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Relevance of the EMG/Grip Relationship in Isometric Anisotonic Conditions

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The aim of the present study was to develop a relationship to evaluate grip force using the electromyogram (EMG) in isometric anisotonic conditions.

The EMGs of the flexor digitorum superficialis (FDS) and the extensor digitorum (ED) were recorded in 3 flexion-extension positions of the wrist (30° flexion, 30° extension, and 60° extension) associated with 3 positions of the forearm (70° pronation, prono-supination, and 70° supination). For each position, the participants had to follow linear ramp targets (2 rates of increase and decrease) displayed on an oscilloscope.

The results show the best fit is a quadratic type force-EMG relationship. Some aspects such as the rate of force variation and the forearm and wrist positions are then discussed along with the limitations of the relationship.

cumulative trauma disorders upper limb electromyogram force isometric contraction anisotonic contraction

1. INTRODUCTION

Musculoskeletal disorders (MSDs) have been on the increase since the beginning of the 1980s, (Ayoub & Wittels, 1989; Caisse Nationale de l'Assurance Maladie des Travailleurs Salariés, 1998). MSDs affect the joints (bursitis, synovitis), tendons (tendinitis, tenosynovitis), and nerves (compression syndromes). MSDs generally result from an imbalance between occupational biomechanical stresses (force, repeated movement, and extreme articular positions) or non-occupational stresses (sport, do-it-yourself, etc.) and individual

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functional capacities, which depend on physiological condition, pathological condition, gender, age, and lifestyle (Silverstein, Fine, & Armstrong, 1986).

Among the MSDs affecting the upper limb, carpal tunnel syndrome (CTS) occupies a predominant position on account of its incidence in certain sectors of activity (Silverstein, Fine, & Armstrong, 1987). Symptoms include pains, numbness, and tingling in the parts of the hand innervated by the median nerve. This involves compression of the nerve in the carpal tunnel, which perturbs intraneural microcirculation and leads to modification of the structure of the nerve fibre. This pathology often results from the repeated microtraumas undergone by the tendons and by the sheaths of the superficial and deep flexors subsequent to high levels of effort (microtearing of collagen fibres), extreme articular positions (compression of the tendons against the carpal bone during extension and against the flexor retinaculum during flexion), and repeated movements (repeated rubbing of the tendons and their sheath against adjacent structures). In this regard, Silverstein et al. (1986) showed that high levels of effort associated with repetitive movements lead to a risk of CTS five times higher than when these constraints are taken into account separately.

The onset of CTS in an occupational context leads to the assumption putting forward the hypothesis that the stresses exerted at the work station are greater than the individual functional capacities; they stem from repeated microtraumatisms of the tendinous structures, which become centres of inflammatory and degenerative symptoms and then result in the compression of the median nerve in the carpal tunnel.

Thus, it appeared necessary firstly to quantify these biomechanical stresses (articular angles, efforts, and movements) at the work station without modification of occupational gestures. The articular positions of the wrist can be measured directly using goniometers fixed to the wrist of an employee. From the articular position of the wrist, it is possible to quantify the repetitiveness of movements by means of the derivative of the signal delivered by the goniometers, each change of sign of this derivative being interpreted as a wrist movement. In contrast, the force is not directly accessible without hindering the occupational gesture, and it is therefore necessary to use an indirect means, namely the integrated surface electromyogram, which is proportional to the force developed by the muscle in certain conditions (see, e.g., Armstrong, Chaffin, & Foulke, 1979).

Relationships allowing the assessment of grip force by means of integrated EMG have already been described in the literature (Duque, Masset, & Malchaire, 1995; Meyer, Didry, Herrault, & Horwat, 1994; Ranaivosoa, 1992). However, these relationships, established in isometric isotonic conditions, are rarely observed at the work station. Indeed, occupational gestures are very often associated with changes of the level of force and with movements of the upper limb. The simultaneous variation in the length of the muscle and the amplitude of the force developed by it has the major disadvantage that the observation of electrical phenomena cannot be attributed to one or the other of these factors. It was for this reason that the study described in this paper investigated only the variation in force, muscle length being kept constant. In practice, this type of situation is mainly found during the use of certain hand-held tools such as power screwdrivers and drills, where the articular position of the upper limb remains fixed but where the buttressing force of the handle of the tools varies as the operation is being carried out.

Thus, the aim of the study was to establish a relationship intended to assess the grip force of the flexor and extensor muscles of the fingers in isometric anisotonic conditions by means of the integrated electromyogram (EMG).

2. METHOD

The study was carried out on 16 right-handed participants (8 men and 8 women). The participants had previously been informed of the content and aims of the experimentation, which was authorised by the French Ministry of Health following the favourable decision reached by the Consultative Committee for the Protection of Individuals in Biomedical Research. All the participants practised sports, were in good health, and free of musculoskeletal disorders of the upper limb. The average age (*SD*) of the participants was 19.9 years (1.7), the average height 170 cm (5.5), and the average weight 64.2 kg (8.4).

The participant sat on a chair that could be adapted to individual anthropometric characteristics (the height and depth of the seat as well as the tilt of the back were adjustable). The trunk was firmly strapped to the back of the seat to prevent modification of posture during the grip force measurements. The elbow was flexed to 90°, and the right forearm placed in a supporting armrest. The right wrist was equipped with a goniometer (Penny-Giles®, Biometrics, UK) allowing measurement of the angles of flexion-extension. The pronation-supination angles were recorded with a torsiometer (Penny-Giles®) placed on the right forearm. The handle used to measure the grip force was equipped with a force sensor (Entran®, Entran, France). The gap between the two sections of the handle could be set to the anthropometric dimensions of the hand of the participants. This setup allowed different fixed positions to be imposed on the forearm and on the wrist (Figure 1).

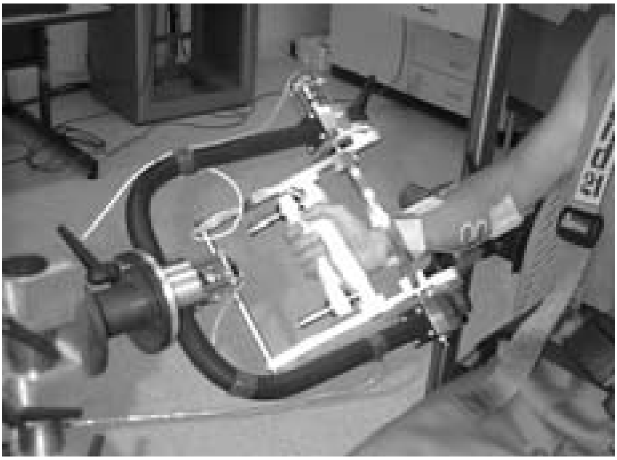

Figure 1. Experimental setup.

Three articular positions of the wrist were retained for this experimentation: 30° flexion, 30° extension, and 60° extension. For each flexion-extension position, the forearm was either 70° pronation, 70° supination, or in the prono-supination position. Thus, a total of 9 hand positions (3 flexion-extension positions of the wrist and 3 pronation-supination positions of the forearm) were imposed on the participant.

For each articular position of the wrist and forearm, each participant had firstly to develop a maximal voluntary contraction (MVC), then exert continuous grip forces varying linearly from 0 to 70%, then from 70 to 0% of the MVC at the given position according to two rates of increase and decrease, namely 10% of the MVC·s^{-1} and 35% of the MVC·s^{-1} (Figure 2).

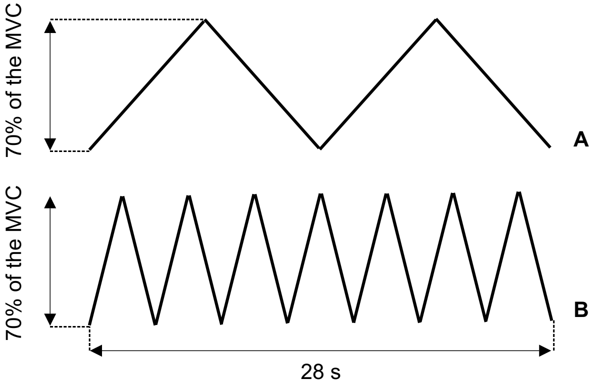

Figure 2. Linear ramp targets. *Notes.* MVC—maximal voluntary contraction, A—rate of 10% of the MVC·s^{-1}, B—rate of 35% of the MVC·s^{-1}.

An electronic device displayed the signal delivered by the force sensor as well as the target values corresponding to the effort required. Each participant was required to superimpose the force sensor signal on that of the target value. The participants were allowed a short period of practice. As was the case in the maximal voluntary contraction/maximum voluntary force tests, each experimental condition was separated by a 3-min break to prevent the onset of muscular fatigue.

The maximal voluntary contractions were systematically repeated twice. The duration of maintaining the MVC was 2 s. Only the higher MVC of the two was retained.

For the forces varying linearly from 0 to 70%, then from 70 to 0% of the MVC, the duration of each experiment was 30 s. This duration allowed for the recording of 2 increasing forces and 2 decreasing forces when the rate of variation was 10% of MVC·s^{-1}, and 7 increasing forces and 7 decreasing forces when the variation rate was 35% of MVC·s^{-1}.

After appropriate preparation of the skin, the surface EMGs of the flexor digitorum superficialis (FDS) and the extensor digitorum (ED) muscles were recorded by means of surface electrodes (Blue Sensor®, Medicotest, Denmark) placed on the forearm about one quarter of the elbow-wrist distance from the elbow (Zipp, 1982). The distance separating the centre of the electrodes

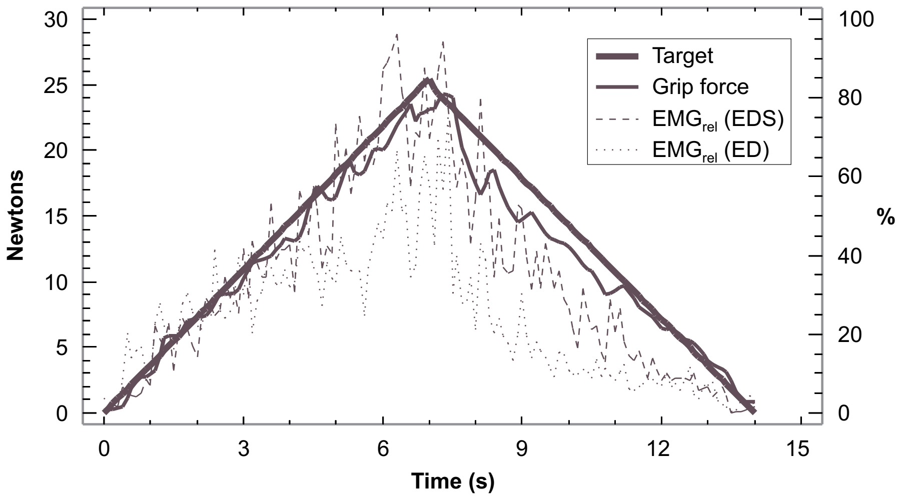

Figure 3. An example of the force sensor signal superimposed on the target signal (left axis) and the integrated EMGs of the flexor digitorum superficialis and the extensor digitorum (right axis). *Notes.* EMG—electromyogram, FDS—flexor digitorum superficialis, ED—extensor digitorum.

was 2 cm. The interelectrode resistance was less than 5 kΩ. The EMG was amplified, rectified, then integrated in periods of 100 ms. An example of the force sensor signal superimposed on the target signal and the integrated EMGs of the flexor digitorum superficialis (FDS) and the extensor digitorum (ED) is shown in Figure 3.

3. RESULTS

3.1. Maximum Voluntary Contractions (MVC)

The highest MVC was obtained with the wrist at medium extension (≈30°) and the forearm in prono-supination. This position was therefore chosen as the reference position for standardisation of the signals. Thus, all the force and EMG values are expressed as a percentage of the values recorded in this reference position according to the procedure proposed by Mirka and Zipp (1994). In the remainder of this paper, the relative grip force and EMG values are termed $force_{rel}$ and EMG_{rel} respectively.

In addition, the MVC was statistically ($p < .001$) higher when the wrist was at 60° extension than when it was at 30° flexion.

Likewise, the MVC was statistically ($p < .001$) higher when the forearm was in supination than when it was in pronation. In contrast, maximal voluntary contraction was not statistically different between the prono-supination and supination positions.

3.2. Influence of the Position of the Hand (Flexion-Extension, Pronation-Supination) and of the Force Target Value (Direction, Rate) on the $force_{rel}$ and the EMG_{rel} of the FDS and ED Muscles

An analysis of variance was carried out on the independent variables $force_{rel}$, EMG_{rel} of the FDS muscle, and EMG_{rel} of the ED muscle, with the angles of wrist flexion-extension and forearm pronation-supination and the direction and rate of the force ramp as independent variables.

The flexion-extension of the wrist was a significant factor on $force_{rel}$ ($p < .001$) and on EMG_{rel} of the FDS muscle ($p < .001$) and on the EMG_{rel} of the ED muscle ($p < .001$). Compared to the reference position, $force_{rel}$ was slightly lower in extension and considerably lower in flexion. The mean amplitude of the EMG_{rel} of the FDS muscle was higher in flexion than in extension. Finally, the mean amplitude of the EMG_{rel} of the ED muscle was the same in extension and in flexion, but higher in the reference position. Likewise, the pronation-supination of the forearm presented a significant effect on the $force_{rel}$ ($p < .001$) and on the EMG_{rel} of the FDS muscle ($p < .005$). Compared to the reference position, the $force_{rel}$ was slightly lower in supination and considerably lower in pronation. Finally, the mean amplitude of the EMG_{rel} of the FDS muscle was lower in supination and in pronation compared to that obtained in the reference position.

The analysis of variance also showed that the effect of the direction of the force ramp (increase vs. decrease) on the mean $force_{rel}$ recorded during the experiments was significant ($p < .001$). Indeed, the mean values of the $force_{rel}$ and the EMG_{rel} of the FDS and ED muscles were higher during increase than during decrease for both rates. This result might be explained by the fact that the participants found it difficult to follow the target value during the decrease phase. Indeed, once the summit of the target value triangle was reached, the participants tended to release their effort brutally at the start of descent.

In contrast, the rate of force variation only exerted a significant action ($p < .001$) for the EMG_{rel} of the extensor digitorum. Indeed, the mean value of the EMG_{rel} of this muscle was lower when the force ramp rate was 10% of $MVC \cdot s^{-1}$ than when it was 35% of $MVC \cdot s^{-1}$ (Figure 3). However, although statistically significant, this difference remains small and, as a result, it can be concluded that, for the present experimental conditions, the rate and direction of the force variation can be ignored in the development of a relationship allowing evaluation of grip force from the EMG.

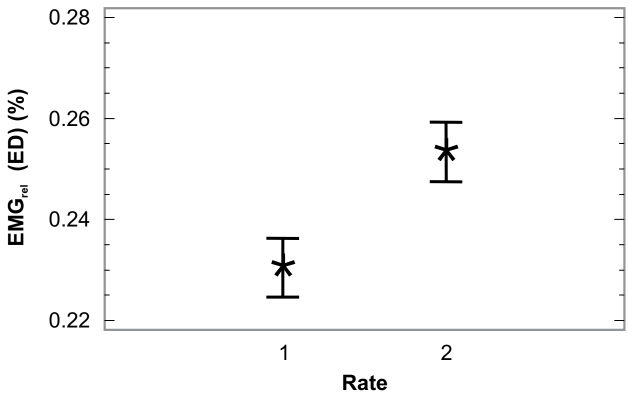

Figure 4. Effect of the rate of force on the mean value of EMG_{rel} (ED). *Notes.* EMG—electromyogram, ED—extensor digitorum, MVC—maximal voluntary contraction, Rate 1—10% of the $MVC \cdot s^{-1}$, Rate 2—35% of the $MVC \cdot s^{-1}$. The vertical lines represent the 95% confidence intervals.

3.3. Empirical Relationship to Predict Relative Grip Force

To establish a relationship allowing assessment of the force$_{rel}$ by means of the EMG$_{rel}$ of the FDS and ED muscles, the initial stage consisted in comparing different types of relationships between force$_{rel}$ and EMG$_{rel}$ (FDS) and between force$_{rel}$ and EMG$_{rel}$ (ED).

Different forms of relationship linking force$_{rel}$ to EMG$_{rel}$ are described in the literature. They concern linear and curvilinear relationships (exponential, power, logarithmic, square root, quadratic; Moritani & Devries, 1978). In the absence of a determining physiological argument in favour of one or another of these relationships, all were applied to the data recorded during the experimentation. Examination of Table 1 shows that a quadratic regression between force$_{rel}$ and EMG$_{rel}$ (FDS) has a correlation coefficient that is frequently higher than that obtained with the other types of relationships. It was therefore the quadratic relationship that was chosen to represent this force-EMG relationship.

TABLE 1. Correlation Coefficients Between Force$_{rel}$ and EMG$_{rel}$ (FDS) According to the 6 Types of Relationships Studied and the Wrist and Forearm Articular Deviations

	Position								
Relationship	**Ext. 60° & Sup.**	**Ext. 60° & Neu.**	**Ext. 60° & Pro.**	**Ext. 30° & Sup.**	**Ext. 30° & Neu.**	**Ext. 30° & Pro.**	**Flex. 60° & Sup.**	**Flex. 60° &Neu.**	**Flex. 60° & Pro.**
Linear model	0.83	0.86	0.85	0.85	0.83	0.79	0.65	0.70	0.70
Exponentinal model	0.73	0.75	0.74	0.75	0.73	0.70	0.58	0.67	–
Multiplicative model	0.86	0.87	0.85	0.88	0.84	0.83	**0.78**	**0.80**	0.76
Logarithmic model	0.80	0.82	0.81	0.80	0.81	0.78	0.74	0.73	0.73
Square root model	0.87	0.89	0.88	0.88	0.87	0.83	0.74	0.76	0.75
Quadratic model	**0.87**	**0.90**	**0.88**	**0.89**	**0.88**	**0.84**	0.72	0.76	**0.77**

Notes. Maximal values are in bold, – —model not available; Ext.—extension, Flex.—flexion, Sup.—supination, Pro.—pronation, Neu.— prono-supination.

A similar analysis was carried out between force$_{rel}$ and EMG$_{rel}$ (ED). The results of Table 2 show that the relationship systematically having the highest correlation coefficient is the quadratic type.

Based on the results presented in Tables 1 and 2, the relative grip force can be calculated by means of the following equation:

$$\text{force}_{rel} = \alpha \bullet \text{EMG}_{rel}(\text{FDS}) + \beta \bullet \text{EMG}_{rel}^{2}(\text{FDS}) + \delta \bullet \text{EMG}_{rel}(\text{ED}) + \lambda \bullet \text{EMG}_{rel}^{2}(\text{ED}) + \mu. \quad (1)$$

TABLE 2. Correlation Coefficients Between Force$_{rel}$ and EMG$_{rel}$ (ED) According to the 6 Types of Relationships Studied

	Position								
Relationship	Ext. 60° & Sup.	Ext. 60° & Neu.	Ext. 60° & Pro.	Ext. 30° & Sup.	Ext. 30° & Neu.	Ext. 30° & Pro.	Flex. 60° & Sup.	Flex. 60° &Neu.	Flex. 60° & Pro.
Linear model	0.77	0.76	0.76	0.79	0.75	0.86	0.76	0.72	0.80
Exponential model	0.66	0.67	0.68	0.69	0.65	0.72	0.68	0.65	–
Multiplicative model	0.71	0.67	0.73	0.78	0.72	0.76	0.76	0.71	0.78
Logarithmic model	0.72	0.66	0.70	0.76	0.72	0.76	0.72	0.67	0.74
Square root model	0.80	0.78	0.79	0.82	0.78	0.88	0.81	0.77	0.84
Quadratic model	**0.81**	**0.78**	**0.79**	**0.82**	**0.78**	**0.88**	**0.81**	**0.78**	**0.85**

Notes. Maximal values are in bold, – —model not available; Ext.—extension, Flex.—flexion, Sup.—supination, Pro.—pronation, Neu.— prono-supination.

Coefficients α, β, δ, and λ vary according to the flexion-extension position of the wrist and the pronation-supination position of the forearm. A multiple linear regression analysis allows the expression of coefficients α, β, δ, and λ according to the following equations:

$$\alpha = a_1 + b_1 \bullet \theta_{f\text{-}e} + c_1 \bullet \theta_{p\text{-}s},$$
$$\beta = a_2 + b_2 \bullet \theta_{f\text{-}e} + c_2 \bullet \theta_{p\text{-}s},$$
$$\delta = a_3 + b_3 \bullet \theta_{f\text{-}e} + c_3 \bullet \theta_{p\text{-}s},$$
$$\lambda = a_4 + b_4 \bullet \theta_{f\text{-}e} + c_4 \bullet \theta_{p\text{-}s},$$

where $\theta_{f\text{-}e}$ represents the angle of flexion-extension of the wrist and $\theta_{p\text{-}s}$ represents the angle of pronation-supination of the forearm.

Thus, Equation 1 is expressed as

$$\begin{aligned}\text{force}_{rel} = {} & a_1 \bullet \text{EMG}_{rel}(\text{FDS}) + b_1 \bullet \text{EMG}_{rel}(\text{FDS}) \bullet \theta_{f\text{-}e} + c_1 \bullet \text{EMG}_{rel}(\text{FDS}) \bullet \theta_{p\text{-}s} \\ & + a_2 \bullet \text{EMG}_{rel}^{2}(\text{FDS}) + b_2 \bullet \text{EMG}_{rel}^{2}(\text{FDS}) \bullet \theta_{f\text{-}e} + c_2 \bullet \text{EMG}_{rel}^{2}(\text{FDS}) \bullet \theta_{p\text{-}s} \\ & + a_3 \bullet \text{EMG}_{rel}(\text{ED}) + b_3 \bullet \text{EMG}_{rel}(\text{ED}) \bullet \theta_{f\text{-}e} + c_3 \bullet \text{EMG}_{rel}(\text{ED}) \bullet \theta_{p\text{-}s} \\ & + a_4 \bullet \text{EMG}_{rel}^{2}(\text{ED}) + b_4 \bullet \text{EMG}_{rel}^{2}(\text{ED}) \bullet \theta_{f\text{-}e} + c_4 \bullet \text{EMG}_{rel}^{2}(\text{ED}) \bullet \theta_{p\text{-}s} \\ & + \mu. \end{aligned} \quad (2)$$

The values of the different coefficients of the aforementioned equations, as well as the standard error, the Student *t* value and the degree of significance *p* are presented in Table 3.

TABLE 3. Coefficients of Polynomial Linear Regression

Independent Variable	Coefficient	*SE*	*t*	*p*
μ	0.039000	0.00200	20.230	.0000
a_1	0.440000	0.01000	42.340	.0000
b_1	0.007600	0.00030	26.460	.0000
c_1	0.000021	0.00030	0.064	.9488*
a_2	–0.220000	0.00900	–25.190	.0000
b_2	–0.005000	0.00026	–18.070	.0000
c_2	–0.000800	0.00020	–3.270	.0011
a_3	0.530000	0.01500	36.300	.0000
b_3	–0.001000	0.00040	–2.880	.0040
c_3	0.001000	0.00040	3.720	.0064
a_4	–0.270000	0.01600	–16.370	.0000
b_4	0.001000	0.00050	2.940	.0033
c_4	–0.001000	0.00050	–1.920	.0544

Notes. *—*ns*, $r = .91$, *SE*: $\sigma = 0.069$.

Figure 4 shows the relationship between the force$_{rel}$ recorded during the experimentation and the force$_{rel}$ obtained by means of Equation 2.

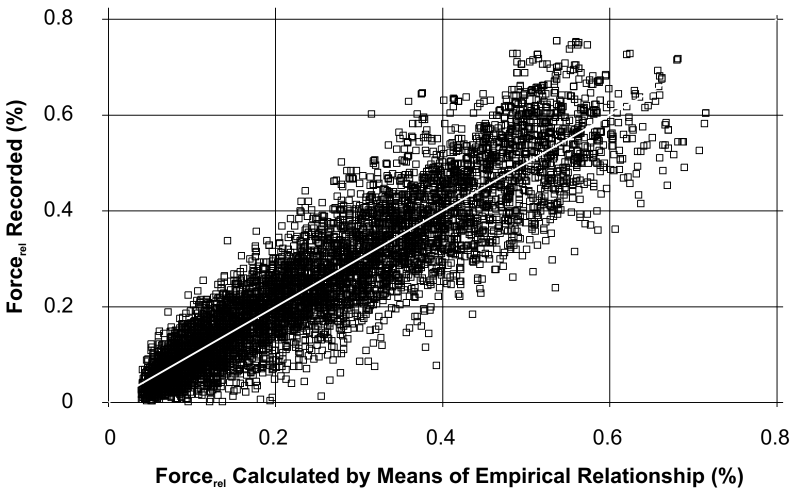

Figure 5. Relationship between the relative force (force$_{rel}$) recorded during the experimentation and the relative force calculated by means of the empirical relationship.

4. DISCUSSION

The aim of the present study was to establish a relationship allowing assessment of grip force by means of the integrated EMG of the muscles involved in gripping with a view to its use in certain ergonomic studies. To achieve this, two force ramp rates combined with nine postural conditions of the wrist and forearm were requested of the participants to construct the required relationship. This discussion therefore looks at the effects of "change of force rate" and "body segment position" factors on the force-EMG relationship as well as at the development and limits of the relationship obtained.

4.1. Effects of Change of Force Rate on the Signals Recorded

The results of previous work (Lawrence & De Luca, 1983; Métral & Cassar, 1986) show that, depending on the muscle studied, the EMGi-force relationship can either be independent of or, in contrast, depend on the change of force rate.

In this respect, Métral and Cassar (1986), during a study carried out on the extensor carpi radialis, found no change in the force-EMGi relationship for force ramp rates varying from 11 to 33% MVC$\cdot s^{-1}$. In contrast, for the deltoid, the biceps, and the first dorsal interosseous, Lawrence and De Luca (1983) showed a significant change ($p < .05$) in the shape of the EMGi-force relationship, principally for low levels of force and for force ramp rates varying from 10 to 40% MVC$\cdot s^{-1}$. Thus, for example, at 20% of the MVC, the amplitude of the EMGi signal is higher for a variation rate of 40% MVC$\cdot s^{-1}$ than for a rate of 20% MVC$\cdot s^{-1}$. Nevertheless, the general pattern of the EMGi-force relationships remains similar whatever the rate of change of force.

The results show that the change of force rate exerts a significant action only for the extensor digitorum. Although statistically significant, this action is very weak and was not taken into account when developing the relationship allowing assessment of grip force. From a physiological point of view, the origin of the difference observed regarding the amplitude of the EMG_{rel} (ED) signal, for the two force rates considered during this experimentation, is likely due to different motor unit (MU) recruitment processes. Indeed, for the extensor indicis, Büdingen and Freund (1976) showed that certain MUs activated at increasingly lower global force levels as the force ramp rate increased. In addition, Lawrence and De Luca (1983) had recourse to the hypothesis of a difference in MU recruitment process to explain the influence of variation in force on the EMGi-force relationships of certain muscular groups. In contrast, again according to Lawrence and De Luca (1983), the rate of force variation had a greater or lesser influence depending on the muscle studied. Thus, in the present study, it was possible to consider the influence of the rate of force variation (within the limit of the values studied) as negligible.

4.2. Effect of Articular Position on the Signals Recorded

The positions of the wrist and forearm also have a significant effect on the grip force developed and on the EMG of the muscles involved in gripping. The variation in grip force according to wrist flexion-extension position and forearm pronation-supination position is comparable to that described in the literature (Ranaivosoa, 1992; Terrel & Purswell, 1976).

Indeed, the grip force was lower when the wrist was at 60° extension or 30° flexion than medium extension (≈30°). In addition, compared to the reference position, the force_{rel} was slightly lower in supination and considerably lower in pronation. The reduction in force between medium extension and 60° flexion as well as between supination and pronation is primarily due to a variation in the length of the flexor digitorum superficialis muscle (Terrel & Purswell, 1976). In contrast, the reduction in MVC between medium extension and extreme extension would appear to be due to less efficient buttressing of the thenar and hypothenar eminence on the wrist (Terrel & Purswell, 1976).

As regards the differences in the amplitude of the EMG signals of the flexor digitorum superficialis and extensor digitorum muscles according to the position of the wrist and the forearm, the origin may be physiological, but the influence of a modification of the reception area of the electrodes due to movement of the electrodes in relation to the muscle cannot be ruled out.

4.3. Development, Accuracy, and Limits of the Empirical Relationship Allowing Evaluation of Grip Force

In the conditions of the present study (isometry, anisotony), the literature describes various types of non-linear relationships according to the muscle groups studied (Bouisset, Goubel & Maton, 1973; Lawrence & De Luca, 1983; Métral & Cassar, 1986; Moritani & Devries, 1978). The construction of the relationship proposed earlier has recourse to quadratic type force-EMG relationships. This type of relationship, widely described in the literature (Lawrence & De Luca, 1983; Moritani & Devries, 1978; Zuniga & Simons, 1969), allowed the best fit for all the $force_{rel}$-EMG_{rel} (ED) and $force_{rel}$-EMG_{rel} (FDS) relationships except for the latter when the wrist was at 60° flexion and the forearm in supination or in prono-supination. Indeed, for these positions, a power type relationship has a higher correlation coefficient than those obtained with the other relationships. For the “general” relationship, the correlation coefficient is .91 and the standard error of estimation 0.069. Figure 4 shows that the prediction of grip force is satisfactory, account taken of the sources of inter- and intraindividual variations.

The relationship was established in isometric anisotonic conditions for grip forces varying between 0 and 70% of the MVC and force ramp rates equal to 10% of the MVC and 35% of the MVC. This relationship must therefore be used in these conditions. It would however appear that it is even more robust for faster grip force ramp rates or for low rates of muscle length variation as long as the conditions remain isometric or quasi-isometric. This therefore allows its use in ergonomic studies.

5. CONCLUSION

This study has allowed the establishment of a relationship linking the integrated EMG of the FDS and ED muscles as well as the angles of wrist flexion-extension and forearm pronation-supination to grip force. This relationship, established in isometric anisotonic conditions, allows assessment of grip force with acceptable accuracy ($\sigma = 0.07$) within the framework of ergonomic studies intended to reduce the influence of the risk factors leading to the onset of MSDs. This is the case in particular during the use of the screwdriver or drill type hand-held tools frequently encountered in assembly operations. It should thus be possible, by means of EMG electrodes and goniometers, to evaluate the grip force of operators at their work stations with no appreciable hindrance to occupational gestures.

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