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A non-MVC EMG normalization technique for the trunk musculature: Part 1. Method development

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Abstract

Normalization of muscle activity has been commonly used to determine the amount of force exerted by a muscle. The most widely used reference point for normalization is the maximum voluntary contraction (MVC). However, MVCs are often subjective, and potentially limited by sensation of pain in injured individuals. The objective of the current study was to develop a normalization technique that predicts an electromyographic (EMG) reference point from sub-maximal exertions. Regression equations predicting maximum exerted trunk moments were developed from anthropometric measurements of 120 subjects. In addition, 20 subjects performed sub-maximal and maximal exertions to determine the necessary characteristic exertions needed for normalization purposes. For most of the trunk muscles, a highly linear relationship was found between EMG muscle activity and trunk moment exerted. This analysis determined that an EMG-moment reference point can be obtained via a set of sub-maximal exertions in combination with a predicted maximal exertion (expected maximum contraction or EMC) based upon anthropometric measurements. This normalization technique overcomes the limitations of the subjective nature for the MVC method providing a viable assessment method of individuals with a low back injury or those unwilling to exert an MVC as well as could be extended to other joints/muscles. © 2001 Published by Elsevier Science Ltd.

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1. Introduction

Accurate documentation of the muscle activities between muscle or between conditions requires some type of normalization, which has typically been referenced relative to a maximum voluntary contraction (MVC) [1]. While the use of an MVC as the reference point for normalization may account for much of the potential variability among recording factors (e.g. skin impedance, electrode position, collection methods and devices, electrode size and pick-up area, etc.), the reproducibility of this reference point also depends upon the level of sincerity or motivation solicited during the exertion. The subjective nature of these exertions may introduce some level of experimental error.

Baratta et al. [2] attempted to overcome this problem through a method that required the subject to perform a maximum exertion, followed by a series of successive exertions that increase by 10%. Once the subject was no longer able to achieve a targeted exertion, the previous successful level was identified as the MVC. While this method was successful in reducing the subjective nature of MVC, it is limited in application because it applies only to healthy subjects and requires substantial rest periods and thus, significant time. In addition, Baratta's MVC method [2] as well as traditional MVC techniques would also have limited utility when evaluating individuals who are suffering from low back pain (LBP) since they may not be willing to generate "true" MVCs.

Keller et al. [3] found that pain during an exertion was the most powerful predictor of strength. Additionally, many researchers have found significant decreases in the strength in individuals who suffer from LBP as compared to individuals with no symptoms [4–12]. These differences may result from the sensation of pain and subsequent guarding or a diminished strength capacity due to disuse. For such individuals, it is unclear

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whether the lower exertion level was due to guarding or if it represents a "true" MVC. In an injured person, exertion sincerity may also be dependent upon the type of exertion, in that, pain may limit strength in one direction of effort more than another. In order to apply an EMG-assisted spinal loading model to an injured population, new normalization techniques must be developed. This study provides the theoretical basis as well as experimental evidence of one such technique.

1.1. Normalization technique

Many MVC normalization techniques assume that there is a linear relationship between force generated and EMG level (see Fig. 1). In other words, as force is uniformly increased, there is a corresponding linear response in the activity of the muscle up to the level of the maximum force exertion. Thus, the proportion of muscle activity for any force level below the MVC can then be represented as a percentage of the EMG level relative to the MVC. The linear assumption has been widely supported in the literature [13–18] although some researchers have found non-linear relationships [18–24]. It is important to determine whether the linear relationship assumption holds for the muscle of interest, in this case trunk muscles. As noted earlier, the MVC normalization method relies upon a "true" maximum exertion. However, as shown by the shaded region in Fig. 1, the MVC may vary depending upon the motivation or pain level of the individual. This variability may result in substantial MVC variability and influence the interpretation of the EMG signal.

The goal of any normalization technique is to identify a relative reference point that is consistent across muscles, across exertions, as well as across subjects. The reference point need not be a maximum exertion as long as it relates to the relative contribution of the muscle. Yang and Winter [25] used sub-maximal exertions to normalize muscle activity and found them to be more reliable than maximum exertions. One could use a sub-maximal reference exertion as long as a means exists to account for the inherent differences in force generating capacity for the various muscles. However, since individuals are stronger in certain directions of motion than others (e.g. extension>flexion>lateral flexion>axial twisting) [6,7,12,26], determining the actual relative level of effort for a given muscle in a multiple muscle system becomes problematic particularly for arbitrarily selected sub-maximum exertions.

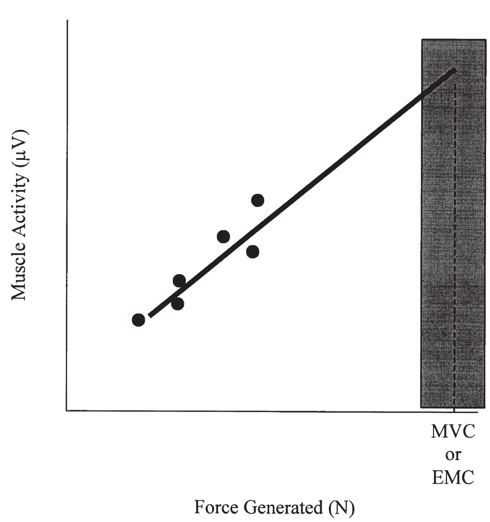

Fig. 1. Graph showing the linear relationship between forces generated and muscle activity with a reference point (either MVC or EMC) (●, muscle activity at a given exerted moment).

An alternative method of normalization might be to predict maximum moment generation capacity of a person and use a linear equation based upon a series of sub-maximal moments and muscle activities in order to establish the EMG–force relationship, thereby providing a stable EMG reference point. In other words, knowledge of the EMG–moment slope relationship in addition to the predicted MVC or EMC, one can help identify the maximum EMG level to be used for normalization purposes. Referring to Fig. 1, the dots represent the muscle activity at the various sub-maximal exertion levels while the diagonal line represents the linear slope of the EMG–moment relationship for these data points. Based on the derived equation, the maximum reference point would correspond to the activity for the predicted expected maximum contraction (EMC) (represented by the vertical line in the figure). Thus, utilization of sub-maximal exertions would allow normalization to be independent of the sincerity of the subject's exertion level.

The objective of this study was to establish whether this theoretical normalization technique is a viable method of EMG normalization for biomechanical modeling purposes. A companion paper [27] reports upon the ability of the EMC normalization method to interface with an EMG-assisted model that predicts spinal loads.

2. Methods

2.1. Approach

The development of this new normalization technique consisted of:

1. development of the moment prediction equations; and
2. the development of the normalization protocol that entails two parts;
   2.1. verification of linear EMG–force relationship for all trunk muscles used in our EMG-assisted model, and
   2.2. determination of necessary exertions needed for the normalization protocol.

2.1.1. Maximum moment (EMC) prediction

A database was constructed from studies performed over the past 5 years in the Biodynamics Laboratory. This database included individual anthropometric measurements (see variables in Table 1) as well as the maximum moments generated during MVC exertions (collected for normalization purposes for previous studies). Regression equations were developed to predict the maximum contraction moments or EMCs from various anthropometric measurements. These equations provide the estimated maximum exertion values that will be inputted into the muscle specific EMG–moment equations.

2.1.2. EMC Activity Prediction

2.1.2.1. EMG–force linearity A separate population of subjects performed several sub-maximal exertions as well as the maximum exertions used to normalize muscle activities [28]. EMG and exerted moment data collected during the *EMC Activity Prediction* phase served as a verification of the linear EMG–moment relationship for the ten trunk muscles evaluated in the EMG-assisted model.

2.1.2.2. EMC normalization protocol The EMG and moment data from these same subjects were also used to determine which exertions provide the best prediction of the EMC reference points. For each subject, EMG–moment relationships were established for each muscle by using the appropriate sub-maximal directional exertions (e.g. extension for erector spinae). These sub-maximal exertions provide points to establish the linear equations that predict the maximum activity given the predicted trunk moment that is based on subjects' anthropometry. This maximum activity provides the reference point for the EMC normalization technique.

In order to minimize the number of exertions, the utility of the various exertions was assessed. Unnecessary exertions were eliminated from the protocol. An exertion was determined to be unnecessary when a MVC for any of the ten trunk muscles occurred during the exertion $<20\%$ of the time. Minimizing the number of exertions reduced the potential effects of fatigue and injury when the technique is employed to assess injured subjects.

2.2. Subjects

The *EMC Prediction* database was constructed using MVC data from 120 subjects (63 males and 57 females). These subjects were recruited from the general university population. The subjects used in the *EMC Activity Prediction* portion of the study consisted of 20 male students. Complete anthropometric data for the subjects recruited for the study are shown in Table 1. All subjects were asymptomatic for LBP in the previous year.

2.3. Apparatus

Electromyographic (EMG) activity was collected through the use of bi-polar silver–silver chloride electrodes that have a 4 mm diameter and were spaced ca. 3 cm apart. Electrodes recorded activity at the ten major trunk muscle sites consisting of: right and left muscle pairs of erector spinae; latissimus dorsi; rectus abdominis; external obliques; and internal obliques. The electrodes were 4 mm in diameter. The locations of the electrodes were as follows:

Table 1
Anthropometry of the subjects used to develop the EMC normalization technique

	EMC prediction				Maximum activity prediction	
	Males (63)		Females (57)		Males (20)	
	Mean	SD	Mean	SD	Mean	SD
Age (years)	23.1	2.8	22.3	3.6	24.6	3.4
Weight (kg)	77.5	13.3	60.4	9.5	81.5	11.4
Standing height (cm)	177.5	7.3	165.6	6.2	179.7	7.1
Shoulder height (cm)	144.2	17.0	137.5	5.9	144.2	23.0
Elbow height (cm)	106.9	13.1	102.2	4.4	110.5	5.5
Upper leg length (cm)	44.7	4.7	44.7	7.1	44.5	7.2
Lower leg length (cm)	55.1	3.9	51.2	4.1	55.7	4.0
Upper arm length (cm)	36.3	5.0	34.5	1.9	37.3	3.1
Lower arm length (cm)	47.1	5.6	43.7	2.3	48.2	2.5
Spine length (cm)	60.5	6.6	52.7	7.5	59.1	3.3
Trunk depth at iliac crest (cm)	22.4	6.0	18.0	2.5	23.6	5.4
Trunk breath at iliac crest (cm)	29.3	3.5	26.8	3.1	30.3	3.8
Trunk depth at xyphoid process (cm)	22.2	3.2	19.2	2.1	22.8	3.1
Trunk breath at xyphoid process (cm)	31.1	6.6	26.4	1.7	31.0	2.0
Trunk circumference (cm)	77.9	17.2	73.3	9.5	73.3	24.1

1. erector spinae muscles — over the largest muscle mass found by palpation and ca. 4 cm from midline of the spine at the third lumbar vertebrae (L3);
2. latissimus dorsi muscles — at the most lateral portion of the muscle at the ninth thoracic vertebrae (T9);
3. rectus abdominus muscles — 3 cm from the midline of the abdomen and 2 cm above the umbilicus;
4. external oblique — 10 cm from the midline of the abdomen and 4 cm above the ilium at an angle of 45°; and
5. internal oblique muscles — 4 cm above ilium in the lumbar triangle at an angle of 45° [29].

The raw EMG signals were pre-amplified, high-passed filtered at 30 Hz, low-passed filtered at 1000 Hz, rectified, and smoothed with a low pass filter of a 20 ms sliding window.

During the static MVC and sub-maximal exertions, subjects were restrained at the shoulder through the asymmetric reference frame (ARF) (Fig. 2) [28,30,31] and at the pelvis by the pelvis support structure (PSS) (Fig. 2) [32]. The ARF provided static resistance against the upper body. The PSS was directly attached to a force plate (Bertec 4060A, Worthington, USA) at the base and restricted pelvic and lower body motion. The forces and moments measured at the center of the forceplate were translated and rotated to L5/S1 by knowing the relative position in three-dimensional space [32]. A computer was employed to display the real-time moments about L5/S1 for all exertions.

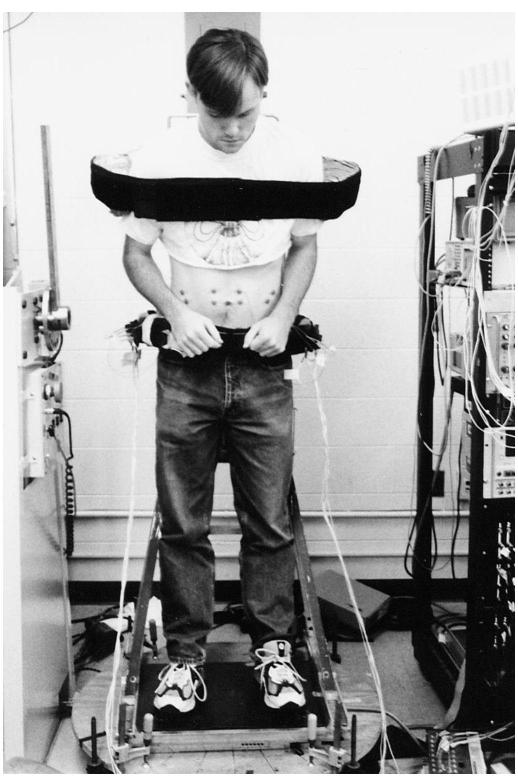

Fig. 2. A subject performing extension exertion within the PSS and ARF.

All signals from the aforementioned equipment were collected simultaneously through customized Windows™-based software developed in the Biodynamics Laboratory. The processed signals were collected at 100 Hz and recorded on a portable computer via an analog-to-digital converter.

2.4. Procedure

EMC Prediction data were gathered from MVC exertions of existing studies collected in the Biodynamics Laboratory. The exertions included: sagittal flexion; right and left lateral bending; and clockwise and counter-clockwise twist in the upright position (e.g. 0° flexion), and sagittal extension with the trunk at a 20° forward flexion angle (described by Marras and Mirka [28]). Each exertion was performed with 2 min of rest between exertions to reduce the effects of fatigue [33]. In addition, anthropometric measurements were taken (see Table 1 for the complete list).

Subjects who participated in the *EMC Activity Prediction* portion of the study performed the MVC exertions as well as a set of sub-maximal exertions in each of the test exertion directions (e.g. extension, flexion, right and left lateral flexion, and clockwise and counter-clockwise twist). Anthropometric measurements were collected and a consent form approved by the University Institutional Review Board was signed. Surface electrodes then were applied using standard placement procedures for muscles of interest [29,34,35]. Skin impedances were kept below 100 KΩ. The subject then was placed into the PSS and ARF where they performed the MVC followed by the sub-maximum exertions that were completed in random order.

The sub-maximal exertions included low-level target values as well as subjective (e.g. half of maximum) exertions. During the low-level moment exertions, the subjects controlled the exertion through feedback using a computer monitor that displayed the appropriate measured moment from the force plate. Subjects maintained the isometric exertion for 3–4 s in a window with a tolerance of ±3%. The low level exertions were dependent on the direction of applied moment. In order to utilize the proposed normalization method in evaluating individuals with LBP, low levels of moments in each of the directions were introduced to minimize the risk of injury as well as limit the influence of potential pain. The levels for extension and flexion were 40, 60, and 80 Nm, for right and left lateral flexion were 30, 60, 90 Nm and for clockwise and counter-clockwise twist were 10, 20, and 30 Nm. The subjective conditions required the subjects exert to the level in which they perceived to be one-third, one-half, and two-thirds of their maximum ability in each of the directions. During these subjective exertions, subjects were not given visual feedback of the exerted moments. All exertions were followed by 2 min of rest [33].

2.5. Data analyses

Multiple linear regression techniques were used to predict the maximum trunk moments exerted from the anthropometric measurements. The moments of interest were: sagittal moment during extension and flexion exertions; lateral moment during right and left lateral flexion exertions; and axial moment, lateral-axial resultant moment, and sagittal-lateral-axial resultant moment during clockwise and counter-clockwise twisting. The final regression models for each of these trunk moments corresponded to the "best" five-factor model resulting in the highest r^2 values. For each moment, an exhaustive search of all five variable combinations as well as stepwise regression techniques was completed using SAS [36] statistical software. The inclusion of additional variables into the models was found to add limited explanatory value, thus, the models were limited to five variables. Moment prediction models based on theoretical considerations such as height, weight, and age (factors known to relate to strength) were initially entered into the models a priori. These variables were also chosen in stepwise regression procedures.

Univariate linear regression techniques were used to evaluate the linear relationship among the various muscles activities and corresponding trunk moments. Descriptive statistics of the level of association between muscle activities and trunk moments were computed across subjects (e.g. mean and standard deviations of the correlations). The set of directional exertions needed for the normalization technique was determined by evaluating the effectiveness of predicting the maximum predicted moments (based on regression equations) as well as determining where maximum values of activity occurred.

3. Results

3.1. EMC Prediction

The best five-variable models based on anthropometry found for each of the trunk moments are shown in Table 2. When gender entered the models, the coefficient was negative indicating lower strength for females. The positive regression coefficients for standing height indicated increased strength for taller individuals. The negative regression coefficients for body weight indicated that heavier individuals had decreased strength when other variables were controlled for in the regression model. The predictability of all the models was moderate with a consistent trend in the amount of variability explained across all models (see correlations in Table 2).

3.2. EMC Activity Prediction

The utility of the *EMC Prediction* regression models was confirmed by evaluating the maximum measured moments obtained during the *EMC Activity Prediction*. While the predicted maximum moments were often found to be greater than actual measurements, the percentage difference between predicted and measured was consistent across all trunk moments. The average difference across all moments was an over-prediction of about 35.6%. For a given moment direction, the average error between exerted and predicted moments ranged from 1 to 52%.

3.2.1. EMG–force linearity

Strong linear relationships were found between the trunk muscle activities and primary trunk moments (Table 3). The strongest linear relationships were found for the erector spinae, rectus abdominus, and external and internal oblique muscles during flexion and extension exertions ($r>0.92$). The correlations for the EMG–moment models for all the muscles when exerting as an agonist (indicated as bold letters in Table 3) were above 0.61, indicating that the assumption of a linear relationship between moment and EMG holds for the trunk muscles of interest. Antagonistic muscles were also found to have a linear relationship between moment exerted and muscle activity ($r>0.54$), indicating increased coactivity with increased force exerted.

3.2.2. EMC normalization protocol

In order to develop a normalization technique that reduces the potential for fatigue or injury when evaluating an injured population, any exertions with limited utility were eliminated to minimize the number of exertions (e.g. limit the number of directions for the normalization exertions). The majority of the maximum moment values during the MVC exertions occurred in either flexion or extension (63%). The MVC maximum values for the specific muscles occurred only 13.5% of the time in both the left and right lateral bending directions while the twisting exertions resulted in maximum values 23.5% of the time. Thus, the lateral exertions resulted in the lowest percentage of maximum values and would have only increased the magnitudes of the values by 14% for the corresponding muscles (e.g. 13.5% of the MVC values). Based on this data, the sub-maximum exertions in extension, flexion, and clockwise and counter-clockwise twist were considered to be the most appropriate set of exertions in obtaining reference values while minimizing the number of exertions.

Table 2
Moment prediction regression results for the three-dimensional exerted trunk moments developed in the *EMC Prediction* phase

Trunk moment direction	Regression equation[a]	Correlation
Sagittal moment for *extension*	−1076.39+13.74×STHT−123.92×GNDR−3.16×WGHT+4.05×ELHT−15.23×LLL	0.460
Sagittal moment for *flexion*	−920.02+11.57×STHT+2.98×ELHT−15.05×LLL−17.06×TBI+2.25×TC	0.433
Lateral bending moment for *right lateral bending*	−589.93+8.72×STHT−78.13×GNDR−1.83×WGHT−9.71×LLL+1.53×TC	0.468
Lateral bending moment for *left lateral bending*	−240.85+6.87×STHT−62.25×GNDR−1.41×WGHT−6.43×LLL−3.75×ULL	0.382
Axial moment for *clockwise twist*	−47.43+0.57×STHT+−16.48×GNDR−.20×WGHT−0.89×TDI+2.71×TDX	0.357
Lateral−axial resultant moment for *clockwise twist*	−69.29+1.87×STHT−43.02×GNDR−0.71×WGHT+3.24×AGE−2.36×ULL	0.383
Sagittal−lateral−axial resultant moment for *counter−clockwise twist*	−823.11+3.98×STHT−2.01×WGHT+6.40×AGE+4.47×LAL+11.96×TDX	0.363
Axial twist for *clockwise twist*	−228.77+1.26×STHT−0.52×WGHT−0.51×ULL+1.32×LAL+5.19×TDX	0.394
Lateral−axial resultant moment for *counter−clockwise twist*	−237.52+3.27×STHT−30.09×GNDR−0.72×WGHT+0.82×ELHT−3.68×LLL	0.380
Sagittal−lateral−axial resultant moment for *counter−clockwise twist*	−596.39+7.47×STHT+5.19×AGE−8.42×LLL−11.78×TBI+1.67×TC	0.454

[a] STHT, standing height (cm); GNDR, gender (males=0, females=1); AGE, age (years); WGHT, weight (kg); ELHT, elbow height (cm); LLL, lower leg length (cm); ULL, upper leg length (cm); TC, trunk circumference (cm); TBI, trunk breadth at illiac crest (cm); TDI, trunk depth at illiac crest (cm); TDX, trunk depth at xyphoid process (cm); and LAL, lower arm length (cm).

Table 3
Summary of the correlations between individual trunk muscles and exerted trunk moments during *Maximum Activity Prediction* (bold correlations indicate agonistic muscle for the corresponding moment generated)

Trunk moment direction	Trunk muscle									
	LLAT	RLAT	LES	RES	LABS	RABS	LEO	REO	LIO	RIO
Sagittal moment for *extension*	0.81	0.83	**0.95**	**0.95**	0.73	0.72	0.86	0.83	**0.92**	**0.93**
Sagittal moment for *flexion*	0.80	0.84	0.73	0.66	**0.96**	**0.95**	**0.95**	**0.96**	0.83	0.86
Lateral bending moment for *right lateral bending*	0.71	0.84	0.66	0.73	0.73	0.82	0.77	**0.90**	0.66	**0.85**
Lateral bending moment for *left lateral bending*	**0.82**	0.76	0.70	0.64	0.89	0.79	**0.92**	0.81	**0.86**	0.67
Axial moment for *clockwise twist*	0.65	**0.81**	0.66	0.68	0.75	0.75	0.83	**0.83**	**0.68**	0.84
Lateral-axial resultant moment for *clockwise twist*	0.61	**0.76**	0.58	0.59	0.68	0.71	0.80	**0.79**	**0.61**	0.76
Sagittal-lateral-axial resultant moment for *clockwise twist*	0.60	**0.67**	0.64	0.69	0.78	0.76	0.75	**0.80**	**0.71**	0.75
Axial twist for *counter-clockwise twist*	**0.84**	0.80	0.72	0.61	0.79	0.79	0.90	**0.90**	0.85	**0.76**
Lateral-axial resultant moment for *counter-clockwise twist*	**0.80**	0.75	0.62	0.54	0.77	0.80	**0.85**	0.87	0.81	0.72
Sagittal-lateral-axial resultant moment for *counter-clockwise twist*	**0.76**	0.74	0.69	0.58	0.83	0.80	**0.84**	0.82	0.77	**0.70**

4. Discussion

This study describes a normalization method that overcomes potential limitations of traditional MVC methods. The EMC normalization technique provides a viable reference point for individuals that may not exert sincerely, in particular, individuals with an injury. This method minimizes the effects of guarding on the reference point prediction since increased guarding would increase coactivity at higher exertion levels. In addition, it minimizes the subjective nature of the exertions due to motivation.

The EMC normalization technique relied upon several underlying relationships. First, we assumed there was a relationship between anthropometric measurements and the exerted trunk moments. The regression equations developed to predict the exerted trunk moments in the six directions of interest (e.g. extension, flexion, right and left lateral bending, and clockwise and counter-clockwise axial twisting) did not explain a large portion of the variability in maximum moment. The lack of accountability of the variability in moment may have resulted from the subjective nature of the MVC exertions. While all possible steps were taken to reduce the potential of non-maximum exertions (e.g. verbally encouraged, visual verification of moment exerted, screened for LBP, etc.), motivation, pain, or other personal factors may still have limited the exertion levels. As expected, this finding may emphasize the highly variable nature of MVCs. The low percentage of variability explained in these models may also reflect the extremely complicated relationship between strength and personal factors. Several other authors have found similar correlations with single anthropometric variables and various trunk strength measurements [37–41]. Mital and Ayoub [42] reported a multivariate regression model that accounted for 46% of the variability in static extension strength. However, one of the variables in the model was static arm strength, indicating that "true" anthropometric measurements would account for less of the variability. Other authors have found leg strength to be mildly related to various anthropometric measurements [43–45]. Thus, the predictability of the current regression models was similar to strength prediction models found in the literature.

These models were also found to be very consistent across the various trunk moments. Standing height was a good predictor of the moment exerted for all the models while weight and gender were also found to be significant predictors in many models. An unexpected trend occurred for body weight in that strength decreased with increased weight. This trend may actually represent the effect of body fat on the strength of the individual, that is, body weight may be a surrogate for non-muscle mass tissue when the other variables in the models are taken into account.

The EMC normalization and MVC normalization techniques rely upon the assumption of a linear relationship between muscle activity and exerted trunk moment. In the current study, the sub-maximum exertions yielded a highly linear moment–EMG relationship with correlations being >0.73 and many exceeding 0.9. The lower correlations were found for the twisting exertions where the exerted trunk moments were much more complex. For extension, flexion, and lateral bending exertions, the primary moment (e.g. sagittal moment during extension and flexion or lateral moment during lateral bending) comprised a greater percentage of the resultant moment (exceeding 90%) while the axial moment during twisting exertions accounted for only 45% of the overall moment. A linear relationship between muscle activity and force exerted has been widely documented for a wide variety of muscles [13–15,17,18,46]. It appears that a linear relationship between muscle activity and force exerted also holds for the trunk muscles evaluated in the current study.

Based on this linear assumption and predicted moment regression models, the reference points for each of the ten muscles were predicted. A set of sub-maximal exertions provided the linear EMG–moment relationship that was extrapolated to the predicted exerted maximum moment resulting in the reference point for the specific muscle. The utility of providing an appropriate reference point indicated the lateral bending exertions could be eliminated from the set of exertions needed for the EMC normalization technique. This allowed the number of exertions to be minimized in order to reduce the chances of fatigue and further injury when the technique is applied to evaluate an injured population. The number of exertions in each direction was chosen to be six for the EMC normalization technique with three low-level and three subjective exertions. It was determined that these six exertion levels provided sufficient variability in exerted moment to accurately represent the linear relationship between moment and muscle activity. For several subjects, the trunk moments during the subjective exertions were very close in magnitude.

During the *EMC Activity Prediction* phase, the predicted moments were found to be greater than the actual MVC by an average of 35%. In support of this finding, Baratta et al. [2] found that subjective MVC exertions were 25–30% below the "true" MVC. This indicates that the predicted values might actually be estimates of a "true" MVC. Further investigation of the moment comparisons revealed that five subjects provided maximum exertions that were significantly lower than that predicted (between 66 and 125% over-prediction). When these five subjects were removed from the summary statistics, the average over-prediction error was 14.6% with all values below 35%. The differences between the measured and predicted maximum trunk moments for all the directions were around the same relative level. These differences may be reflective of the wide range of motivations when performing MVCs in either the original *EMC Prediction* or *EMC Activity Prediction* datasets.

In order to properly utilize this method, one must understand a few of the limitations of the method and its development. One must recognize that the moment prediction equations explained a small proportion of the overall variability in trunk moment. But, this is probably comparable to the variation associated with sincerity. In addition, the current method could be easily adjusted to incorporate better predictive models, possibly including factors such as muscle size, muscle fiber type, and physical fitness of the individual. The current regression models do, however, seem to provide a similar relative level of moment for the various exertion directions.

During the *EMC Activity Prediction* phase, the subjects were all males. The objective of this phase was to verify the linear relationship between muscle activity and moments exerted as well as develop the method protocol. While it is possible that the linear relationships may not be appropriate for females, females were tested in the validation part of this study (published in the companion paper [27]) and were found to have a linear EMG–moment relationship.

Finally, the EMC protocol seems to rely heavily on the fact that a linear relationship exists between muscle activity and moment exerted. While this method has been initially developed using this relationship, the method could be easily adapted to incorporate more complex relationships such as higher-order polynomial or exponential relationship. One must only determine which is the most appropriate relationship to adopt when predicting the EMC value. Thus, this method could be adapted to muscles groups that exhibit non-linear EMG–force trends.

5. Conclusion

The current study provides the theoretical foundation of a normalization technique (EMC) other than the traditional MVC method. An EMG–moment reference point can be obtained by completing a set of sub-maximal exertions combined with an EMC moment that is based on anthropometric measurements. The EMC normalization technique overcomes the limitations of the subjective nature for the MVC method, particularly when evaluating individuals with injuries. A companion paper [27] provides an assessment of the utility of this method when evaluating muscle activities and predicting spinal loads. Furthermore, the method could easily be adjusted to predict reference points for other muscles and joints as well as account for non-linear EMG–force relationships.

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