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The development and validation of equations to predict grip force in the workplace: contributions of muscle activity and posture

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The inherent difficulty of measuring forces on the hand in ergonomic workplace assessments has led to the need for equations to predict grip force. A family of equations was developed, and validated, for the prediction of grip force using forearm electromyography (six finger and wrist muscles) as well as posture of the wrist (flexed, neutral and extended) and forearm (pronated, neutral, supinated). Inclusion of muscle activity was necessary to explain over 85% of the grip force variance and was further improved with wrist posture but not forearm posture. Posture itself had little predictive power without muscle activity (<1%). Nominal wrist posture improved predictive power more than the measured wrist angle. Inclusion of baseline muscle activity, the activity required to simply hold the grip dynamometer, greatly improved grip force predictions, especially at low force levels. While the complete model using six muscles and posture was the most accurate, the detailed validation and error analysis revealed that equations based on fewer components often resulted in a negligible reduction in predictive strength. Error was typically less than 10% under 50% of maximal grip force and around 15% over 50% of maximal grip force. This study presents detailed error analyses to both improve upon previous studies and to allow an educated decision to be made on which muscles to monitor depending on expected force levels, costs and error deemed acceptable by the potential user.

Keywords: Grip force; EMG; Regression; Prediction; Ergonomic tool

1. Introduction

The inherent difficulty of measuring grip and hand forces in the workplace has led to the development of alternate indirect methods by which to predict grip force. The methodology to predict hand forces in the workplace developed by Armstrong *et al.* (1979) has been widely adopted in various forms but more recent efforts to develop

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equations to predict grip force remain limited (Duque *et al.* 1995, Claudon 1998, 2003). The need to determine grip forces in the workplace stems from the link between pinch and grip forces, especially when combined with non-neutral postures, and musculoskeletal disorders of the upper extremity, such as carpal tunnel syndrome and tendonitis (Silverstein *et al.* 1986).

Armstrong *et al.* (1979) presented a method to predict forces at the hand from the position of the hand and fingers combined with muscle activity, which has been used to assess the injury risk potential of manual jobs (e.g. Silverstein *et al.* 1986). Their method created multiple posture-specific muscle activity – hand force calibration curves for each individual, and matched task components to the closest posture using video analysis. Since both the calibration and implementation procedures of Armstrong *et al.* (1979) are relatively time consuming and specific to the individual, researchers have attempted to develop less intensive procedures. Most efforts to simplify the process have related finger (or wrist) muscle electromyography (EMG) to grip force, which require only a single (maximal) calibration trial and result in an equation that may be applied generally and is not subject specific (Duque *et al.* 1995, Claudon 1998, 2003). This simplifies the relationship between EMG and grip force, as co-contraction of the wrist muscles is necessary to stabilize the wrist during gripping tasks (Snijders *et al.* 1987), as well as maintain wrist posture against gravity (Mogk and Keir 2003a). Furthermore, whilst these predictive equations have provided useful insights regarding grip force and muscle activity, they are limited in their scope. For example, forearm rotation (pronation/supination) has not been examined and the muscle activity required to hold a tool has been disregarded (Duque *et al.* 1995) or eliminated by using a supported grip dynamometer (Claudon 1998, 2003). These factors appear to limit the applicability of these equations within the workplace.

Although it might seem intuitive that the activity of the finger flexor muscles should provide the best estimate of grip force, the redundant nature of the forearm musculature and the need for co-contraction to maintain wrist posture complicate the relationship between EMG and grip force (Mogk and Keir 2003a). Both the degree of synergist activation and co-contraction of antagonist muscles have the potential to alter the net force measured (Lawrence and De Luca 1983), as well as the muscle activity required to produce a given force. While a nearly perfect linear relationship has been reported between finger flexor EMG and finger force in the absence of extensor co-activation (Danion *et al.* 2002), both flexors and extensors are active during gripping tasks (Snijders *et al.* 1987, Claudon 1998, 2003, Mogk and Keir 2003a). Wrist and forearm posture affect both muscle and moment arm lengths and thus the moment potential of a muscle (Loren *et al.* 1996), which in turn alters EMG amplitude (Inbar *et al.* 1987) as well as muscle synergies (Buchanan *et al.* 1989, Sergio and Ostry 1995). Whilst the relative muscle activity associated with grip force in several postures was previously evaluated (Mogk and Keir 2003a), a comprehensive investigation of forearm muscle activity and posture contributions to grip force is needed to determine the nature of muscle selection on grip force predictions across postures.

The purpose of this study was to present an equation or, more correctly, a family of equations, to predict grip force by further analysing a comprehensive dataset, which evaluated the effects of posture and grip force on forearm muscle activity (Mogk and Keir 2003a). The process for developing equations to predict isometric grip force based on forearm EMG and posture has been presented. To address limitations of previous equations and to improve transferability to the workplace, both wrist and finger musculature were included and participants were also required to support the grip dynamometer. The complete family of equations is presented, as is the rationale for selecting an acceptable equation to predict grip force based on maximizing accuracy of the prediction and minimizing the number of input variables.

2. Methods

The data used in this study were collected previously and fully described in Mogk and Keir (2003a), thus a brief overview is provided here. Ten healthy volunteers (five males and five females) had their maximum grip force ($Grip_{max}$) determined in a mid-prone forearm and neutral wrist posture using a grip dynamometer (MIE Medical Research Ltd., UK; mass = 450 g). Participants then performed exertions at five force levels (5, 50, 70 and 100% $Grip_{max}$, and 50 N) using a grip dynamometer (grip span of 5 cm) in each combination (nine in total) of three forearm (pronation, neutral/mid-prone and supination) and three wrist postures (45° extension, neutral and 45° flexion). Participants were seated upright with their right forearm resting on an adjustable horizontal platform, while the hand, wrist and dynamometer were left unsupported. Posture was monitored using a mirror apparatus that allowed wrist radioulnar deviation and flexion-extension angles to be recorded with a single video camera. Radioulnar deviation was maintained in neutral for all tests. Surface EMG was recorded from six forearm muscles: flexor carpi radialis (FCR); flexor carpi ulnaris (FCU); flexor digitorum superficialis (FDS); extensor carpi radialis (ECR); extensor carpi ulnaris (ECU); and extensor digitorum communis (EDC). EMG signals were normalized to maximal voluntary electrical activation (MVE) determined through a series of trials including maximal grip force with voluntary isometric wrist flexion and extension, forceful voluntary wrist circumduction, as well as resisted finger flexion and extension. Each experimental trial lasted 10 s, in which the participant held the dynamometer without exerting force ('baseline') then ramped up to the target force level, which was held for 3 s before returning to baseline. Average EMG (AEMG) was calculated from the 3 Hz linear envelope EMG over the 3 s plateau at the target force, as well as during baseline prior to each exertion. The relative grip force achieved for each trial was calculated as the average force exerted over the same 3 s plateau. Visual feedback, using an oscilloscope, enabled participants to maintain grip force exertions within 1.5% of the target force level for most trials, with the exception of the 100% target level and 70% and 100% trials with a flexed wrist. All trials were performed on each of 2 days ('Day 1' and 'Day 2') separated by a minimum of 4 and maximum of 7 days. The complete dataset for each day comprised 900 data points, which were used to develop (Day 1) and validate (Day 2) the equations.

2.1. *Equation development*

Multiple regression analyses were used to predict grip force from AEMG and postural data (figure 1). Analyses included linear, factorial and polynomial regressions (STATISTICA, v. 6.0, StatSoft Inc., Tulsa, OK, USA). A decision was made *a priori* to create the equations from Day 1 data and validate those equations using the data from Day 2. Relative grip force (% $Grip_{max}$) was input as the dependent variable, to be predicted by various combinations of AEMG and posture data (independent variables). Equations were initially created using all possible data from Day 1 ('full dataset').

Equations were developed using posture alone (forearm and/or wrist), muscle(s) alone and muscle(s) with posture. Wrist posture was input as nominal data (extension = 1,

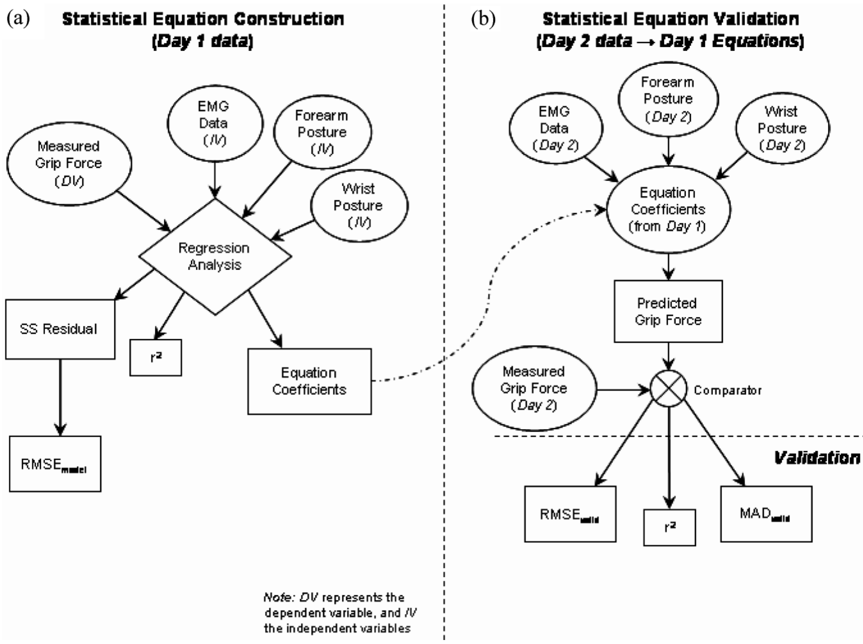

Figure 1. The process of constructing (a) and validating (b) the statistical equations. Ellipses represent inputs and rectangles represent outputs. Note that the equation coefficients determined from Day 1 data are carried over to the Day 2 data for the purpose of validating each equation. EMG = electromyography; RMSE = root mean square error; MAD = mean absolute difference.

neutral = 2, flexion = 3) and as the wrist angle in degrees as measured from videotape. Forearm posture was input only as a nominal variable (pronation = 1, neutral = 2 and supination = 3). The number and combination of muscles input into each model was manipulated to determine whether certain combinations of muscles would predict grip forces better than others (17 combinations in total). Muscle combinations included: 1) all muscles; 2) all flexor muscles; 3) all extensor muscles; 4) wrist flexors and extensors; 5) wrist flexors; 6) wrist extensors; and 7) finger muscles. As occupational studies often report the placement of electrodes over a 'common muscle mass', the following combinations of wrist muscles were also examined: 1) FCR and ECR; 2) FCR and ECU; 3) FCU and ECR; and 4) FCU and ECU. Equations were developed for each muscle individually as well. Gender was not included as a variable as it had previously been determined that no significant differences existed once the data were normalized (Mogk and Keir 2003a). Coefficients were included in each model if they were significant at a level of $p < 0.05$; however, coefficients were typically significant at $p < 0.001$.

The importance of including muscle activity from simply holding the grip dynamometer (i.e. 0% $Grip_{max}$) was examined by developing the equations in four ways: 1) including all baseline (0% $Grip_{max}$) data (also called 'full dataset'; 900 data points in total); 2) including baseline data as the mean posture-specific activation for each individual (also called 'average baseline'; 540 data points); 3) subtracting baseline activity from the AEMG of each grip exertion (also called 'baseline subtracted'; 450 data points); and 4) excluding baseline data (no zero point, also called 'no baseline'; 450 data points). Although baseline activity was recorded prior to each grip force exertion

(450 data points), previous tests showed that baseline activity did not vary with the target force to be exerted (Mogk and Keir 2003a). Consequently, the activity prior to each exertion in a given posture was used to calculate a mean posture-specific baseline activation level for each individual, effectively reducing baseline data from 450 to 90 data points ('average baseline' dataset). Additionally, based on the apparent nonlinear relationship between grip force and EMG (Duque *et al.* 1995, Claudon 1998), the effect of splitting the dataset and generating one group of equations for forces $\leq$50% Grip_{max}, and another for forces $>$50% Grip_{max} was examined.

The predictive ability of each model was judged based on the adjusted r^2 (explained variance) as a measure of fit and an overall root mean square error ($\text{RMSE}_{\text{model}}$, measured in % Grip_{max}) as a measure of predictive error magnitude, using the residual sum of the squares.

2.2. *Equation validation*

Each equation was developed from Day 1 data and validated using data from Day 2 (figure 1). Goodness of fit was determined by the r^2, validation RMSE ($\text{RMSE}_{\text{valid}}$) and the mean absolute difference (MAD) between the observed and predicted force data. Both $\text{RMSE}_{\text{valid}}$ and MAD are reported in % Grip_{max} and summarize the overall error for each model, with MAD thought to better represent the difference for the layperson. In addition to overall error measures for each equation, EMG and posture were input for specific levels of force (baseline, 50 N, forces $\leq$50% Grip_{max} and forces $>$50% Grip_{max}) to examine the ability to predict grip force across force levels. This provided a more detailed examination of the potential for certain muscles to be better predictors within specific grip force ranges, and will aid future users in appropriate equation selection based on the needs of their particular application.

3. Results

Equations developed using the full dataset (900 data points) resulted in the largest r^2 and smallest $\text{RMSE}_{\text{model}}$ values compared to the reduced datasets (table 1). However, with marginally larger error terms, use of the 'average baseline' dataset (540 data points) was deemed more representative of the activity prior to grip exertions in each posture, and less likely to bias the overall error of each model. Consequently, all equations presented account for the average muscle activity required by each individual to simply hold the dynamometer in each posture, without exerting a grip force.

3.1. *Equation*

The generic form of the complete model is found in equation 1. Second order models improved r^2 by nearly 4% and $\text{RMSE}_{\text{model}}$ by 2% over simple linear multiple regression. Third order polynomial and factorial models did not markedly improve r^2 or $\text{RMSE}_{\text{model}}$. These relationships remained true for all test conditions.

$$\text{Grip force} = \sum_{i=1}^{6}(a_i \cdot m_i + b_i \cdot m_i^2) + a_7 \cdot FA + b_7 \cdot FA^2 + a_8 \cdot W + b_8 \cdot W^2 + c \qquad (1)$$

where, Grip force is measured in % Grip_{max}, m_i is muscle activation (in % MVE) for each muscle (from 1 to 6), FA is forearm posture (pronation = 1, neutral = 2 and

Table 1. Comparison of the four approaches used to examine the effect of developing equations with and without baseline (0% $Grip_{max}$) data. Explained variance (r^2) and root mean square error ($RMSE_{model}$) for each polynomial equation developed from each Day 1 dataset variation, for equations created using muscle and wrist posture data as input variables.

	Muscle(s) (number included)	All baseline		Average baseline		Baseline subtracted		No baseline	
		r^2	$RMSE_{model}$	r^2	$RMSE_{model}$	r^2	$RMSE_{model}$	r^2	$RMSE_{model}$
Muscle groups	All (6)	0.911	8.8	0.892	10.3	0.862	11.1	0.870	10.7
	Flexors (3)	0.893	9.7	0.864	11.6	0.832	12.2	0.835	12.1
	Extensors (3)	0.853	11.3	0.832	12.8	0.799	13.4	0.797	13.5
	Wrist muscles (4)	0.905	9.1	0.885	12.1	0.859	11.2	0.863	11.1
	Wrist flexors (2)	0.885	10.0	0.856	11.9	0.820	12.7	0.829	12.4
	Wrist extensors (2)	0.817	12.7	0.793	14.2	0.799	13.4	0.756	14.8
Muscle pairs	Finger muscles	0.894	9.7	0.874	11.1	0.835	12.1	0.849	11.6
	FCR + ECR	0.898	9.5	0.882	10.7	0.849	11.6	0.860	11.2
	FCR + ECU	0.887	9.9	0.875	11.1	0.844	11.8	0.854	11.4
	FCU + ECR	0.886	10.0	0.858	11.8	0.835	12.1	0.827	12.4
	FCU + ECU	0.880	10.2	0.845	12.3	0.818	12.7	0.809	13.1
Single muscles	FCR	0.846	11.6	0.839	12.6	0.794	13.6	0.817	12.8
	FCU	0.869	10.7	0.825	13.1	0.783	13.9	0.783	14.0
	FDS	0.875	10.5	0.841	12.5	0.809	13.1	0.806	13.2
	ECR	0.805	13.1	0.784	14.6	0.786	13.8	0.748	15.0
	ECU	0.764	14.4	0.728	16.4	0.739	15.3	0.676	17.1
	EDC	0.818	12.7	0.803	13.9	0.740	15.3	0.766	14.5
Total no. of data points		900		540		450		450	

FCR = flexor carpi radialis; ECR = extensor carpi radialis; ECU = extensor carpi ulnaris; FCU = flexor carpi ulnaris; FDS = flexor digitorum superficialis; EDC = extensor digitorum communis.

supination = 3), W is wrist posture (extension = 1, neutral = 2 and flexion = 3), c is a constant and represents the y-intercept, and a_i and b_i represent the first and second order coefficients for each variable, respectively. A select list of coefficients and error measures for many equations, with and without wrist posture, is found in table 2.

3.1.1. Contribution of model components. Posture alone (wrist and/or forearm angle) was not a good predictor of grip force, explaining less than 3.5% of the variance. However, including wrist posture improved all multiple muscle equations to r^2 values of 0.84 – 0.89 with $RMSE_{model}$ less than 13%, except when only extensor muscles were used (table 2). When used in combination with muscle activity, wrist posture increased the r^2 of all models containing wrist muscles by approximately 3% and reduced $RMSE_{model}$ by 1%. Similarly, r^2 and $RMSE_{model}$ for equations derived from finger EMG were improved by 2.9 – 8.8% and 1.2 – 2.7%, respectively, with the addition of wrist posture. Using the measured wrist angle (in degrees) resulted in models of similar strength to those of nominal form (i.e. 1, 2 or 3), thus nominal wrist posture was used. Forearm posture did little to improve the predictive ability of each model, often not altering either r^2 or $RMSE_{model}$ compared to those created from muscle activity alone (comparing across columns in table 3). As a result, forearm posture was excluded to minimize the number of input variables.

When only muscle activity was used (i.e. no wrist or forearm angles), equations developed from individual or multiple flexor muscles only were always better predictors of grip force than those based solely on extensor muscles. Predictive strength also improved with the number of muscles included, with the exception that single flexor muscles were better than any combination of extensor muscles. Further examination of the EMG input revealed increased predictive strength when baseline data (0% $Grip_{max}$) were included with the other force levels, improving r^2 and $RMSE_{model}$ values by 2 – 6% and 0.4 – 2.0%, respectively.

The most accurate model was based on all six muscles with wrist posture; however, the coefficient for the finger flexor activity (FDS) was not significant and fell out of the final equation (table 2). The paired wrist muscle models were almost as accurate as the full model. For example, the equation based on the combination of FCR and ECR (with wrist posture) had an r^2 of nearly 90% and $RMSE_{model}$ only 0.4% higher than that of the full model. As a rule, single muscles were not as good as multiple muscle models. Of the single muscle models, the finger muscles (FDS and EDC) were the best predictors of grip force for the flexor and extensor muscles, respectively, once wrist posture was included. As a single input variable, ECU had the least predictive power, explaining less than 75% of the variance.

3.2. *Model validation (application of Day 2 data)*

3.2.1. Full data range equations. Day 2 grip forces were well predicted using the equations developed from Day 1 data ($r^2 = 0.819 \pm 0.056$ over all equations; figure 2). Both r^2 and RMSE values were remarkably similar between equation development (Day 1 data) and validation with Day 2 data (table 3 vs. table 4). The MAD between the observed and predicted Day 2 grip forces was always 2.4 – 5.4% lower than the $RMSE_{valid}$ (table 4).

3.2.2. Full data range equations: prediction dependence on input force range. To test the usability of the equations based on the full data range, specific ranges of data were evaluated in isolation. Error increased as the grip force became greater (table 5).

Table 2. Coefficients, to be substituted into Equation 1, for models with both flexor and extensor muscles as input developed using the 'average baseline' Day 1 dataset ('Muscles only' and 'Muscles + Wrist').

	All*		Wrist muscles†		Finger muscles†		FCR + ECR†		FCR + ECU†		FCU + ECR†		FCU + ECU†	
	Muscles only	Muscles + Wrist	Muscles only	Muscles + Wrist	Muscles only	Muscles + Wrist	Muscles only	Muscles + Wrist	Muscles only	Muscles + Wrist	Muscles only	Muscles + Wrist	Muscles only	Muscles + Wrist
c – Int.	−6.1405	0.6749	−3.4310	2.2850	−5.8447	3.7730	−6.8685	−6.2082	−3.7619	4.4625	−4.6463	−6.7826	−2.3828	5.6585
a_1 – FCR	1.3075	1.1284	1.3422	1.4565			1.4570	1.5329	1.6184	1.7363				
b_1 – FCR^2	−0.0071	−0.0069	−0.0087	−0.0099				−0.0110	−0.0105	−0.0127				
a_2 – FCU	0.1653	0.1357	0.2472								1.6261	1.5855	1.8572	1.9445
b_2 – FCU^2											−0.0121	−0.0129	−0.0146	−0.0168
a_3 – FDS					1.6598	1.3643								
b_3 – FDS^2					−0.0114	−0.0089								
a_4 – ECR	1.0101	0.7113		1.0703			1.3393	1.3199			1.1477	1.2165		
b_4 – ECR^2	−0.0095	−0.0058		−0.0105			−0.0134	−0.0117			−0.0110	−0.0102		
a_5 – ECU	0.2603	0.1462	0.9045	0.2078					0.8967	0.8682			0.8276	0.7875
b_5 – ECU^2			−0.0081						−0.0068	−0.0061			−0.0074	−0.0060
a_6 – EDC		0.7466			1.1707	1.4882								
b_6 – EDC^2	−0.0040	−0.0095			−0.0126	−0.0140								
a_7 – FA														
b_7 – FA^2														
a_8 – Wrist								9.4734				12.157		
b_8 – Wrist^2		−2.0264		−2.0097		−2.5888		−4.3863		−1.9081		−4.9709		−1.8148
r^2	0.854	0.892	0.840	0.885	0.807	0.874	0.837	0.882	0.836	0.875	0.815	0.858	0.810	0.845
$RMSE_{model}$	11.9	10.3	12.6	12.1	13.8	11.1	12.7	10.7	12.7	11.1	13.5	11.8	13.7	12.3
No. of terms	8	10	6	7	5	6	4	7	5	6	5	7	5	6

*Refers to the use of all six muscles.

†The muscles included in each model.

'Wrist muscles' includes flexor carpi radialis (FCR), flexor carpi ulnaris (FCU), extensor carpi radialis (ECR) and extensor carpi ulnaris (ECU). 'Finger muscles' corresponds to flexor digitorum superficialis (FDS) and extensor digitorum communis (EDC).

Table 3. Equation development. Explained variance (r^2) and root mean square error ($RMSE_{model}$) for each polynomial equation developed from the 'average baseline' Day 1 dataset, for each combination of input variables.

	Muscle(s) (number included)	Muscles only		Muscles + Wrist		Muscles + Forearm		Muscles + Forearm + Wrist	
		r^2	$RMSE_{model}$	r^2	$RMSE_{model}$	r^2	$RMSE_{model}$	r^2	$RMSE_{model}$
Muscle groups	All (6)	0.854	11.9	0.892	10.3	0.858	11.8	0.895	10.1
	Flexors (3)	0.826	13.1	0.864	11.6	0.826	13.1	0.864	11.6
	Extensors (3)	0.757	15.5	0.832	12.8	0.772	15.0	0.849	12.1
	Wrist muscles (4)	0.840	12.6	0.885	12.1	0.842	12.4	0.889	10.4
	Wrist flexors (2)	0.822	13.2	0.856	11.9	0.822	13.2	0.856	11.9
	Wrist extensors (2)	0.737	16.1	0.793	14.2	0.772	15.0	0.827	13.0
Muscle pairs	Finger muscles	0.807	13.8	0.874	11.1	0.807	13.8	0.874	11.1
	FCR + ECR	0.837	12.7	0.882	10.7	0.837	12.7	0.882	10.7
	FCR + ECU	0.836	12.7	0.875	11.1	0.837	12.6	0.877	11.0
	FCU + ECR	0.815	13.5	0.858	11.8	0.826	13.1	0.869	11.3
	FCU + ECU	0.810	13.7	0.845	12.3	0.826	13.1	0.862	11.6
Single muscles	FCR	0.802	14.0	0.839	12.6	0.806	13.8	0.843	12.4
	FCU	0.793	14.3	0.825	13.1	0.797	14.2	0.828	13.0
	FDS	0.790	14.4	0.841	12.5	0.790	14.4	0.842	12.5
	ECR	0.721	16.6	0.784	14.6	0.741	16.0	0.805	13.8
	ECU	0.687	17.6	0.728	16.4	0.743	15.9	0.786	14.5
	EDC	0.704	17.1	0.803	13.9	0.709	16.9	0.809	13.7

FCR = flexor carpi radialis; ECR = extensor carpi radialis; ECU = extensor carpi ulnaris; FCU = flexor carpi ulnaris; FDS = flexor digitorum superficialis; EDC = extensor digitorum communis.

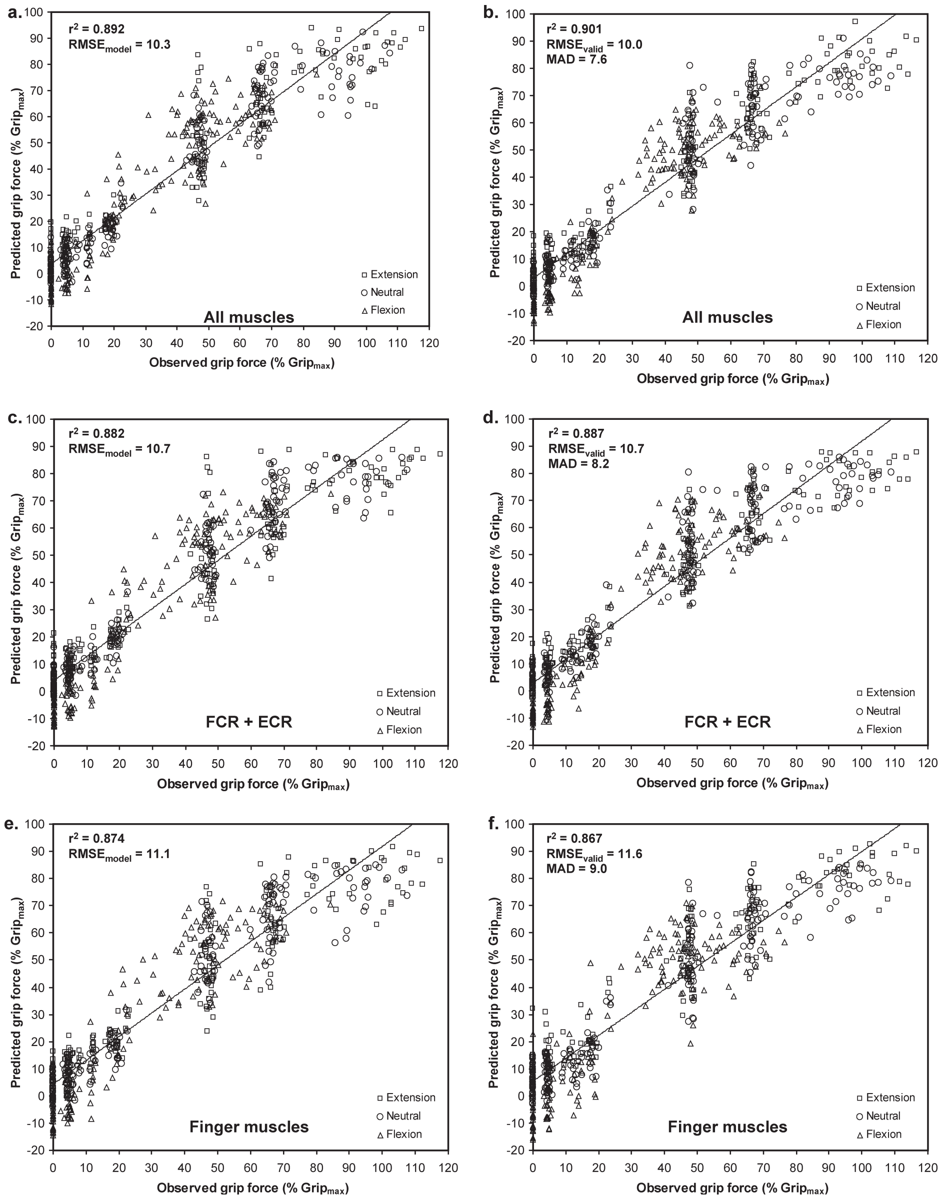

Figure 2. Correlation between observed grip force values and those predicted by models developed from 'All' muscle data (a and b), 'flexor carpi radialis (FCR) + extensor carpi radialis (ECR)' (c and d) and 'Finger muscles' (e and f), plus wrist posture. Graphs a, c and e represent the development of the model (Day 1 data → Day 1 model), whilst graphs b, d and f are for model validation (Day 2 data → Day 1 model). Different symbols are used to distinguish between the data points for each wrist posture, and provide information on the error distribution between force levels. RMSE = root mean square error; MAD = mean absolute difference.

Table 4. Equation validation. Explained variance (r^2), root mean square error ($RMSE_{valid}$) and mean absolute difference (MAD) between measured and predicted grip force levels (in % $Grip_{max}$) for the validation of each model in Table 1 using the Day 2 dataset.

	Muscle(s) (number included)	Muscles only			Muscles + Wrist			Muscles + Forearm			Muscles + Forearm + Wrist		
		r^2	$RMSE_{valid}$	MAD	r^2	$RMSE_{valid}$	MAD	r^2	$RMSE_{valid}$	MAD	r^2	$RMSE_{valid}$	MAD
Muscle groups	All (6)	0.868	11.6	8.1	0.901	10.0	7.6	0.868	11.7	8.0	0.901	10.1	7.6
	Flexors (3)	0.838	12.7	9.1	0.868	11.5	8.8	0.838	12.7	9.1	0.868	11.5	8.8
	Extensors (3)	0.760	15.6	11.6	0.826	13.2	10.4	0.757	15.9	11.2	0.835	12.9	10.1
	Wrist muscles (4)	0.838	13.0	9.2	0.881	11.0	8.3	0.836	13.1	9.2	0.884	10.9	8.3
	Wrist flexors (2)	0.840	12.7	9.2	0.866	11.6	9.1	0.840	12.7	9.2	0.866	11.6	9.1
	Wrist extensors (2)	0.735	16.4	12.0	0.793	14.5	11.3	0.757	15.9	11.2	0.812	13.9	10.5
Muscle pairs	Finger muscles	0.810	13.8	10.2	0.867	11.6	9.0	0.810	13.8	10.2	0.867	11.6	9.0
	FCR + ECR	0.847	12.4	8.7	0.887	10.7	8.2	0.847	12.4	8.7	0.887	10.7	8.2
	FCR + ECU	0.831	13.3	9.4	0.861	12.1	9.3	0.829	13.4	9.4	0.859	12.3	9.4
	FCU + ECR	0.835	12.9	9.0	0.870	11.4	8.8	0.840	12.6	8.9	0.876	11.1	8.7
	FCU + ECU	0.816	13.7	9.6	0.839	12.8	9.8	0.818	13.7	9.6	0.842	12.8	9.8
Single muscles	FCR	0.830	13.1	9.4	0.859	11.9	9.4	0.832	13.0	9.4	0.861	11.9	9.3
	FCU	0.814	13.6	9.8	0.839	12.7	9.8	0.816	13.6	9.9	0.839	12.7	9.8
	FDS	0.788	14.6	10.5	0.835	12.8	9.8	0.788	14.6	10.5	0.835	12.8	9.8
	ECR	0.719	16.8	12.6	0.785	14.7	11.5	0.732	16.3	11.9	0.799	14.2	11.0
	ECU	0.660	19.1	13.7	0.692	18.2	13.7	0.697	18.2	13.0	0.730	17.3	12.8
	EDC	0.686	17.7	13.4	0.774	15.0	11.8	0.689	17.6	13.3	0.777	14.9	11.8

FCR = flexor carpi radialis; ECR = extensor carpi radialis; ECU = extensor carpi ulnaris; FCU = flexor carpi ulnaris; FDS = flexor digitorum superficialis; EDC = extensor digitorum communis.

Table 5. Error associated with testing equations with limited input data ranges of Day 2 data. Root mean square error (RMSE) and mean absolute difference (MAD), in % $Grip_{max}$, in specific ranges of force for each equation developed from flexor and extensor muscle combinations, based on model validation.

Muscles included in equation	Error measure	Baseline (0% $Grip_{max}$)		50 N*		≤50% $Grip_{max}$†		≤50% $Grip_{max}$ (no baseline)‡		>50% $Grip_{max}$		Full range	
		Muscles only	Muscles + Wrist	Muscles only	Muscles + Wrist	Muscles only	Muscles + Wrist	Muscles only	Muscles + Wrist	Muscles only	Muscles + Wrist	Muscles only	Muscles + Wrist
All	RMSE	2.9	6.3	6.1	6.0	8.8	8.1	9.9	8.6	16.9	13.8	11.6	10.0
	MAD	2.3	5.2	4.9	4.6	6.0	6.3	7.1	6.7	13.5	11.0	8.1	7.6
Wrist	RMSE	3.5	6.8	7.0	6.0	10.2	8.8	11.4	9.3	18.4	15.4	13.0	11.0
	MAD	2.7	5.8	6.0	4.6	7.0	6.8	8.3	7.1	15.1	12.4	9.2	8.3
Finger	RMSE	8.3	9.3	9.9	8.8	12.0	10.4	12.9	10.7	17.7	14.2	13.8	11.6
	MAD	5.7	7.6	7.5	6.4	8.7	8.1	9.6	8.3	14.2	11.1	10.2	9.0
FCR + ECR	RMSE	3.3	7.1	6.5	6.1	9.5	8.9	10.7	9.4	17.9	14.3	12.4	10.7
	MAD	2.5	6.1	5.1	4.6	6.4	6.9	7.6	7.1	14.4	11.6	8.7	8.2
FCR + ECU	RMSE	3.7	6.3	7.2	7.6	10.2	9.8	11.5	10.6	19.1	16.7	13.3	12.1
	MAD	2.9	5.3	6.1	6.1	7.0	7.6	8.3	8.3	15.6	13.6	9.4	9.3
FCU + ECR	RMSE	3.4	7.4	7.4	7.7	10.4	9.9	11.7	10.6	17.7	14.6	12.9	11.4
	MAD	2.6	6.4	5.5	6.0	6.8	7.8	8.1	7.8	14.6	11.6	9.0	8.8
FCU + ECU	RMSE	3.1	6.0	7.5	8.3	10.8	10.5	12.2	11.5	19.2	17.5	13.7	12.8
	MAD	2.3	5.2	5.9	6.6	7.1	8.1	8.6	9.0	15.9	14.2	9.6	9.8
No. of observations		90		90		390		300		150		540	

*The relative effort level of the 50 N trials was dependent on each participant's strength and ranged from 8.8 – 24.2% $Grip_{max}$.

†Includes baseline, 5% $Grip_{max}$ trials, 50 N trials and any other trials where grip force was ≤ 50% $Grip_{max}$. Note that the full range column corresponds to the $RMSE_{valid}$ in table 4.

‡Includes 5% $Grip_{max}$ trials, 50 N trials and any other trials where ≤ 50% $Grip_{max}$ was achieved.

FCR = flexor carpi radialis; ECR = extensor carpi radialis; FCU = flexor carpi ulnaris; ECU = extensor carpi ulnaris.

Specifically, when grip force was limited to $\leq$ 50% $Grip_{max}$ (including baseline), RMSE was 1.2–3.3% lower than the overall (full range) $RMSE_{valid}$, whilst the RMSE for forces above 50% was 2.6–5.8% higher, regardless of the input variables. When the baseline data were excluded from the $\leq$ 50% $Grip_{max}$ validation tests, error became 0.8–1.8% lower than that of the full range validation ($RMSE_{valid}$). As with the model development, $RMSE_{valid}$ decreased with the inclusion of wrist posture, largely due to improved prediction (1.7–3.6% lower) of forces greater than 50% $Grip_{max}$ (table 5). Models using some combination of wrist flexor and extensor muscles were better predictors of grip forces below 50% $Grip_{max}$ than finger muscles, but their ability to predict forces above 50% $Grip_{max}$ was similar.

4. Discussion

In this study, a family of equations was developed to predict grip force (from zero to maximum) using muscle activity and posture, with an emphasis on the lower range (0 to 50%) to reflect the distribution of forces in the workplace. A number of equations were developed using reduced datasets and provided useful alternatives to the complete model by requiring fewer muscles while maintaining accuracy. By using six forearm muscles under a wide range of conditions in both men and women, and a validation process that included a detailed assessment of error, this study represents a more comprehensive evaluation of grip force prediction than previously available. Previous efforts in this area have included only one or two muscles, avoided forces below 20–30% of maximum and have generally been limited to men (Duque *et al.* 1995) or women (Claudon 1998). It was found that muscle activity was the most important input, being required to predict over 85% of the variance. Although posture alone had little predictive power (<4%), the accuracy of equations developed using muscle activations improved with wrist posture but were not markedly changed with forearm posture. In addition to monitoring more muscles than previous research, lower forces (0%, 5% and 50 N, the latter ranging from about 8–25% maximum) were included, which relate more closely with a suggested limit of 17% of maximum to prevent fatigue during intermittent hand gripping and 10% during continuous gripping (Byström and Fransson-Hall 1994).

Although the full equation with all six muscles and wrist posture provided the best statistical estimate of grip force, reducing the number of input muscles did not necessarily compromise the usefulness of the resulting equation. For example, equations developed from a combination of flexor and extensor muscles (e.g. 'FCR + ECR' or the 'finger muscles') resulted in RMSE and r^2 values comparable to the full model (table 3 and figure 2). Despite the finger muscles (FDS and EDC) representing the best single muscle predictors for each respective muscle group (table 3), the finger flexor (FDS) coefficients were non-significant when all six muscles were used to develop the equations and thus did not appear in the final model (table 2). Flexor muscles proved to be stronger predictors of grip force than any of the extensors, as was previously reported during low-level pinch force exertions (Maier and Hepp-Reymond 1995). Under controlled conditions that allowed activation of the extrinsic finger flexors in the absence of extensor co-activation, finger flexor EMG has been nearly perfectly related to measured finger forces (Danion *et al.* 2002). However, gripping tasks require concurrent activation of the finger flexor and extensor muscles (Claudon 1998, 2003, Mogk and Keir 2003a) resulting in co-contraction, which would alter the EMG–force relationship. It is unlikely that the statistical similarity between models could be explained by EMG cross-talk between forearm muscles, as suggested by a previous study examining electrode placement and spacing (Mogk and Keir 2003b).

The inclusion of posture had a relatively small effect on the overall predictive strength in the development (table 3) and validation (table 4) of most models. The benefit of including wrist posture was most evident above 50% $Grip_{max}$ resulting in a 2.0 – 3.5% decrease in RMSE, whilst the error at baseline increased by 1 – 4% over muscle activity alone (table 5). Finger muscle models were the most sensitive to wrist posture (table 3). Forearm posture made little to no improvement over muscles alone, with the exception of those models consisting of only wrist extensor muscles, either individually or in combination, particularly with ECU (table 3). Perhaps the most interesting finding in this study was the similar predictive power between nominal and actual wrist posture. Whilst this may have been a result of reduced variation in the wrist angles set by the protocol (within about 5° of the desired posture), the use of nominal wrist posture may allow ergonomists the liberty of eschewing the sometimes cumbersome and expensive wrist goniometers, as previously required (Duque *et al.* 1995, Claudon 1998, 2003). In addition, it may allow a more normal activity profile with fewer potential interruptions to the worker. However, further testing is required to determine whether nominal wrist angles effectively represent the continuum of wrist postures, with potential issues at the boundaries of the categories (e.g. 'neutral' vs. 'flexion'). Given the trade-off in predictive accuracy with the addition of nominal wrist posture between high ($>50\%$) and low forces, it could be argued that wrist posture is not necessary to estimate grip force under certain conditions.

The decision to include baseline muscle activity (the activity associated with holding the grip force dynamometer in each posture) was particularly important given that the inclusion of baseline data (0% $Grip_{max}$) increased explained variance by 6% and reduced RMSE by up to 2%. This effect was seen mainly at low force levels. Baseline muscle activity was included, in part, because it has previously been reported to have a large effect on muscle force estimation at the wrist (Buchanan *et al.* 1993). In the current study, most equations resulted in a negative y-intercept (table 2), which would predict negative grip forces in the absence of muscle activity (which is possible mathematically but not physiologically). This was expected as the grip dynamometer was reset to zero prior to each exertion, and relates to the force required to hold the dynamometer, which amounted to approximately 10 N or 2 – 5% of each participant's maximum. Negative intercepts could have been avoided by zeroing the dynamometer prior to it being held by the participant; however, this would have introduced errors at low force levels due to the orientation of the dynamometer. Previous attempts to develop equations to predict grip force avoided baseline muscle activity in different ways. Duque *et al.* (1995) deemed baseline activity 'negligible' and thus omitted it, despite results indicating that almost 30% activation was required to hold the flexed posture. Claudon (1998, 2003) supported the dynamometer, but also imposed each posture by fixing the orientation of the dynamometer, which may alter muscle activation during gripping (Johansson *et al.* 2004). The current results indicate that the muscle activity associated with holding the dynamometer in specific postures is a large determinant of low level grip force and should not be disregarded as has been done previously, especially considering that the dynamometer used is much lighter (by over 50%) than even light tools (e.g. pneumatic nutrunner; Lin *et al.* 2003).

One goal of this study was to determine the nature of the errors associated with the equations, especially relating to error magnitude for different ranges of force, as previous research has been very limited in this regard. Comprehensive error analysis was conducted by incorporating a test (Day 2) dataset in its entirety and by isolating specific grip force ranges (table 5). It was found that error was not constant across force ranges. Whilst almost 70% of all grip force predictions were within 10% $Grip_{max}$ of the measured values (tables 2 and 5), error increased with relative grip force (table 5). By limiting the range of validation grip force data, grip forces were generally predicted within 7% $Grip_{max}$ (RMSE and MAD) at baseline, increasing by 1 – 2% for grip forces less than 50% (including baseline), to near 10% for the same range without using baseline data. Predictions above 50% were still within 15% maximum when all components were included (table 5). It should be noted that, although table 1 may give reason to use the 'full dataset' version of the equations, the additional 'baseline' data points biased the error measures at the lower grip forces with higher error over 50% $Grip_{max}$, leading to the decision to use the average baseline. Previous studies have been very limited in their assessments of error. Claudon (1998) reported an overall error of 9.9% and, although it was not formally presented, suggested the predictive error was less than 20% for forces below 50% and less than 40% for forces above 50% maximum for a two muscle model. Duque *et al.* (1995) presented only correlation coefficients and suggested that there was 'a good approximation' of forces under 60% of maximum for their one muscle model. It should be noted that the overall error presented by Claudon (1998) is comparable to the present equations based on muscle pairs and wrist posture ($RMSE_{valid} = 10.3$–12.3%; table 4), with a marginally larger error in the current study likely due to the increased variance in grip force of using of an unsupported dynamometer. In a recent study, Claudon (2003) reported an error of 6.9% using continuous and linearly increasing forces between 0 and 70%, which is similar to the current study when restricted to the same input data range. The validation analysis presented in this study was included to fill a void of such measurements in previous reports and to allow educated decisions on which muscles to monitor depending on expected force levels and error deemed acceptable by the user.

There are limitations to the current study. Wrist posture was maintained in neutral radioulnar deviation and the elbow angle remained constant, thus the grip force predictions may be affected if these angles changed. Although the procedure consisted of a time varying force profile, the analysis used a 3 s mean during the isometric isotonic portion. The dynamometer was set at a grip span of 50 mm for all participants, other studies have used 45 mm (Duque *et al.* 1995, Claudon 1998, 2003), thus subtle differences between studies may exist. Additionally, the mass of the dynamometer (450 g) is less than many hand tools; for example, the lightest pneumatic nutrunner used in a recent study was 1.4 kg (Lin *et al.* 2003). All of the equations were developed using the full range of grip forces (0 to maximum), thus it could be argued that the results may have differed had the equations been based on grip force ranges. However, the equations were generated by splitting the dataset at 50% maximum and no benefit was found over the equations based on the full data range. Finally, the validation process used in this study may be criticized for consisting of the same participants repeating the conditions used in the development of the equations. However, the validation data is a complete dataset collected on a separate occasion allowing assessment of day-to-day reliability of the equation(s) and represents a large improvement over the limited validations of previous efforts.

Rather than suggesting that an 'ideal' equation exists for all circumstances, a family of equations was presented, developed using a number of muscles and postures with a comprehensive assessment of the errors involved. It was found that inclusion of muscle activity in the model was required to explain over 85% of the grip force variance, and whilst posture alone had little predictive power, the inclusion of wrist posture improved predictions with muscle activity. It was also found that nominal wrist posture was more effective than measured angle. Baseline muscle activity, the activity required to simply hold the dynamometer without actively exerting a grip force, was an important inclusion and improved prediction of grip, especially at low levels. Detailed error analysis revealed that equations based on fewer components often resulted in a negligible reduction in predictive strength, thus users may select an equation that reflects the limitations of their ergonomic assessment or may allow the user to choose to reduce the detail of their assessment based on knowing the errors associated with each equation. Additionally, if the ergonomist requires an actual grip force, the relative value predicted by the equation may be multiplied by the worker's maximum. Realism was improved by requiring the grip dynamometer to be supported by the subject; however, there remains a need to further examine force varying isometric contractions as well as unconstrained dynamic tasks typical in the workplace, including those with a pinch grip. Whilst these equations stand on their own, their full value might be realized combined with the extra activation required to stabilize a tool in the hand, such as described with the 'moment wrench' (Wells and Greig 2001). These findings provide a useful refinement to long-standing grip force prediction methods by providing insight as to the muscle(s) to monitor and the need to directly monitor posture.

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