the wrist. Each experiment for each subject began by simultaneous recordings of the hand force and the corresponding surface-integrated EMG level during MVC on the JAMAR dynamometer, with the wrist in neutral position. This test was repeated three times and the average values for these two parameters were used as reference maximum values throughout the experiment.

The pilot study consisted of recording the EMG levels during efforts of 10, 20, 30 etc to 100% of the MVC with the hand griping the dynamometer in neutral posture, in full voluntary flexion and in full voluntary extension, the neutral position being maintained in the ulnar–radial plane of movement. The angles of the wrist were displayed in front of the subject, who was invited to maintain the predetermined angle values. Similarly, the force exerted was displayed and the subject was invited to maintain the desired level. The order of presentation of the different levels of force in the different positions of the wrist was determined at random.

The main study was performed with 20 other male subjects, voluntarily enlisted, in good health and without musculoskeletal past history. The age varied between 18 and 39 years, the weight between 64 and 90 kg and the height between 1.70 and 1.96 m; 18 of them were right-handed but the tests were always performed on the right hand.

The sequence of efforts in the main study was as follows: three maximum voluntary efforts, followed by three psychophysical tests in which each subject was asked to exert a light, a medium and a heavy force; finally, the random sequence of tests, made at two relative levels of force (30% and 70% of MVC) in 11 postures of the wrist, as defined in *Table 1*.

The data from the main study (forces, angles and corresponding EMG) were statistically analysed using a multiple regression analysis: the dependent variable was the relative force and the independent variables were the relative EMG as well as the relative angles in flexion, extension and ulnar–radial deviation. The model was developed using the average of the relative values for the 20 subjects for each combination of the 11 postures of the wrist and the two force levels. It was then validated using the data from the pilot study and from the psychophysical tests of the main study. As the model is intended to be used in the field to study a particular worker situation, individual rather than average values were used in this validation study.

Results

Pilot study

Figure 2 presents the results of the pilot study (means ±1 standard error). It gives, for the three postures of the wrists and in average for the five subjects, the relative grip forces actively exerted (in ordinate) as a function of the forces that the subjects were asked to exert. *Figure 3* gives similarly the relative values of the corresponding integrated EMG levels. These two figures show that, for the neutral posture as well as in full voluntary extension, both the force and electrical muscular activity increase continuously as the level of the expected relative force increases. In context, for the wrist in full voluntary flexion, the muscular activity and the grip forces reach a ceiling value for about 60% of the MVC.

The regression analysis between the relative values of EMG and forces suggests a linear relationship between the natural logarithms of the two parameters for the three postures ($R = 0.997$, 0.977 and 0.981 in neutral posture, full extension and full flexion respectively). The relationships in extension and in neutral position are not significantly different. The electrical activity of the muscles during flexion is greatly increased for the same level of force.

Main study

For the 20 subjects participating in the main study, the MVC observed in neutral posture of the wrist ranged from 257 to 557 N with a mean value and a standard deviation of 407 ± 81 N.

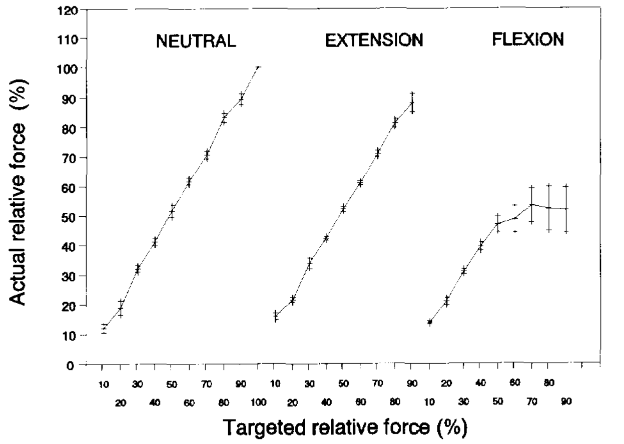

Figure 2 Influence of wrist posture on the grip force actually exerted for each level of force anticipated (means ±1 standard error)

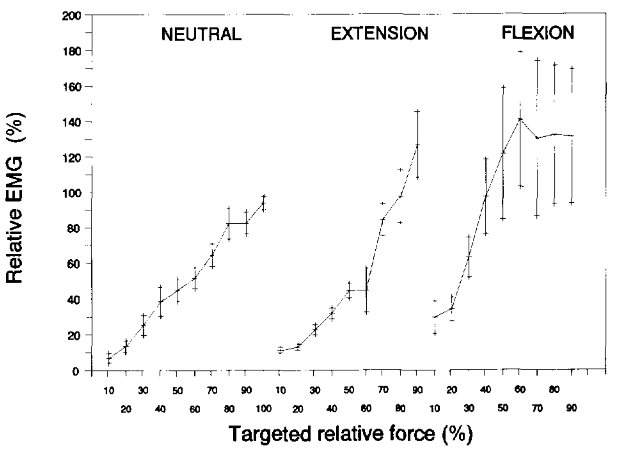

Figure 3 Influence of wrist posture on the integrated EMG level for each level of force anticipated (means ±1 standard error)

All subjects were able to maintain efforts of about 30% of their MVC in the 11 wrist postures. However, when asked to maintain 70% MVC, the real efforts were limited to 59% MVC on average in full voluntary flexion and 65% MVC in full extension and reached 70% MVC only in neutral posture. They were also reduced slightly with the wrist in full deviation.

Table 2 gives the results of the regression analysis conducted on the data of the main study: the coefficients of the final model, their standard error and their degree of significance. The multiple correlation coefficient is equal to 0.996. The final empirical model can be written

$$\ln(\text{Force\%}) = 1.13 + 0.714 \ln(\text{EMG\%}) + 3.19 \times 10^{-3}\,\text{Flex} + 0.83 \times 10^{-3}\,\text{Dev} - 26.9 \times 10^{-6}\,\text{Flex}^2 + 15.7 \times 10^{-6}\,\text{Flex} \cdot \text{Dev} \qquad (1)$$

where ln(Force%) is the natural logarithm of the relative force; ln(EMG%) is the natural logarithm of

Table 1 Angles adopted by the wrist during the test

	Flexion (−)/Extension (+)				
	−100%	−50%	Neutral	+50%	+100%
Deviation					
Neutral	X	X	X	X	X
Radial	X		X		X
Ulnar	X		X		X

Table 2 **Model for the prediction of the relative force as a function of relative EMG and relative angles of the wrist (dependent variable = ln (Force%))**

Independent variable	Coefficient	Standard error	Significance, p
Constant	1.13	0.09	< 0.0001
ln (EMG%)	0.714	0.02	< 0.0001
Flex	3.19×10^{-3}	0.21×10^{-3}	< 0.0001
Dev	0.83×10^{-3}	0.25×10^{-3}	< 0.005
Flex·Flex	-26.9×10^{-6}	4.8×10^{-6}	< 0.0001
Flex·Dev	15.7×10^{-6}	4.5×10^{-6}	< 0.005

where ln(Force %) = natural logarithm of the relative force; ln(EMG %) = natural logarithm of relative EMG; Flex = relative angle in extension–flexion (in %); Dev = relative angle in ulnar–radial deviation (in %).

relative EMG; Flex is the relative angle in extension–flexion (%); Dev is the relative angle in ulnar–radial deviation (%).

Figure 4 illustrates the results of the model as a function of the angle in the flexion–extension plane and in neutral posture for the ulnar–radial deviation. The 'Flexion' curve relates to the case with the wrist in 100% flexion (therefore Flex = −100%), while the 'Extension' curve relates to the full voluntary extension at +100%. This figure shows clearly the influence of flexion: with a relative force of 30% of MVC, the relative EMG level reaches about 24% in neutral posture, 22% in full extension but 56% in full flexion. For a force of 70% MVC, the EMG reaches 79% in neutral posture and 73% in full extension. In full flexion, however, it should correspond to 179%, according to the model. However, as shown during the pilot study, the force (*Figure 2*) as well as the EMG (*Figure 3*) tend to reach maximum values when the wrist is in full flexion, and the subjects are unable to exceed levels of force greater than 52% of MVC. The relative EMG reaches 100% for a force equal to 46% of the MVC in flexion instead of 84 and 87% in neutral posture and in extension respectively. In theory, these last two values should be about 100%.

Figure 5 depicts the same information for the maximum ulnar and radial deviations, the posture of the wrist remaining neutral in the flexion–extension plane. The 'radial' curve relates to the case where the wrist is in extreme radial posture (deviation = +100%) and the 'ulnar' curve relates to the opposite position (deviation = −100%). This figure clearly shows the lower influence of the deviation angle.

Validation of the model was conducted using the data not yet used from the pilot study and the main study. The model was used to derive the relative force in each case, and the regression line between the relative forces observed ($Force_{ob}$) and the relative forces predicted by the model ($Force_p$) was computed. The regression equation is

$$Force_{ob} = 5.66 + 0.901 \cdot Force_p \qquad (2)$$

The correlation coefficient was 0.895.

Figure 6 shows the relationship between these two sets of values. The slope of the linear relation is

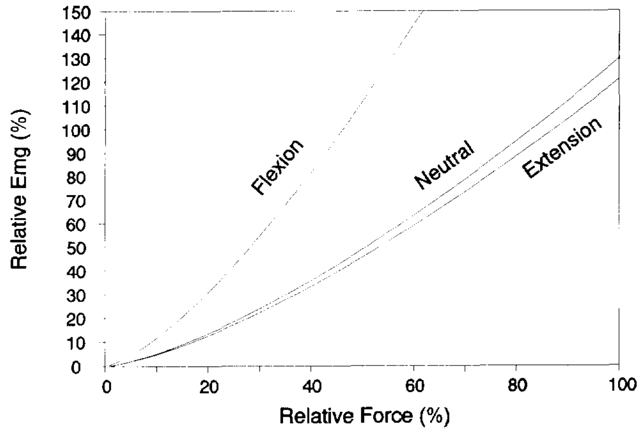

Figure 4 Relationship provided by the empirical model between relative EMG and relative grip force, the three wrist postures being neutral, full flexion and full extension (neutral posture in ulnar–radial deviation)

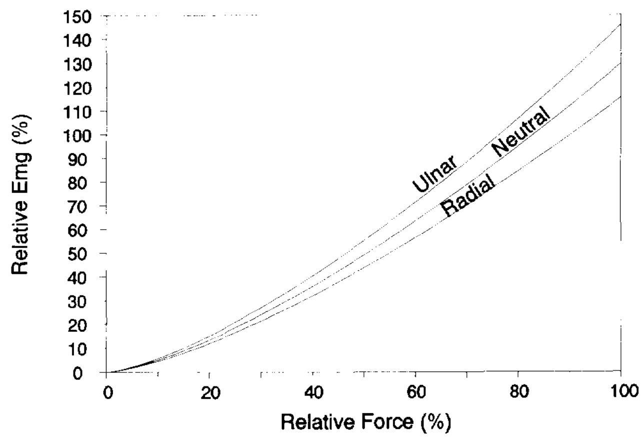

Figure 5 Relationship provided by the empirical model between relative EMG and relative force, the wrist being in neutral posture or in radial or ulnar maximum deviation (neutral posture in flexion–extension)

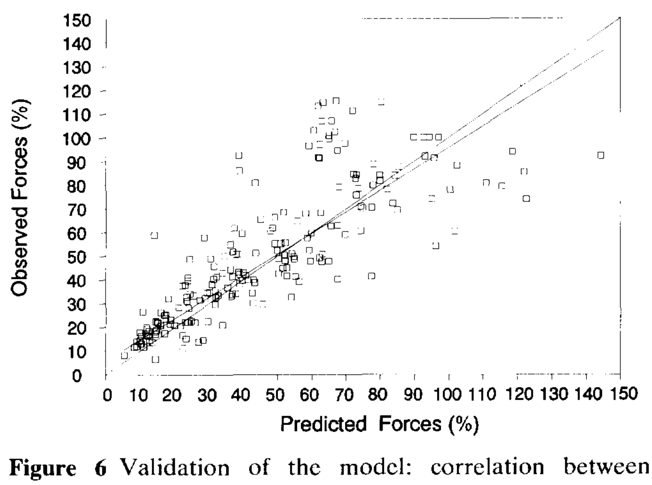

Figure 6 Validation of the model: correlation between observed relative forces and relative values predicted by the model

significantly different from 1 and the intercept is significantly different from the null value.

Discussion

The present study was aimed at developing a model making it possible to quantify the relationship between the electrical activity of the muscle and grip force and taking into account the posture adopted simultaneously by the wrist in the two main planes of movement: flexion–extension and ulnar–radial deviation.

The final model obtained from the mean results from the main study, although remaining relatively simple, is statistically highly significant ($R = 0.996$). In the validation step, the relationship between predicted and observed values is also highly significant. Attention must be drawn to the fact that, while the model was derived from data collected for 11 wrist postures and only two levels of force (30% and 70% of the MVC), the validation was carried out using, among others, individual data for 10 levels of relative force. As shown by *Figure 6*, the bisectrix line provides a good approximation of the low and average levels (lower than 60%) that are most frequently met, and which are therefore where the model could be particularly interesting in practice.

Although the results are very encouraging, they must be used with care. The model uses the relative angles of the wrist flexion, extension, ulnar and radial deviations. These relative angles are calculated as the ratios between the observed values and the maximum values, with the hand grasping the handle of the dynamometer. These maximum values are smaller than those expected with the hand flexed but the fingers extended.

However, the main restrictions concern the method of EMG recordings. Indeed, as surface electrodes must be used instead of fine wire electrodes, the actual muscle tissues whose electrical activity is monitored change with each posture as the skin slides over the muscles. However, this happens mainly as a function of the orientation of the hand in pronation and supination. As a neutral position was maintained in our experiments, this sliding is quite small and can affect the validity of the mathematical model only in a limited way. However, the model would provide questionable results in practical working conditions, where this orientation would vary constantly.

In addition to this, the posture of the wrist, especially in the flexion–extension plane, alters the length of the muscles, therefore modifying the length–strength relationship and the force–EMG relationship. This is clearly taken into account by the model as shown in *Figures 2–4*: in full flexion mainly, the subjects were not able to exert a force above 52% of the MVC, and for a given force the integrated EMG level varied greatly for the different wrist postures. This is consistent with the findings of other authors (Hazelton *et al*, 1975; Pryce, 1980), and does not constitute a limitation of the empirical model, which takes into account the non-linear role of the wrist angles.

It can be further argued that the changing of posture angles for the wrist leads to different cocontractions of the antagonistic extensor muscles and, most of all, of the flexor carpi radialis. Again, this might be considered as being included in the empirical model as long as no voluntary effort is required from these muscles except for maintaining the posture. This, however, constitutes the main limitation of the model: in many working conditions, flexion of the fingers is accompanied by wrist flexion. As the wrist flexion muscles are more superficial, their contribution to the EMG may become predominant. The integrated EMG level can then be larger than the reference value and lead to an over-estimation of the force exerted by the finger flexor muscles. It must be noted that, in the context of ergonomics, this error leads to an overestimation of the risk of musculoskeletal disorders and therefore constitutes a safety factor. However, the result is in those cases an unreliable estimate of the hand force.

This raises the question of the calibration of the force and EMG. The procedure uses as reference values only the maximum voluntary force and the corresponding EMG RMS value (EMG_{max}) recorded with the hand in neutral position. The calibration in any posture is done for each parameter by simply dividing the observed values by the reference values, and therefore negelects the EMG levels at rest (EMG_{min}), when the hand is simply holding the dynamometer.

This appears to be valid with the hand in neutral posture, as independent tests showed that EMG_{min} in this case is of the order of 2% of EMG_{max}. It would still hold with the hand in full voluntary extension, as EMG_{min} remains less than about 5% of EMG_{max}. However, in full voluntary flexion, EMG_{min} appeared to increase drastically to about 30% of EMG_{max} due to the activity of the flexor carpi radialis needed to maintain this posture.

As shown in *Figure 3*, the EMG maximum level also increases greatly in full voluntary extension and flexion with the hand exerting a maximum grip force. Therefore, calibration of the EMG should ideally be made using the following expression (Mirka, 1991):

$$\%\ EMG_p = \frac{EMG_p - EMG_{min,p}}{EMG_{max,p} - EMG_{min,p}} \tag{3}$$

where $EMG_{max,p}$ and $EMG_{min,p}$ would be the maximum and minimum RMS levels of the EMG recorded in a given posture (p), with respectively a maximum and no handgrip force.

This calibration for several postures would not be satisfactory, as it would consume time and effort, and would be valid only for these postures and not for the intermediate postures most often encountered in the workplace. It can additionally be debated whether this constitutes a real limitation of the empirical model, which does not aim to depict the physiological or biomechanical behaviour of the hand–forearm system, but simply to provide an order-of-magnitude estimate of the force exerted. This aim appears to be reached satisfactorily as the correlation coefficient obtained during the validation study is unexpectedly very high (0.895), although based on the comparison between predicted and observed individual force values.

The validity of the empirical model therefore appears to be limited to conditions with the hand in neutral pro-supination, with no activity of the wrist flexor muscles other than for maintaining the posture of the wrist. One must add, however, that the model also assumes a constant temperature of the muscles involved and no fatigue, as these two factors can also influence the amplitudes of the EMG signal for a given handgrip force (Petrofsky, 1979). It therefore appears reasonable to advise against using the model in conditions with high repetitivity of forces.

Finally, the use of the model cannot be extended to any type of manual handling, but is specific to the types of effort that were simulated using the JAMAR dynamometer. For other types of grip, such as the pinch grip, different muscular groups are involved, and the model should not be used: the relative force derived from the model could be a very large underestimation of the real efforts exerted by these other muscles, and therefore would provide a very false picture of the severity of these efforts. Similar models should then be established for these other types of grip.

Conclusions

The study was undertaken to verify whether an empirical mathematical model could be derived to predict the relative force exerted during a handgrip as a function of the integrated EMG level and the angles of the wrist. This can be answered positively, as a rather simple mathematical formula was found to predict the observed force with great accuracy (correlation coefficient $R = 0.895$). The main limitations have been discussed. This model concerns only the type of handgrip forces that can be simulated with the particular dynamometer used, with no voluntary contraction of the wrist flexor muscles (in particular, the flexor carpi radialis) and while the hand is in neutral position in pro-supination.

When using heart rate to estimate the metabolic rate of a worker at a workplace, the user must therefore keep in mind these limitations and be very cautious about the use of the results. However, at a time where ambulatory surface EMG recorders are becoming more available, it is hoped that the use of this mathematical model based on a single reference value in force and EMG can reduce the number of misinterpretations in the field. The model is therefore proposed as one of the methods of gaining some insight into the force requirements of a working condition.

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