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Testing apparatus and experimental procedure for position specific normalization of electromyographic measurements of distal upper extremity musculature

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Abstract

Objective. An apparatus and procedure are described to determine position specific normalization coefficients for surface EMG of upper extremity musculature.

Study design. Thirty-nine subjects were tested three times. Repeatability of EMG measurements across test sessions was determined by computing intraclass correlation coefficients. Two-way analysis of variance was used to test upper extremity position dependent differences in EMG measurements.

Background. EMG measurements are susceptible to error from skin movement and muscle length changes, both of which may occur when upper extremity positions vary. Normalization of the EMG signal without consideration for such positional influences may lead to erroneous conclusions regarding muscle activation during functional tasks.

Method. An apparatus was designed that allowed subjects to perform three repetitions of maximum elbow flexion, forearm pronation, wrist extension, and wrist flexion with the forearm in neutral and pronated positions. Surface EMG was sampled from eight muscles. Mean EMG on maximum voluntary contraction was computed, and resting EMG was subtracted to obtain EMG normalization coefficients.

Results. Upper extremity position affected the EMG normalization coefficient for biceps brachii, which was lower in the pronated position, and extensor carpi radialis, which was higher in the pronated position ($P < 0.00625$).

Conclusions. The apparatus accommodates various combined positions of the elbow, forearm and wrist. The normalization procedure is efficient for testing subjects who are being observed during functional tasks. Only two muscles were affected by upper extremity position, but group trends were not always consistent with individual behavior. This method would ensure the use of appropriate EMG normalization coefficients regardless of individual variation.

Relevance

This method is effective for normalizing EMG signals using task specific upper extremity positions. It may be used to test isometric exertions of distal upper extremity musculature for clinical and research purposes.

Keywords: Biomechanics; Apparatus; EMG normalization; Upper extremity musculature; Elbow; Forearm; Wrist

1. Introduction

Surface electromyography (EMG) is used widely to estimate muscle activation intensity during functional tasks. Surface electrodes are non-invasive and, because they have a large pick up area, they are appropriate for the study of gross muscular function. They are well suited for the study of the temporal relationship between the EMG signal and muscle contraction dynamics and, to a limited extent, the magnitude of muscle contraction force [1,2]. The disadvantages of surface EMG technique are that it lacks muscle fiber specificity, is

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susceptible to signal artifacts due to poor contact between the detection surface and the skin, is highly influenced by the filtering and impedance effects of surrounding tissues, and is susceptible to cross-talk from adjacent muscles.

Normalization of the EMG signal amplitude is a form of force calibration and is necessary if comparisons between muscle contraction intensities within and between subjects are to be made [3,4]. The most common method of normalization is to obtain reference isometric contractions of maximum effort, or maximum voluntary contractions (MVC), for each muscle or muscle group under investigation [3,5,6]. All other EMG signals obtained during testing are then expressed as a percentage of the appropriate reference EMG signal obtained during MVC.

Armstrong et al. [8] presented an EMG-force normalization method using a range of hand grip force exertions for investigations of dynamic (i.e. non-isometric) muscle contractions. They suggested sampling a series of isometric grip forces for each electrode site at several fixed hand positions that were likely to occur during the performance of sewing machine operation. This produced a series of calibration curves for each electrode at each hand position. The force exerted by the hand of the operator could be estimated from the experimentally obtained EMG signal and the appropriate calibration curve if the hand position was also measured. This technique requires the simultaneous sampling of joint kinematics and is desirable if the estimation of muscle force from EMG is intended. The biggest drawback to this technique is the time required to sample and process the family of EMG-force-position values and then to select the appropriate normalization coefficient for each region of the task related EMG signal. If such precise relationships between EMG, force and position are not required, a more simplified normalization procedure may be appropriate.

The purpose of this article is to describe the design of an apparatus and the experimental procedure used to normalize surface EMG data sampled from eight muscles of the distal upper extremity (UE). The effect of UE position on the EMG normalization signals was determined.

In this study, an adaptation of the method suggested by Armstrong et al. [8] was used. EMG signals were sampled during isometric MVC of each muscle with the distal UE maintained in two different combinations of elbow, forearm and wrist positions. These positions were chosen based upon those adopted by subjects during two specific functional tasks that we intended to compare in a future study and were, therefore, task specific. Other task specific positions could have been chosen to illustrate the usefulness of this procedure.

2. Methods

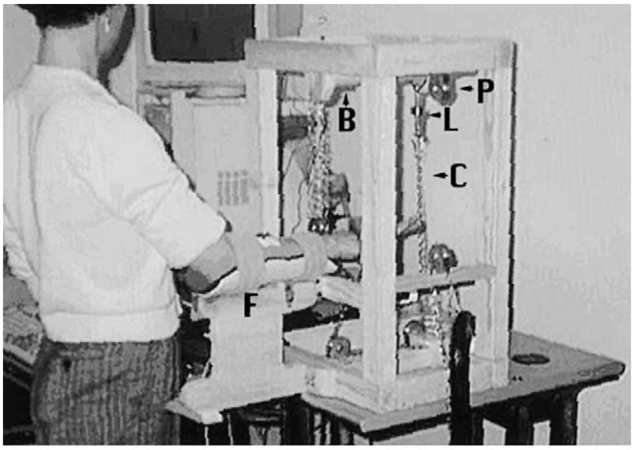
2.1. Apparatus and instrumentation

The EMG normalization apparatus was designed for use in both laboratory and field settings. It was light weight, portable, easy to assemble, comfortable to subjects, and sufficiently flexible to permit unencumbered performance of elbow, forearm, and wrist exertions in multiple directions.

The following anthropometric dimensions were taken into consideration in the design of the apparatus: forearm-hand length, hand breadth, hand length, hand circumference, and forearm circumference [9]. Both male and female US adults were the population of interest. In order to accommodate the 5th to 95th percentile of this population, the principle of design for extreme individuals was applied. This enabled the relatively large forelimb dimensions of 95th percentile males to be accommodated with adjustable features added to ensure that smaller forelimbs could also be tested.

The apparatus, which is shown in Figs. 1 and 2, consisted of a solid wooden frame, 0.6 m $\times$ 0.6 m $\times$ 0.5 m, with a padded, concave plastic forearm support (F) attached to the front of the frame and an adjustable padded handle (H). The forearm support allowed for the positioning of the distal forearm and the hand within the interior of the frame, and the padded, wooden handle encircled the transverse metacarpal arch of the hand, thereby leaving rotation about the wrist unconstrained in the flexion–extension plane. This handle was attached by a length of chain (C) to a uniaxial, tension–compression load cell (L) (Model LCF-500, Omega Engineering, Stamford, CT, USA) either directly or through a system of pulleys (P). The load cell was in turn attached to a metal ring (R) that encircled a horizontal, cylindrical beam (B) that bisected the top face of the wooden frame. This ring attachment allowed the load cell to slide along the beam until it was positioned directly over the handle, regardless of wrist position or forelimb length.

Through the system of pulleys, the application of a tensile force to the load cell resulted from any isometric exertion of elbow flexion, forearm pronation, or wrist flexion–extension. Furthermore, once a position was established, it was not necessary to reposition the UE for the performance of the different contractions. For example, maximum isometric wrist extension (Fig. 1, top) could be tested in the same UE position as maximum isometric wrist flexion (Fig. 1, bottom) by simply rerouting the chain from the bottom pulleys to attach directly to the top of the handle. Fig. 2 shows the chain and pulley configurations available to test the right UE in the forearm pronated position for wrist extension and elbow flexion (C1) and wrist flexion (C3). In the pronated forearm position, forearm pronation was tested



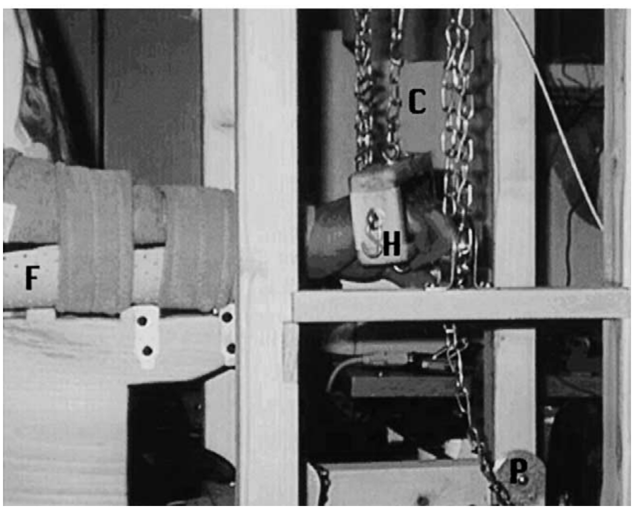


Fig. 1. Photographs of apparatus designed for position specific normalization of electromyographic recordings of the muscles of the distal forearm. Attached to the wooden frame is a padded forearm support (F) with wide straps for limb stabilization. The four fingers are inserted to the level of the transverse metacarpal arch into a wooden handle (H) that is lined with a soft, foam pad. The handle is attached to a load cell (L) either directly or through a system of pulleys (P) by lengths of chain (C). The load cell is in turn attached with a metal ring to a cylindrical beam (B) that bisects the top face of the wooden frame. In the top figure, the handle is attached to the load cell through two lower pulleys on the bottom face of the frame for application of resistance to maximum wrist extension in the forearm pronated position. In the bottom figure, the top of the handle is attached directly to the load cell for application of resistance to maximum wrist flexion in the forearm pronated position. These exertions could be tested in the forearm neutral position by supinating the forearm 90° then routing the chain through the side mounted pulleys.

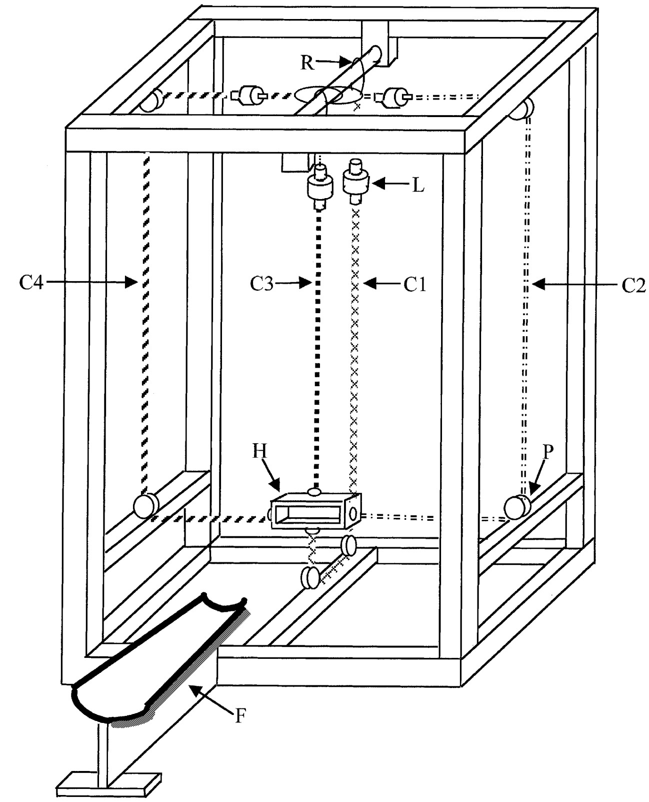


Fig. 2. Diagram of apparatus designed for position specific normalization of electromyographic recordings of the muscles of the distal forearm. F = forearm, support, H = padded handle, L = load cell, R = metal ring attaching load cell to cross beam, C = chain and P = pulley. The handle is oriented in the diagram for the pronated forearm position. With the right forearm pronated, the chain configurations needed to test maximum voluntary contractions of the eight UE muscles are as follows: for elbow flexion and wrist extension, C1 attaches to palmar (bottom) handle attachment site; for wrist flexion, C3 attaches to dorsal (top) handle attachment site; for forearm pronation, C3 attaches to radial (left) handle attachment site. Rotation of the handle 90° about the long axis of the forearm permits testing in the neutral forearm position. With the right forearm in neutral pronation–supination, the chain configurations needed to test maximum voluntary contractions of the eight UE muscles are as follows: for elbow flexion, C1 attaches to ulnar (bottom) handle attachment site; for wrist extension, C4 attaches to palmar (left) handle attachment site; for wrist flexion, C2 attaches to dorsal (right) handle attachment site; for forearm pronation, C2 attaches to radial (top) handle attachment site.

by attaching C3 in Fig. 2 to the radial (or left) attachment site of the handle. The subject was then instructed to turn the palm downward. In the neutral forearm position, the handle in Fig. 2 was rotated 90° about the long axis of the forearm so the opening in the handle could accommodate the breadth of the hand. In this case, the chain and pulley configurations are shown in Fig. 2 for testing elbow flexion (C1), wrist flexion (C2) and wrist extension (C4). Forearm pronation was tested in the neutral forearm position by attaching C2 to the radial (or upper) attachment site of the handle.

EMG data were collected at a sampling rate of 1000 Hz with a system that utilizes a differential amplification method and possesses a gain of 1400, a CMRR of 130 dB, an input impedance of 10 GΩ, an 8th-order low pass Bessel anti-aliasing input filter with a corner frequency of 1000 Hz, and a high pass filter of 25 Hz (Innovative Computer Solutions, Fresh Meadows, NY, USA). Although this recording arrangement might have permitted signal aliasing, we determined through frequency analysis that there was no appreciable noise in the frequency band between the highest signal frequency and the cutoff frequency of the lowpass filter [10]. Thus, a 1000 Hz sampling rate adequately captured the EMG signals obtained in this study (up to 450 Hz in the worst case) and the signals were not appreciably distorted by the negligible noise in our recordings. The EMG system was interfaced with an analog-to-digital data acquisition system consisting of an IBM PS/Value Point computer (IBM model 6387) and a multichannel A/D board (Model AM-TIO-64F-5, National Instruments, Austin, TX, USA) using a Microsoft Windows™ Version 3.1 operating system. LabVIEW® for Windows™ (Version 3.0, National Instruments) was used to control the simultaneous collection of EMG and load cell data. MATLAB® (version 4.2, the MathWorks, Natick, MA, USA) was used to perform root mean square (RMS) processing of the raw EMG signal with a 40 ms time constant of integration. Bipolar, silver–silver chloride pregelled disposable adhesive electrodes, 10 mm in diameter for each Ag–AgCl surface, were used (Classic Medical Products, Muskego, WI, USA).

2.2. Subjects

Thirty-nine healthy subjects, 11 males and 28 females, participated by informed consent according to the procedures for human subject experimentation at the Hospital for Joint Diseases, New York, USA. All participants were screened by a physical examination to rule out the presence of signs and symptoms of neuromusculoskeletal disorders of the dominant or preferred upper extremity. The average age of the subjects was 36.2 years (SD, 9.4) for the females and 36.8 (SD, 7.7) for the males. The subjects' average height was 165.4 cm (SD, 5.0) for the females and 176.8 (SD, 7.4) for the males. The average weight of the subjects was 63.4 kg (SD, 11.9) for the females and 75.2 kg (SD, 11.6) for the males. Two of the female subjects were left hand dominant. Therefore, 37 right and 2 left UE were tested in this study.

2.3. Procedure

Following cleansing of the skin with rubbing alcohol, pairs of electrodes were placed with an inter-electrode distance of 14 mm. The following electrode placement guidelines recommended by Soderberg [2] (indicated by the superscript a) or by Perotto [11] (indicated by the superscript b) were adopted. Eight electrode sites were used. For the biceps brachii (BB)[a], electrodes were placed 1/3 of the length from the cubital fossa to the acromion process. For the pronator teres (PT)[b], electrodes were placed 3 cm distal to the midpoint of a line connecting the medial epicondyle of the humerus and the biceps tendon. For the flexor carpi radialis (FCR)[b], electrodes were placed 4.5 cm distal to the midpoint of a line connecting the medial epicondyle of the humerus and the biceps tendon. For the flexor carpi ulnaris (FCU)[b], electrodes were placed 3 cm medial to the ulna at a point 1/3 of the length of the forearm from the elbow crease. For the extensors carpi radialis longus and brevis (ECR)[b], electrodes were placed 3 cm distal to the lateral epicondyle of the humerus with the forearm pronated. For the extensor digitorum communis (EDC)[a], electrodes were placed 1/4 to 1/3 of the distance between the midpoint between the radial and ulnar styloid processes as measured from the wrist dorsum and the olecranon process. For the extensor carpi ulnaris (ECU)[a], electrodes were placed 1/3 of the distance between the midpoint between the lateral epicondyle of the humerus and the olecranon process and the styloid process of the ulna. For the pronator quadratus (PQ)[a], electrodes were placed at the point of intersection between the end of the palmaris longus tendon and a line drawn 2.5 cm proximal and parallel to the volar skinfold of the wrist. EMG electrode placement was controlled across test sessions by recording careful measurements for each subject with a tape measure during the first test session and using these same measurements for the placement of electrodes in the subsequent two test sessions.

The mean of each muscle's RMS EMG signal obtained during three sequential repetitions of an isometric MVC were determined for two fixed upper limb positions. These positions consisted of: (1) a pronated forearm position (full forearm pronation, 15° of wrist ulnar deviation and 15° of wrist extension), and (2) a neutral forearm position (neutral forearm pronation–supination, neutral wrist radial–ulnar deviation and neutral wrist flexion–extension). The elbow was maintained in 60° of flexion for both UE positions. These positions were chosen based upon those previously determined for particular functional tasks to be investigated in a future study. A similar comparison could have been made with other UE positions.

The pronated forearm position was tested first. However, fatigue was avoided by allowing a sufficient rest period between test exertions. All muscles were monitored during three sequential 5 s repetitions of an MVC separated by a 45–60 s rest period. The BB was monitored during an elbow flexion exertion, the PT and PQ during a forearm pronation exertion, the ECU, ECR and EDC during a wrist extension exertion, and the FCU and FCR during a wrist flexion exertion. EMG system baseline noise was determined for each muscle while the subject relaxed with the UE supported. Each subject underwent the normalization procedure on 3 occasions at 2 week intervals over a period of 4 weeks.

The mean RMS EMG signal over a 1.0 s maximum force period was computed for the three MVC repetitions for each muscle. This period was determined automatically using MATLAB® as the earliest 1.0 s period during which the load cell output was maximized during the MVC. Fig. 3 shows the load cell and EMG output (both raw and RMS processed) of the ECR from a randomly chosen female subject during the performance of three sequential trials of maximum wrist extension in the forearm pronated position on the first test session. As this figure indicates, maximum load was consistent across trials, as was the mean EMG signal amplitude. The resulting maximum RMS EMG was adjusted for system baseline noise by subtracting the mean RMS EMG amplitude of the resting EMG signals for each muscle. This gave the denominator, or EMG normalization coefficient, typically used for the normalization of EMG signals obtained during functional task performance [3,6,7]. This procedure was carried out for each muscle in each of the two UE positions across the three test sessions.

2.4. Data analysis

The calculated EMG normalization coefficients were not normally distributed. Therefore, they were subjected to log transformation in order to fulfill the assumptions required for parametric statistical analysis. The log adjusted EMG normalization coefficients were tested for repeatability by computing the intraclass correlation coefficients (ICCs) of the values across test sessions. For this purpose, ICC model 3 for specific instances of the factor test session was used [12]. The log adjusted EMG normalization coefficients obtained in the two UE positions were compared by two-way analysis of variance (ANOVA) with the factors UE position (2 levels) and test session (3 levels) for each of the eight muscles. In order to control for type I error due to multiple statistical comparisons, the chosen level of significance was adjusted using the Bonferroni correction [13]. Therefore, the level of significance was obtained by dividing 0.05 by 8, resulting in a level of significance of 0.00625.

3. Results

The mean values for the log adjusted EMG normalization coefficients are summarized in Table 1 for both UE positions and for each of the three test sessions. The ICCs for the EMG normalization coefficients are summarized in Table 2. The test–retest ICCs of the log adjusted EMG normalization coefficients ranged from 0.04 for the ECU in the pronated position to 0.70 for the PT in the pronated position, indicating poor to moderate agreement across test sessions.

The results of the two-way ANOVAs are shown in Tables 3–10. Table 3 indicates that there was a significant difference between log adjusted EMG normalization coefficients in the two UE positions for the BB ($P = 0.0004$). Table 1 shows that the log adjusted EMG normalization coefficient of the BB was lower in the pronated forearm position than in the neutral forearm position. Neither the factor test session nor the UE

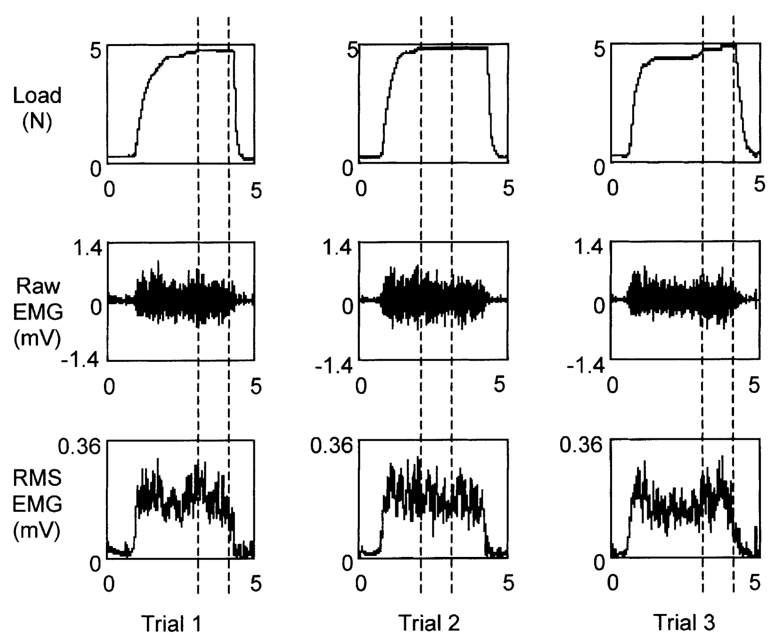


Fig. 3. Plots of load cell and EMG output from the extensors carpi radialis longus and brevis during a maximum voluntary wrist extension by a randomly chosen subject (female) showing three repeated trials on test session 1. The load cell signal is shown in the top row of plots. The raw EMG signal is shown in the middle row of plots. The root mean square (RMS) processed EMG signal is shown in the bottom row of plots. The vertical dashed lines depict the 1 s maximum force window over which the RMS EMG was averaged to obtain the maximum RMS EMG value used to determine the EMG normalization coefficient. The boundaries of the 1 s maximum force window were defined as the earliest 1 s period in which force was maximized during the entire 5 s data collection period.

Table 1
Summary of the log adjusted EMG normalization coefficients ($n = 39$) and standard deviations, in parentheses, for eight muscles of the elbow and forearm in two upper extremity positions over three separate test sessions[a]

Muscle	Neutral forearm position			Pronated forearm position		
	Session 1	Session 2	Session 3	Session 1	Session 2	Session 3
Biceps brachialis	2.351 (0.400)	2.396 (0.302)	2.432 (0.277)	2.160 (0.373)	2.225 (0.315)	2.322 (0.304)
Pronator teres	2.214 (0.367)	2.222 (0.268)	2.189 (0.300)	2.250 (0.323)	2.207 (0.303)	2.213 (0.244)
Pronator quadratus	1.818 (0.403)	1.841 (0.302)	1.741 (0.398)	1.863 (0.506)	1.888 (0.528)	1.825 (0.340)
Extensor carpi radialis	2.052 (0.637)	2.218 (0.251)	2.219 (0.273)	2.228 (0.595)	2.363 (0.256)	2.382 (0.267)
Extensor carpi ulnaris	2.257 (0.222)	2.300 (0.235)	2.263 (0.323)	2.182 (0.557)	2.295 (0.181)	2.239 (0.314)
Extensor digitorum communis	2.174 (0.347)	2.137 (0.398)	2.272 (0.256)	2.140 (0.589)	2.218 (0.246)	2.249 (0.226)
Flexor carpi radialis	2.441 (0.252)	2.399 (0.227)	2.477 (0.322)	2.383 (0.251)	2.455 (0.303)	2.424 (0.262)
Flexor carpi ulnaris	2.242 (0.224)	2.198 (0.200)	2.225 (0.166)	2.223 (0.198)	2.236 (0.192)	2.268 (0.159)

[a] Values were derived by subtracting the mean amplitude of the resting (baseline) RMS EMG signal (in mV) from the mean amplitude of the RMS EMG signal (in mV) averaged over three repetitions of an MVC of 1 s duration, then performing a log transformation on the difference.

Table 2
Intraclass correlation coefficients for log adjusted EMG normalization coefficients across test sessions for both the forearm pronated and forearm neutral upper extremity positions ($n = 39$)[a]

Muscle	Neutral forearm position	Pronated forearm position
	ICC (3,1)	ICC (3,1)
Biceps brachialis	0.54	0.51
Pronator teres	0.64	0.70
Pronator quadratus	0.52	0.35
Extensor carpi radialis	0.22	0.28
Extensor carpi ulnaris	0.42	0.04
Extensor digitorum communis	0.35	0.14
Flexor carpi radialis	0.37	0.33
Flexor carpi ulnaris	0.64	0.55

Note: ICC below 0.75 are indicative of poor to moderate reliability.[a] Model 3 for specific measurements of the factor test session was used (ICC (3,1)).

position × test session interaction was significant for the BB ($P > 0.00625$). As shown in Table 4, there was a significant difference between log adjusted EMG normalization coefficients in the two UE positions for the ECR ($P = 0.0033$). Table 1 shows that the log adjusted EMG normalization coefficient of the ECR was lower in the neutral forearm position than in the pronated forearm position. Neither the factor test session nor the UE position × test session interaction was significant for the ECR ($P > 0.00625$). Tables 5–10 indicate that there were no other significant main effects or interactions for the remaining six muscles tested ($P > 0.0625$).

4. Discussion and conclusions

The only muscles whose log adjusted EMG normalization coefficients were affected by UE position were the BB and the ECR. The log adjusted EMG normalization coefficient for the BB in the pronated forearm position was lower than that obtained with the UE in the neutral forearm position. Since a change in forearm orientation from neutral to full pronation would not change the relative location of the biceps muscle belly relative to the skin underlying the electrodes, the difference in EMG signal amplitude cannot be attributed to skin movement. It may reflect instead a reduction of the mechanical advantage of the biceps through excessive elongation with forearm pronation, especially since the elbow was flexed by only 60°. Basmajian and DeLuca reported that the contribution of the biceps brachii to elbow flexion is optimal in the range of forearm position from full supination to neutral pronation-supination and that the BB plays only a small role in elbow

Table 3
Two-way analysis of variance for the log adjusted EMG normalization coefficients of the biceps brachii with the factors upper extremity position (2 levels) and test session (3 levels)

Source of variance	Degrees of freedom	Sum of squares	Mean square	F	P
Upper extremity position	1	1.433	1.433	13.011	0.0004*
Test session	2	0.572	0.286	2.594	0.0769
UE position × Test session	2	0.067	0.033	0.302	0.7393
Error	225	24.786	0.110		

* Significant at Bonferroni adjusted alpha level of 0.00625.

Table 4
Two-way analysis of variance for the log adjusted EMG normalization coefficients of the extensors carpi radialis longus and brevis with the factors upper extremity position (2 levels) and test session (3 levels)

Source of variance	Degrees of freedom	Sum of squares	Mean square	*F*	*P*
Upper extremity position	1	1.511	1.511	8.808	0.0033*
Test session	2	1.250	0.625	3.644	0.0277
UE position × Test session	2	0.010	0.005	0.028	0.9720
Error	227	38.939	0.172		

* Significant at Bonferroni adjusted alpha level of 0.00625.

Table 5
Two-way analysis of variance for the log adjusted EMG normalization coefficients of the extensor carpi ulnaris with the factors upper extremity position (2 levels) and test session (3 levels)

Source of variance	Degrees of freedom	Sum of squares	Mean square	*F*	*P*
Upper extremity position	1	0.068	0.068	0.625	0.4299
Test session	2	0.235	0.118	1.080	0.3414
UE position × Test session	2	0.050	0.025	0.228	0.7966
Error	223	24.307	0.109		

Table 6
Two-way analysis of variance for the log adjusted EMG normalization coefficients of the extensor digitorum communis with the factors upper extremity position (2 levels) and test session (3 levels)

Source of variance	Degrees of freedom	Sum of squares	Mean square	*F*	*P*
Upper extremity position	1	0.004	0.004	0.027	0.8695
Test session	2	0.457	0.229	1.689	0.1871
UE position × Test session	2	0.152	0.076	0.561	0.5714
Error	222	30.057	0.135		

Table 7
Two-way analysis of variance for the log adjusted EMG normalization coefficients of the flexor carpi radialis with the factors upper extremity position (2 levels) and test session (3 levels)

Source of variance	Degrees of freedom	Sum of squares	Mean square	*F*	*P*
Upper extremity position	1	0.019	0.019	0.252	0.6162
Test session	2	0.060	0.030	0.406	0.6670
UE position × Test session	2	0.162	0.081	1.102	0.3341
Error	228	16.801	0.074		

Table 8
Two-way analysis of variance for the log adjusted EMG normalization coefficients of the flexor carpi ulnaris with the factors upper extremity position (2 levels) and test session (3 levels)

Source of variance	Degrees of freedom	Sum of squares	Mean square	*F*	*P*
Upper extremity position	1	0.026	0.026	0.717	0.3981
Test session	2	0.034	0.017	0.460	0.6320
UE position × Test session	2	0.047	0.023	0.638	0.5294
Error	226	8.280	0.037		

flexion when the forearm is pronated [1]. An early EMG study by Beevor showed this phenomenon [14]. Jamison and Caldwell more recently found that forearm supination facilitated BB activation, whereas forearm pronation inhibited BB activation [15]. This finding is consistent with our own results and supports the idea

Table 9
Two-way analysis of variance for the log adjusted EMG normalization coefficients of the pronator quadratus with the factors upper extremity position (2 levels) and test session (3 levels)

Source of variance	Degrees of freedom	Sum of squares	Mean square	F	P
Upper extremity position	1	0.201	0.201	1.136	0.2875
Test session	2	0.274	0.137	0.776	0.4614
UE position × Test session	2	0.018	0.009	0.052	0.9494
Error	225	39.722	0.177		

Table 10
Two-way analysis of variance for the log adjusted EMG normalization coefficients of the pronator teres with the factors upper extremity position (2 levels) and test session (3 levels)

Source of variance	Degrees of freedom	Sum of squares	Mean square	F	P
Upper extremity position	1	0.013	0.013	0.144	0.7042
Test session	2	0.037	0.018	0.200	0.8186
UE position × Test session	2	0.028	0.014	0.153	0.8581
Error	228	20.962	0.092		

that other elbow flexor muscles, namely the brachioradialis and brachialis, predominate during elbow flexion exertions with a pronated forearm. Such studies illustrate the substantial effect that muscle length changes and musculotendinous orientation can have on the EMG signal and strengthens the argument for using position specific EMG normalization coefficients. For example, if only the neutral forearm EMG normalization coefficient was used, biceps activation would be overestimated in that position as compared to the pronated position.

The log adjusted EMG normalization coefficient for the ECR in the neutral forearm position was lower than that in the pronated forearm position. Recall that in the neutral forearm position, the wrist was also in a neutral position, whereas in the pronated forearm position the wrist was extended 15° and ulnar deviated 15°. This latter combination forearm–wrist position may have provided the ECR with a mechanical advantage through optimization of the muscle's length–tension relationship. Additionally, the transition from the neutral to the pronated forearm position may have resulted in skin movement over the ECR leading to an alteration in the pick up area of that electrode site. This skin movement is one source of error that this normalization procedure was undertaken to control. Were only the normalization coefficient obtained in the pronated forearm position used, ECR activation would be underestimated in the neutral forearm position as compared to the pronated forearm position. In a worst case scenario, it is possible that cross-talk was detected by the ECR electrodes from the brachioradialis in the neutral forearm posture and from the EDC in the pronated forearm posture despite the care taken in electrode placement. The position-related discrepancy in ECR activation is consistent with this possibility, since the EDC participates in wrist extension but the brachioradialis does not [1]. The effect of such cross talk on task analysis would be most detrimental when attempts are made to distinguish ECR and EDC activation during wrist extension excursions with the forearm pronated. The availability of EMG data from electrodes placed over the EDC would be helpful in this regard, since EMG signals from two sets of surface electrodes that detect activity simultaneously from the same muscle often show similar phase and amplitude variations. Investigators should consider such measures in interpreting their data.

The poor to moderate agreement between log adjusted EMG normalization coefficients for the same muscles across the three test sessions illustrates the importance of normalization for comparisons of individual muscles over multiple test sessions. These findings were anticipated and confirm previous reports in the literature [16]. Even though the replacement of the surface EMG electrodes was controlled in this study through careful measurements across test sessions, the filtering and impedance characteristics of the subjects' skin and underlying tissues is an uncontrollable source of error in test–retest EMG measurements.

The results of the two-way ANOVAs comparing log adjusted EMG normalization coefficients suggest that alternating between the two UE positions used in this study did not have a deleterious effect on the amplitude of the MVC for the EDC, ECU, FCR, FCU, PQ, or PT muscles. Calculations based on the data from these six muscles indicated that with 39 subjects, the power to detect differences between the log adjusted EMG normalization coefficients in the two UE positions ranged from as low as 5.3% for the EDC to 17.5% for the PQ. Although one could interpret this result as a basis for repeating these experiments with very large numbers of subjects, we suggest alternatively that the UE position-dependent differences in the EMG normalization coefficients for these six muscles were so small as to be clinically meaningless. These results do not, however, guarantee that these same six muscles will behave similarly in other UE positions and should not be interpreted as such.

One limitation of this study design was that the order of testing in the two UE positions was not randomized. Because the pronated forearm position was always tested first, MVC exertions in the neutral forearm position could have been affected by muscle fatigue. Fatigue was controlled for by allowing subjects a rest period of 45–60 s between each MVC exertion. Several of our findings suggest that this rest period was sufficient. For example, as exemplified in Fig. 3, the force and EMG amplitudes were consistent across repeated trials of MVC within subjects. In addition, the BB exhibited a lower EMG amplitude in the pronated forearm position, whereas the ECR exhibited a lower EMG amplitude in the neutral forearm position. If fatigue had occurred, we would anticipate that the EMG amplitude would always be higher in the second (neutral) forearm position [17,18]. Finally, the lack of difference in EMG amplitude between the two UE positions of the other six muscles tested suggests further that muscle fatigue was not a confounding factor in this study.

EMG technique remains a useful means of comparing different task conditions among subjects performing functional tasks. Because such tasks may be performed in a variety of different limb positions, it is important to account for the potential artifacts in the EMG signal associated with skin movement of surface electrodes and muscle length changes. One way to partially ameliorate such artifacts is to normalize the task dependent EMG signals to position specific reference signals. When using the maximum voluntary contraction as the reference effort for distal UE activities, it is necessary to supply sufficient resistance for the multi-directional exertions about the elbow, forearm, and wrist joints while maintaining limb position and to accomplish the procedure in a reasonable period of time.

The present apparatus solved the problem of maintaining limb position and providing resistance. It also permitted the simultaneous measurement of tensile load and EMG and allowed for the performance of 36 separate isometric exertions of 5 s duration within approximately 45 min with only one limb position change. The apparatus can also be used to obtain submaximal voluntary contractions through reliance on the real time load cell output for visual feedback. Furthermore, the ability to sample force information is helpful to those investigations where estimates of muscle forces from EMG output are sought.

The use of the reference EMG signals obtained during this procedure will reduce errors in comparisons of muscle activation intensities during tasks performed in different UE positions. This is important in studies where functional tasks may substantially alter limb position. Mirka, for example, showed that errors as high as 75% resulted from the use of EMG normalization signals of the lumbar erector spinae when obtained at a trunk angle that differed from that of the functional task [19]. Miaki et al. showed that both ankle and knee angle affect the amplitude of the EMG from the soleus during MVC of the triceps surae [20]. Since the soleus does not cross the knee, this study suggests that both mechancial and motor control factors may influence intensity of muscle activation when limb orientation is altered.

Although every muscle's EMG normalization coefficient was not affected by UE position in this study, concern for relative skin movement or changing mechanical function of individual muscles warrants using the methodological approach described here when limb positions are anticipated to vary substantially between the functional tasks to be compared. Furthermore, the grouped analysis can only estimate general trends within a population, whereas certain individuals may be affected differently by limb position changes. By using the method presented here, the pick up area of the surface electrodes is consistent between the task and the reference EMG signals used for normalization for each individual subject.

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