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A non-MVC EMG normalization technique for the trunk musculature: Part 2. Validation and use to predict spinal loads

W.S. Marras *, K.G. Davis , A.B. Maronitis

The Biodynamics Laboratory, The Ohio State University, 210 Baker Systems, 1971 Neil Avenue, Columbus OH 43210, USA

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Abstract

Estimates of the amount of force exerted by a muscle using electromyography (EMG) rely partially upon the accuracy of the reference point used in the normalization technique. Accurate representations of muscle activities are essential for use in EMG-driven spinal loading models. The expected maximum contraction (EMC) normalization method was evaluated to explore whether it could be used to assess individuals who are not capable of performing a maximum exertion such as a person with a low back injury. Hence, this study evaluated the utility of an EMG normalization method (Marras and Davis, A non-MVC EMG normalization technique, Part 1, method development. Journal of Electromyography and Kinesiology 2000) that draws upon sub-maximal exertions to determine the reference points needed for normalization of the muscle activities. The EMC normalization technique was compared to traditional MVC-based EMG normalization by evaluating the spinal loads for 20 subjects (10 males and 10 females) performing dynamic lifts. The spinal loads (estimated via an EMG-assisted model) for the two normalization techniques were very similar with differences being <8%. The model performance variables indicated that both normalization techniques performed well (r^2>0.9 and average error below 6%) with only the muscle gain being affected by normalization method as a result in different reference points. Based on these results, the proposed normalization technique was considered to be a viable method for EMG normalization and for use in EMG-assisted models. This technique should permit the quantitative evaluation of muscle activity for subjects unable to produce maximum exertions.

Keywords: Electromyography; Spinal loading; Anthropometry; Strength; Low back injury

1. Introduction

Electromyography (EMG) has become a valuable tool for the evaluation of the activity of the musculoskeletal system. Initially, the magnitude of EMG activity was used as an indicator of the level of force exerted [1–10] EMG data has also been used as an input into spinal loading models that predicts muscle forces [11–21]. In order to accurately estimate muscle forces, these models must account for the complex relationship between muscle activity, length and velocity of the muscle, cross-sectional area of the muscles, and muscle force capacity.

Traditionally, accurate representation of the muscle activity has relied upon EMG normalization relative to a maximum voluntary contraction (MVC). In a companion paper, Marras and Davis [22] proposed a new protocol that incorporates sub-maximal exertions and the estimation of the expected maximum contraction (EMC) to predict a reference point to be used for normalization. The sub-maximal exertions provide a range of moments with corresponding muscle activity levels. Based on these moments and activity levels, an EMG-force(moment) relationship was determined. A reference point (EMC) was determined when the predicted maximum moment was used in the resulting equation. Thus, this EMC technique does not require maximum exertions, allowing EMG assessments of individuals unable to provide a MVC.

Before the EMC normalization technique could be considered as an alternative to traditional MVC-based normalization, it must be demonstrated that the EMC normalized muscle activities would produce similar results in an EMG-assisted spinal loading model. It would be important that muscle activity results provide the same trends across a set of independent variables and

* Corresponding author. Tel.: +1-614-292-6670; fax: +1-614-292-7852.

E-mail address: marras.1@osu.edu (W.S. Marras).

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more importantly, the estimated spinal loads should be independent of the normalization method. Thus, the objective of this study was to investigate whether the EMC normalization technique was a viable method of normalization for biomechanical modeling purposes. The EMC reference values were evaluated by determining their effect on the fidelity of a biomechanical model as well as spinal load predictions. Since the MVCs for different muscles are independent of each other, the effectiveness of the EMC normalization is not just direct change in the model output variables. In other words, a complex EMG-assisted model relies heavily on the accurate estimation of the MVC, both for model fidelity and spinal load prediction.

2. Methods

2.1. Approach

The EMC normalization method was compared to actual MVC normalization by evaluating the model performance variables and predicted loads for individuals performing several lifting conditions. The normalization protocol developed in Marras and Davis [22] was employed to predict the reference points. For the same lifting conditions, both sets of reference points (EMC and MVC-based methods) were used to normalize the muscle activity and then inputted into an EMG-assisted spinal loading model [11–14,19,23–28].

2.2. Subjects

Twenty students (10 male and 10 female) who were asymptomatic for low back pain in the previous year participated in this study. These subjects were not part of the subject population in the companion paper [22]. Complete anthropometric data for the subjects recruited for the study are shown in Table 1.

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

	Males (10)		Females (10)	
	Mean	SD	Mean	SD
Age (years)	23.4	2.7	22.7	2.6
Weight (kg)	74.2	8.3	60.0	5.9
Standing height (cm)	177.2	6.2	165.2	4.7
Shoulder height (cm)	146.5	5.6	136.3	4.6
Elbow height (cm)	107.0	4.2	100.6	4.0
Upper leg length (cm)	41.1	4.2	43.0	3.1
Lower leg length (cm)	53.5	4.0	51.3	3.7
Upper arm length (cm)	37.4	2.4	33.9	1.1
Lower arm length (cm)	47.7	2.1	43.9	1.5
Spine length (cm)	59.6	2.0	54.0	4.3
Trunk depth at iliac crest (cm)	20.2	1.8	17.8	2.4
Trunk breath at iliac crest (cm)	28.7	1.8	25.7	2.6
Trunk depth at xyphoid process (cm)	21.2	1.6	18.1	3.0
Trunk breath at xyphoid process (cm)	30.2	1.5	26.3	1.0
Trunk circumference (cm)	82.9	5.8	71.3	6.8

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. The electrodes were placed at the ten major trunk muscle sites which consist of: right and left muscle pairs of erector spinae; latissimus dorsi; rectus abdominis; external obliques; and internal obliques [22,27]. 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 20 ms sliding window.

During the static MVC exertions and the sub-maximal exertions of the used as part of the EMC normalization protocol, subjects were restrained at the shoulder through the asymmetric reference frame (ARF) [7,8,29] and at the pelvis by the pelvis support structure (PSS) [30]. The ARF provided static resistance against the upper body during exertions. 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 are translated and rotated to L5/S1 by knowing the relative position in three-dimensional space [30]. A computer was employed to display the real-time moments about L5/S1 for all exertions.

During the dynamic lifting (test) exertions, subjects wore a lumbar motion monitor (LMM) that recorded the three-dimensional trunk motion. The LMM is essentially an exoskeleton of the spine in the form of a triaxial electro-goniometer that measured instantaneous three-dimensional position, velocity, and acceleration of the trunk. For more information on the design, accuracy, and application of the LMM, refer to Marras et al. [31]. Subjects lifted a box while controlling sagittal position from 55° of sagittal flexion to an upright position while their lower body was placed in the PSS. The subjects controlled their trunk velocity through the use of a computer display that displayed the real-time sagittal position of the trunk.

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. Study design

An EMG-assisted biomechanical model was utilized to compare the predicted spinal loads as well as model fidelity variables for the two types of normalization procedures (e.g. reference points obtained during the actual MVC exertions and based on the EMC maximum moment). The spinal loads were estimated using an EMG-assisted biomechanical model under development in the Biodynamics Laboratory over the past 18 years [11–14,19,23–28] and consisted of: the maximum values of compression, anterior–posterior shear, and lateral shear forces on the lower back at the lumbosacral joint (L5/S1). Model fidelity was evaluated by three model performance variables: muscle gain, r-square (r^2), and average absolute error (AAE) [18,24]. Muscle gain consists of the maximum force production capacity of the muscles represented in the model (physiological range of 30–100 N/cm^2) [24,32,33]. The r^2 and AAE variables provide an indication of the ability of the model to predict trunk moments relative to the moments that are measured by the experimental apparatus. The r^2 values provide information about how well the measured trunk moment “trend” matches the predicted trunk moment “trend” while AAE indicates how well the magnitude of these two “trends” matches.

2.5. Procedure

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 the subjects preformed the MVC exertions as well as the set of exertions required for the EMC-based normalization protocol (e.g. extension, flexion, and right and left twist).

After the necessary exertions for both normalization methods were completed, the subjects performed several sagittally symmetric dynamic lifts while positioned in the PSS (the pelvis was fixed). The subjects lifted a box weighing 6.81 and 13.63 kg at several trunk velocities (15, 30, 45, and 60 deg/s).

2.6. Data analyses

Repeated-measures analysis of variance (ANOVA) statistical analyses were performed for the model fidelity and spinal load variables to explore whether differences exist between the two types of normalization techniques for the dynamic lifting trails. For all significant independent variables, post-hoc analyses, in the form of Tukey multiple pairwise comparisons were performed to determine the source of the significant effect(s).

3. Results

The normalized muscle activities were significantly affected by the normalization method used for the left latissimus, right erector spinae, and left and right internal obliques ($\alpha<0.05$). The normalized activities for these muscles were lower for the EMC method than the MVC technique (by ca. 30%). Similar trends across the two box weight conditions were found for all the trunk muscles using both normalization techniques as seen by comparing Figs. 1 and 2. While there were slight deviations in the actual increases in muscle activity seen between the weight levels under each of the normalization methods, the overall picture of muscle activity was strikingly similar, that is, the relative change was the same for both techniques. The 27.2 kg weight resulted in about 30% higher muscle activities for the latissimus dorsi, external oblique, and internal oblique muscles than for the 13.6 kg weight while the erector spinae and rectus abdominus muscles increased by ca. 15%. The muscle activity responded similarly to two normalization techniques across the various trunk velocities.

The model performance and predicted spinal loads and trunk moments are shown in Table 2. The type of normalization technique used altered only one of the model performance variables (muscle gain). The EMC normalization technique resulted in higher gains than the MVC method, by about 30%. While there was a difference in muscle gain, the muscle gains for both methods were within physiological limits [24,32,33]. Both normalization methods resulted in the same r^2 and AAE values, above 0.9 and below 6.5%, respectively. Thus, the model performed extremely well under both normalization techniques.

There was no statistically significant effect of normalization technique on either the predicted sagittal trunk moment or the three-dimensional spinal loads (Table 2). The difference between the maximum predicted sagittal trunk moments between the two methods was <1 Nm (0.3%). The normalization method was found to affect maximum compression force by <50 N or 1.3%. Similar results were found for A–P shear forces (<1 N or 0.1%). The largest difference between the normalization methods was for lateral shear where the EMC-based technique predicted slightly higher loads than the MVC method. None of these differences were statistically significant for the main effect of normalization technique. The only significant interaction between normalization technique with gender, weight, or trunk velocity was for lateral shear force where there was a 29 N (22.5%) difference between normalization techniques for the males and no difference for the females.

This significant difference in lateral shear force between methods found for males did not affect any of the trends for the other independent variables (e.g. no significant three-way interactions). There were consist-

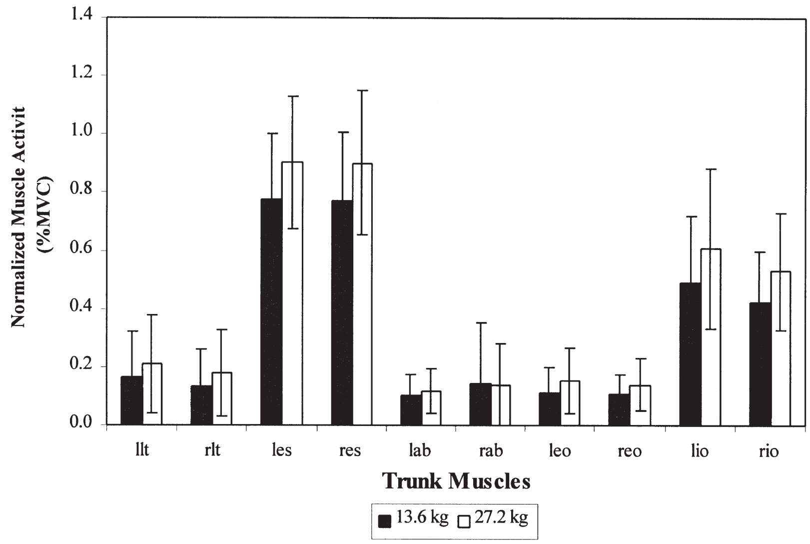

Fig. 1. Normalized muscle activity relative to the **MVC** reference values as a function of box weight (llt, left latissimus dorsi; rlt, right latissimus dorsi; les, left erector spinae; res, right erector spinae; lab, left rectus abdominus; rab, right rectus abdominus; leo, left external oblique; reo, right external oblique; lio, left internal oblique; rio, right internal oblique).

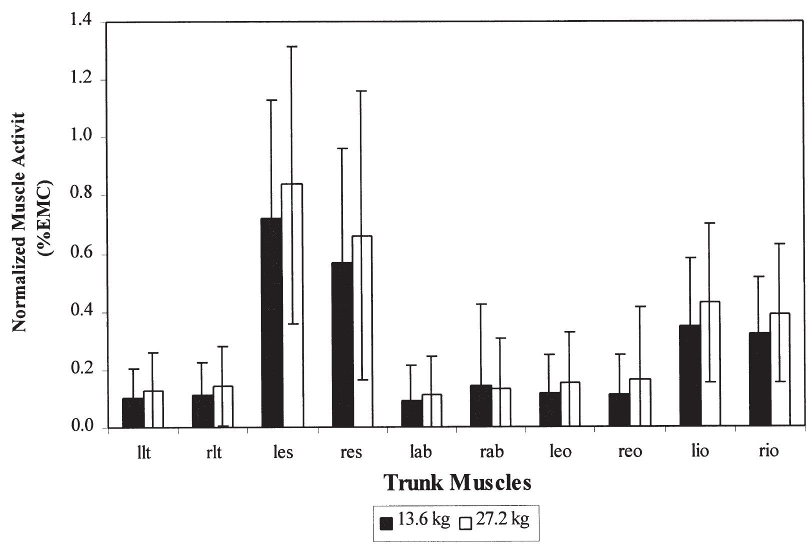

Fig. 2. Normalized muscle activity relative to the **Estimated Maximum Contraction (EMC)** reference values as a function of box weight (llt, left latissimus dorsi; rlt, right latissimus dorsi; les, left erector spinae; res, right erector spinae; lab, left rectus abdominus; rab, right rectus abdominus; leo, left external oblique; reo, right external oblique; lio, left internal oblique; rio, right internal oblique).

ent trends for loading across the velocity and weight conditions with differences between the two methods being within 6–8%, 1%, and 3% for lateral shear, A–P shear, and compression, respectively. A typical trial that was normalized to both methods can be seen in Fig. 3. Notice both normalization methods provide remarkably similar trends throughout the trial.

4. Discussion

The normalization technique evaluated in the current study was found to be equally proficient in predicting spinal loads and trunk moments as traditional MVC-based normalization. The only significant difference occurred in lateral shear (<30 N) and was apparent specifically for males. However, the trends across the other independent variables, box weight and trunk velocity, were found to be nearly identical between the two normalization techniques. The model performance variables for both normalization techniques indicated that the EMG-assisted model was robust in predicting the three-dimensional spinal loads.

The muscle gain values were slightly higher for the current normalization technique than the MVC method, but both were well within the physiological range

Table 2
Mean (standard deviations) for the model performance and spinal loading variables for the two normalization techniques (MVC and EMC EMG values) as a function of gender, trunk velocity, and box weight (bold indicates statistical significant effect due to normalization technique)

Independent variables		Gain (N/cm^2)	r^2	AAE per unit moment (%)	Maximum sagittal moment (Nm)	Maximum lateral shear force (N)	Maximum A–P shear force (N)	Maximum compression force (N)
Gender								
Male	MVC	**41.2 (19.9)**	0.91 (0.13)	6.2 (3.0)	150.0 (42.6)	**130.3 (69.1)**	740.3 (115.3)	3750.1 (749.4)
	EMC	**55.9 (23.2)**	0.91 (0.13)	6.5 (3.1)	151.1 (42.1)	**159.6 (80.8)**	767.7 (174.3)	3760.6 (815.6)
Female	MVC	**61.3 (27.5)**	0.92 (0.14)	5.9 (2.9)	142.4 (65.9)	154.8 (147.9)	737.0 (186.7)	3744.6 (1224.8)
	EMC	**79.4 (46.4)**	0.92 (0.13)	5.9 (2.8)	142.1 (66.5)	144.4 (152.0)	704.5 (155.1)	3628.4 (1164.5)
Trunk velocity								
15 deg/s	MVC	**55.6 (28.6)**	0.87 (0.17)	6.3 (2.6)	148.6 (52.7)	143.1 (121.7)	745.9 (170.6)	3788.8 (970.3)
	EMC	**73.7 (41.7)**	0.87 (0.16)	6.2 (2.5)	148.9 (52.5)	155.4 (130.0)	743.2 (177.7)	3739.3 (961.8)
30 deg/s	MVC	**52.1 (25.2)**	0.92 (0.14)	6.1 (3.0)	148.5 (55.5)	146.3 (105.8)	763.4 (144.1)	3778.0 (988.2)
	EMC	**68.6 (35.8)**	0.92 (0.12)	6.2 (3.0)	149.1 (56.1)	155.5 (112.6)	767.2 (175.2)	3734.0 (1000.6)
45 deg/s	MVC	**48.4 (25.2)**	0.94 (0.10)	5.6 (3.1)	143.6 (54.7)	138.4 (116.2)	701.4 (142.9)	3701.7 (1018.1)
	EMC	**64.2 (37.9)**	0.94 (0.10)	5.8 (3.1)	143.9 (54.6)	148.0 (121.8)	699.3 (161.2)	3642.4 (995.2)
60 deg/s	MVC	**46.5 (23.5)**	0.94 (0.13)	6.3 (3.3)	145.1 (57.1)	139.7 (112.7)	744.9 (148.7)	3723.5 (1036.5)
	EMC	**61.5 (35.2)**	0.93 (0.13)	6.5 (3.3)	145.7 (57.5)	151.3 (116.3)	743.1 (154.4)	3680.2 (1038.2)
Box weight								
13.6 kg	MVC	**49.8 (25.5)**	0.92 (0.15)	5.7 (2.7)	127.2 (47.0)	127.5 (96.5)	678.6 (113.1)	3298.3 (794.2)
	EMC	**64.8 (36.8)**	0.92 (0.15)	5.8 (2.8)	127.8 (47.1)	135.6 (99.9)	676.1 (125.0)	3264.6 (811.3)
22.7 kg	MVC	**51.5 (26.2)**	0.92 (0.12)	6.6 (3.2)	168.1 (55.0)	158.0 (128.7)	806.6 (163.3)	4254.1 (966.7)
	EMC	**69.3 (38.9)**	0.92 (0.11)	6.7 (3.1)	168.4 (55.4)	171.6 (136.5)	807.9 (183.2)	4187.6 (958.5)

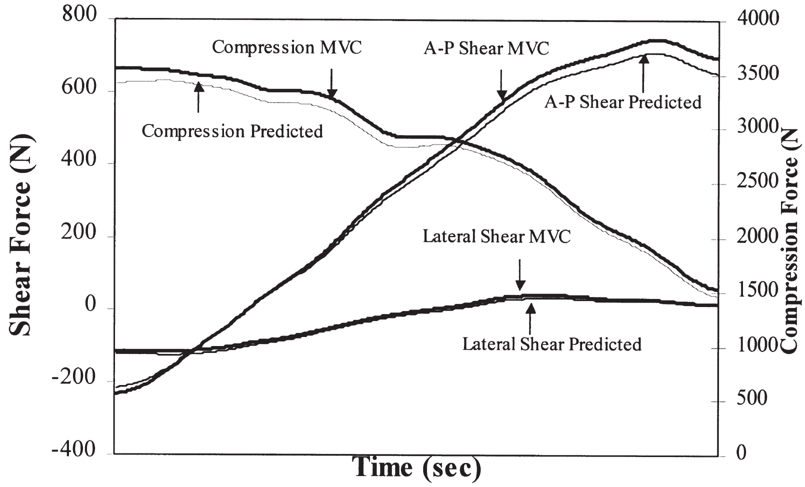

Fig. 3. A typical trial showing the instantaneous spinal loads for the two normalization techniques (MVC and EMC).

[24,32,33]. The difference in muscle gains was due to the overestimation of maximum moment (EMC vs MVC) that resulted in lower normalized muscle activity. The over-prediction for the current subject population of ca. 30% was similar to the over-prediction of MVC exertion levels found in the companion paper [22] as well as that published by Baratta et al. [36]. Baratta and associates [36] found that the "true" MVC was about 25–30% greater than the subjective MVC performed initially. Thus, the over-prediction of the MVC by the EMC is not surprising, and may reflect a more "true" estimate of the MVC. Muscle gain was estimated by comparing the predicted trunk moment to the measured moment. Increases in model gain estimation are not surprising. Any underestimation of the agonistic muscle activity such as the erector spinae, internal obliques and latissimus dorsi muscles would increase muscle gain. The predicted moment is calculated by summing the individual muscle forces multiplied by their corresponding moment arms from the spine. Muscle force is calculated by multiplying the normalized muscle activity by the cross-sectional area, the length–strength and force–velocity relationships and the muscle gain [11,13]. Muscle gain then is calculated by minimizing the difference between the measured and predicted trunk moments. Thus, there is a direct relationship muscle activity and muscle gain and an inverse relationship between muscle gain and the normalization reference point.

The difference in muscle gains between the two techniques had little effect on the prediction of the trends in trunk moments during the lifts. The r^2 ($>$0.91) and AAE ($<$6%) values for both normalization methods indicated the trend in predicted moment was very similar to the trend in measured moment. Thus, the EMC normalization procedure proved to be a viable way to normalize muscle activity when predicting spinal loads with the only exception being male lateral shear force.

The EMC normalization technique yielded remarkably similar trends in the trunk muscle activities when compared to MVC normalization across the various experimental parameters (e.g. box weight and trunk velocity). While the EMC normalization method resulted in lower muscle activities than for the MVC method, the relative trends were almost identical. Hence, the current method now provides a way to accurately evaluate the muscle activity of an individual is unable to exert MVCs. Sensation of pain may inhibit a MVC, causing activities normalized to these values to be highly variable where as one would expect the EMC method to remain stable. The sensation of pain during the exertion was found to be the most powerful predictor of strength [37]. The sub-maximal exertion method eliminates this inhibition by using the regression equations to predict "normal" exertion levels. This method now allows the evaluation of individuals with non-muscular based LBP to go beyond the evaluation of on/off patterns [38] as well as overcome the inaccuracies of evaluating non-normalized activities [39–42].

The current method evaluated effectiveness of sub-maximal normalization of muscle activity for trunk muscles. These same principles could easily be adapted to other joints by developing strength predictive models and measuring a series of sub-maximal exertions (in appropriate directions) to determine the linear slope between muscle activity and exerted moment. Thus, the underlying concepts of the method can be easily adopted to provide information about muscle exertion levels in a wide variety of situation where maximum exertions are unattainable.

5. Conclusion

The current study evaluated the potential of a sub-maximal EMC-based normalization technique as an alternative to the traditional MVC-based method. Muscle forces can be accurately estimated from the normalized muscle activity based on references points predicted by the protocol developed by Marras and Davis [22]. This normalization technique overcomes the limitations of the subjective nature for the MVC method. When applied to an EMG-assisted model, the model fidelity was minimally influenced by the use of the EMC normalization technique. Although muscle gain was slightly higher for the EMC normalization technique, spinal load prediction was nearly identical to the values obtained for the MVC normalization. Thus, the EMC normalization procedure provides a viable means of evaluating muscle activity and subsequent spinal loads and load patterns for an individuals who are not willing to provide accurate MVC exertions such individuals with a low back injury. This work may also provide a basis for EMC-based normalization for other parts of the body.

Acknowledgements

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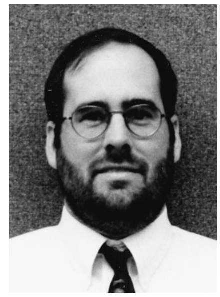
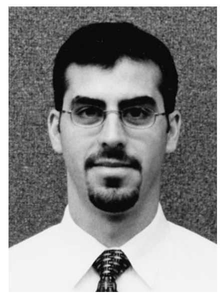
Professor William S. Marras holds the Honda Endowed Chair in Ergonomics in the Department of Industrial and Systems Engineenng at the Ohio State University. He is also the director of the Biodynamics Laboratory and holds joint appointments in the Departments of Physical Medicine, and Biomedical Engineering. Professor Marras is also the co-director of the Ohio State University Institute for Ergonomics. Dr Marras received his Ph.D. in Bioengineering and Ergonomics from Wayne State University in Detroit, MI. His research is centered around industrial biomechanics issues. Specifically, his research ivolves workplace biomechanical epidemiologic studies, laboratory biomechanic studies, mathematical modeling, and clinical studies of the back and wrist. His findings have been published in over 120 refereed journal articles and numerous books and book chapters. He also holds two patents including one for the lumbar motion monitor (LMM). His work has also attracted national as well as international recognition. Internationally, he has won both the prestigious Swedish Volvo Award for Low Back Pain Research as well as Austria's Vienna Award for Physical Medicine. Nationally, he has been recognized for his research efforts through awards from the Human Factors and Ergonomics Society as well as the Jnstitue for lndustnal Engineers. He has been elected to fellow status in the Human Factors and Ergonomics Society, the Ergonomics Society, and the American Institute for Medical and Biological Engineering.



Kermit C. Davis received his Bachelors of Science in Industrial Engineering from The University of Toledo in 1992, Masters of Science in Industrial and Systems Engineering at The Ohio State University in 1996. He is presently working on his Ph.D. in Industrial Engineering while performing research in the Ohio State Biodynamics Laboratory. His research interests include industrial ergonomics and biomechanics, and modeling with a specialization in the lower back.



Anthony B. Maronitis received his Bachelors of Science in Industrial and Systems Engineering from The Ohio State University in 1997. He is presently working on his Masters of Science in Industrial and Systems Engineering at The Ohio State University while doing research at the Ohio State Biodynamics Laboratory. His research activities focus on low back injury, applied ergonomic and biomechanical studies, and modeling.