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Testing the efficacy of existing force-endurance models to account for the prevalence of obesity in the workforce

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ABSTRACT

This study evaluates whether the existing force-endurance relationship models are predictive of endurance time for overweight and obese individuals, and if not, provide revised models that can be applied for ergonomics practice. Data was collected from 141 participants (49 normal weight, 50 overweight, 42 obese) who each performed isometric endurance tasks of hand grip, shoulder flexion, and trunk extension at four levels of relative workload. Subject-specific fatigue rates and a general model of the force-endurance relationship were determined and compared to two fatigue models from the literature. There was a lack of fit between previous models and the current data for the grip (ICC = 0.8), with a shift toward lower endurance times for the new data. Application of the revised models can facilitate improved workplace design and job evaluation to accommodate the capacities of the current workforce.

Introduction

Over two-thirds (68.6%) of the U.S. population is either obese ($30 \leq$ body mass index (BMI) kg/m^2) or overweight ($25 \leq$ BMI < 30 kg/m^2).[1] The increasing prevalence of obesity and its association with increased risk of work-related musculoskeletal disorders (WMSDs) and injuries have become areas of concern.[2] WMSDs of the upper extremity (mainly shoulder and hand) and trunk are the most common, with 346,170 and 269,290 incidences, respectively, requiring days away from work in 2014.[3] The higher rates of WMSDs reported associated with obesity,[4,5] particularly during prolonged submaximal exertions, might originate from associated physiological and neuromuscular changes. These changes include a higher proportion of fast-twitch, fatigable muscle fibers,[6,7] vascular occlusion,[8,9] reduced neuromuscular activation with fatigue,[10,11] and reduced oxidative enzyme activity,[12,13] all of which suggest altered physical fatigue development with obesity. On the other hand, an adaptive weight bearing training effect with obesity has been postulated, especially for the trunk and lower extremity postural muscles, which may enhance fatigue resistance ability.[14]

The highlighted demographic changes toward more obese individuals, who may be more prone to fatigue, along with industrial automation with higher proportion of static tasks with small range of motions in modern workplaces,[15] necessitates updated practical ergonomic guidelines to prevent fatigue as a risk factor of WMSDs during sustained static tasks.[16] In the literature and in ergonomics practice, localized muscle fatigue (LMF) development has been modeled by the intensity-maximum endurance time (MET) relationship, dating back over 50 years to generalized muscle fatigue models by Rohmert[17] and Monod and Scherrer.[18] Subsequently, researchers have focused on modeling joint specific LMFs on frequently used joints such as the elbow, shoulder, hip, and knee, when the inaccuracy of using a generalized model for all muscle groups was reported due to a posture- and muscle-dependent relationship between endurance and relative loads.[19] The force intensity (fMVC) in this relationship is typically defined relative to maximum isometric contraction (MVC) of a joint or a muscle group.

A lack of generalizability to conditions beyond those tested in addition to either small sample size or very low statistical power has limited the applicability of the empirical models. While the coefficient of determination (R^2) values reported from these models indicate good fits with the data, these may be the result of overfitting due to

the small samples. A well-structured summary of these models is provided by El Ahrache et al.[20] They summarized and categorized the empirical models into general, upper limb, and back/hip models to provide the most commonly used percentiles (i.e., 5th, 10th, and 15th) of the MET for ergonomic applications. A larger meta-analysis of 194 papers on the intensity-static MET relationships was conducted by Frey Law and Avin[21] and identified a significant joint dependency of the relationship leading to the development of best-fit power and exponential curves for each joint. This joint dependency is hypothesized to result from differences in muscle fiber type composition by joint.[22] Despite the importance, any inter-individual variability in muscle fiber type distribution that could affect intensity-MET relationship has been disregarded in these empirical LMF models.

To account for the limitations associated with experimental models, theoretical models were proposed based on the theories of muscle activation and fiber type composition. However, most theoretical models are complex and therefore, not occupationally relevant. Moreover, their validation lacks reliable experimental power or large sample size. Liu et al.[23] proposed a three-compartment model, assuming constant fatigue and recovery rates, and a constant maximum brain effort rate, which could only be valid during the short maximum exertions. Compatible with the assumptions, the model was only validated under the maximum relative load (100% MVC) and with only 10 subjects for the hand grip. Later on, Xia and Frey Law[24] added a time varying function, rather than a constant brain intensity, as a submaximal exertion controller, with joint-specific optimal fatigue and recovery rates included in a subsequent revision.[25]

For dynamic tasks, Ma et al.[26] proposed an alternative model that begins to account for some individual differences. The model was validated based on a set of experimental tests and against the static models provided in previous reviews.[20,26–28] Even after adjusting the fatigue rate (k) for each muscle group, models of the back/hip still deviated largely from the theoretical results. The remaining variability was mainly attributed to individual factors in addition to experimental methods.

Knowing the importance of individual factors, Ma et al.[27] designed an experiment to compare the degree of the compatibility of the Ma et al.[26] model to newly collected experimental data and to investigate inter-individual variability of fatigue rate. Results from a drilling task designed to fatigue the shoulder were reasonably fit to the theoretical model for 87% of the subjects. A high positive correlation between moment arm and fatigue rate was found indicating a significant difference between the fatigue rates of the two groups of participants, the 10 with the highest moment arm and the 10 with the lowest. This significant difference was attributed to the different muscle fiber type distribution for each group. They did not find a significant correlation between BMI and fatigue rate; however, their sample was limited to normal weight and a few overweight participants.

To date, authors have added more complexity to the theoretical models to account for the variability observed in the experimental fatigue data. However, these models have only been validated with samples that did not include individuals who are obese, and thus are not representative of the current U.S. workforce demographics of over two-thirds overweight and obese. Based on the differences in fatigability for endurance tasks that has been observed with obesity,[29–32] it is important that the theoretical models are re-evaluated. Therefore, one objective of the current study was to compare fatigue rates based on a sample of individuals with a range of BMIs. To achieve this objective, first, the subject-specific fatigue rates were determined and compared to the experimental results from Ma et al. (Ma model).[27] Second, the collected data was also tested against the empirical models from Frey Law and Avin (FLA model)[21] to assess the predictive ability of the models. Finally, a revised model of the force-MET relationship for hand grip, should flexion, and trunk extension was determined from a large sample representative of the proportions of normal weight, overweight, and obese individuals in the general population. Since ergonomics in practice is dependent on designing to accommodate the largest segment of the population feasible, and specific models by personal characteristics is impractical, this study focused on developing models based on being representative of the current demographic trends rather than specific models based on obesity level.

Methods

The data used for this article originated from a study conducted by Mehta and Cavuoto[32] on the effects of obesity on endurance time. In that study, a higher fatigability or shorter endurance was found for males versus females during shoulder exertions and for obese compared to normal weight individuals during shoulder and trunk exertions at lower workloads. Based on these findings, this article focuses on the development of endurance models, and a comparison to previously published models, to account for the range of BMI levels and representative proportions of obesity that are present in the U.S. workforce. For a more detailed explanation of the other aspects of the study procedure see Mehta and Cavuoto.[32]

Table 1. Participants' information presented as mean (SD). Those groups that do not share a letter (A, B, C in the order of normal, overweight, and obese) are significantly different at $p < 0.05$ based on a t-test with gender pooled in each group.

	Normal (n = 49)	Overweight (n = 50)	Obese (n = 42)
Age (yr)	31.4(8.5)	33.4(9.8)	31.4(9.4)
Body mass (kg)[A, B, C]	64.8(8.4)	76.7(8.1)	100.0(15.5)
Stature (cm)	169.3(8.4)	168.2(8.4)	169.7(7.5)
BMI (kg/m^2)[A, B, C]	22.6(2.1)	27.1(1.4)	34.6(3.6)
Body fat (%)[A, B, C]	21.7(8.8)	29.9(7.7)	36.3(7.4)
Fat free mass (kg)[A, A, B]	50.9(10.1)	54.0(10.0)	63.9(13.0)
Waist circumference (cm)[A, B, C]	79.8(10.6)	90.8(7.8)	108.1(12.0)
Hip circumference (cm)[A, B, C]	95.3(12.5)	103.2(7.4)	118.5(13.3)
Waist to hip ratio[A, B, C]	0.84(0.06)	0.88(0.06)	0.91(0.06)

Subjects and ethical approval

Following the approval of the standard study protocol by the Institutional Review Boards of University at Buffalo, Buffalo, NY and Texas A&M University, College Station, TX, a total of 142 healthy adults (141 of which were used due to missing data from one participant) with a mean (standard deviation) age of 32.1 (9.2) were recruited between both sites. The sample consisted of 70 females and 71 males, with 34.7% non-obese, 35.5% overweight, and 29.8% obese. These proportions are consistent with the U.S. adult population breakdown.[1] Detailed demographic and anthropometric information is provided in Table 1. All participants provided signed informed consent to participate and completed demographic, health history, and physical activity questionnaires prior to the experiment. Only healthy individuals who did not perform extensive physical activity, determined by self-report, were included in this experiment. An electronic impedance scale (BC-568 Inner Scan, TANITA Corporation, Tokyo, Japan) was used for measuring weight and body fat percentage (%BF).

Experimental setup and protocol

Isometric endurance times for three tasks (shoulder flexion, trunk extension, and hand grip) were tested at four relative workloads (20, 40, 60, and 80% MVC). One workload for each task was tested during each session. The order of loads over the four sessions and tasks within a session were counterbalanced across subjects. Sessions were separated by at least 48 hr to minimize the effect of any residual fatigue on performance. The relative endurance targets were established based on the maximum of three consecutive MVCs completed during the first session for each task. After measurement of MVC, a sustained isometric endurance task at the assigned workload was performed. The participant was asked to exert at the target level for as long as possible, and the time at which they could no longer exert at the required level was recorded as the endurance time. Real-time visual feedback of the force or torque output was presented on a monitor in front of the participant and included a fixed line indicating the target force or torque for the task.

The shoulder flexion and trunk extension tasks were tested using an isokinetic dynamometer (Cybex Humac NORM, Ronkonkoma, NY). Shoulder flexion of the right arm was tested with participants laying supine on the dynamometer chair with a seat belt around pelvis, arm flexed at 90° and elbow extended (Figure 1a). The dynamometer's axis of rotation and shoulder adaptor height were set with respect to the acromion process and arm length, respectively. For trunk extension, participants stood upright on the dynamometer footplate with slightly flexed (<5°) trunk against the sacral pad (Figure 1b). The
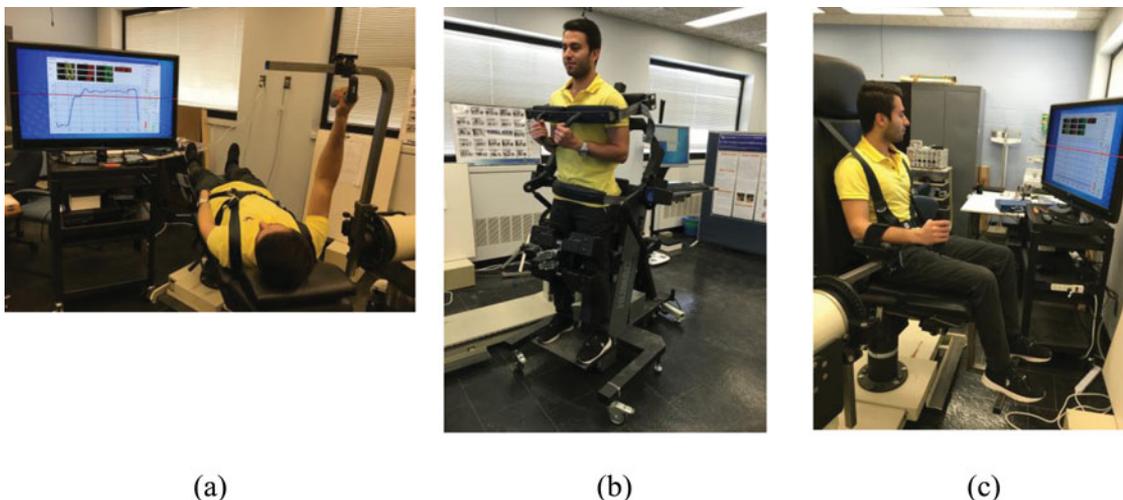

(a)　　　　　(b)　　　　　(c)

Figure 1. Experimental set-up: (a) shoulder flexion, (b) trunk extension, and (c) hand grip.

dynamometer's axis of rotation was aligned based on the iliac crest and L5/S1 location.

Hand grip force was tested with a hand dynamometer (SS25LA, BIOPAC Systems, Inc, Aero Camino Goleta, CA) and data acquisition system (BIOPAC Systems, Inc, Aero Camino Goleta, CA) with the sampling rate of 1 KHz. Participants were seated upright on the dynamometer chair with their arms at their side, resting the lower arm on a stabilizer tube, with the shoulder joint in a neutral position, and the elbow flexed at 90° (Figure 1c).

Statistical analysis

Statistical analyses were conducted in R (Foundation for Statistical Computing, Vienna, Austria) version 3.3.1, and SPSS version 24. Level of statistical significance was set at $\alpha = 0.05$. Relative loads of 20, 40, 60, and 80% were set according to the MVC of each individual in the first session. In order to minimize any potential confounding based on this condition and differences in strength between days, these relative loads were recalculated based on the MVC of the corresponding session. Nonlinear model parameters were estimated using the nonlinear (weighted) least-squares method.

The fatigue rate (k) during isometric sustained endurance test was calculated for each fMVC-MET combination ($N = 548$ for shoulder, 556 for grip, 501 for trunk) using the Ma et al.[26] theoretical model presented in Equation (1). Those individuals with missing data for any of the four points (either MVC or MET) and those with extreme values (i.e., where fMVC exceeded 100% at any time or where the function could not be fit with the experimental data) were excluded from the analysis:

$$MET = -\frac{\ln(f_{MVC})}{k\,f_{MVC}} \quad (1)$$

Fatigue rates were natural log transformed to meet the normality assumption. Homogeneity of variance, and independency of residual errors were verified by Leven's

Table 2. k (min^{-1}) are presented as mean (SD) and [min, max]. The number of participants in each group of Obese (Ob), Overweight (Ov) and Normal weight (N) is shown in the bracket in the last row.

	Ma model Shoulder	New model		
		Shoulder	Trunk	Grip
k (min^{-1})	1.02 (0.49) [0.37, 2.29]	1.76 (0.88) [0.49, 4.96]	0.61 (0.64) [0.08, 4.50]	1.95 (0.86) [0.65, 6.92]
R^2	0.87 (0.14) [0.23, 0.99]	0.90 (0.14) [0.20, 1]	0.82 (0.23) [0.07, 1]	0.81 (0.18) [0.05, 1]
Participants	40	119 [35 Ob, 42 Ov, 42 N]	61 [15 Ob, 25 Ov, 21 N]	122 [35 Ob, 42 Ov, 45 N]

and Durbin-Watson tests, respectively, and by visual inspection of the transferred fatigue rates.

Model comparison and results

In order to compare the fatigue rates to the results from Ma et al.,[27] fMVC-MET was modeled using Equation (1) for each individual with their four data points as the inputs for curve fitting. A comparison of the model parameters and R^2 values with the Ma model are reported in Table 2. As Ma et al.,[27] only provided results for a shoulder task, comparison can only be made for this task. For the shoulder task, a higher fatigue rate with a greater variability was observed in the current sample.

The fMVC-MET relationship for the full sample was also modeled in power and exponential function formats. These new models for grip, shoulder, and trunk were compared to the corresponding FLA models of the grip, shoulder and trunk. The model parameters and comparison measures are presented in Table 3. One-way-random ICC$_1$ was used to compare the models. For the shoulder and trunk-related tasks, a good agreement (ICC$_1 > 0.9$) was observed between the models. RMS and average deviations, differences between FLA model and our models for each task, were also calculated. For the grip task, the FLA model overestimated endurance times. Coefficient of determination was calculated to measure the goodness of

Table 3. Power and exponential model fits and comparisons.

|	FLA model		New model			Model Comparison		
	b_0	b_1	b_0	b_1	Adj-R^2	Average deviations (s)	RMS(s)	ICC$_1$
				$MET = b_0 * exp^{(fMVC*b1)}$				
Shoulder	685.46	−4.97	541.71	−4.27	0.68	0.95	12.86	0.99
Trunk	1165.09	−4.51	1240.05	−3.35	0.33	−98.34	111.59	0.90
Grip	808.15	−4.01	420.37	−3.67	0.71	67.18	89.95	0.80
				Power: $MET = b_0 * (fMVC)^{b1}$				
Shoulder	14.86	−1.83	13.91	−1.87	0.67	1.36	1.65	1
Trunk	22.69	−2.27	70.51	−1.46	0.32	−14.45	174.86	0.93
Grip	33.55	−1.61	18.21	−1.57	0.69	89.97	126.37	0.77
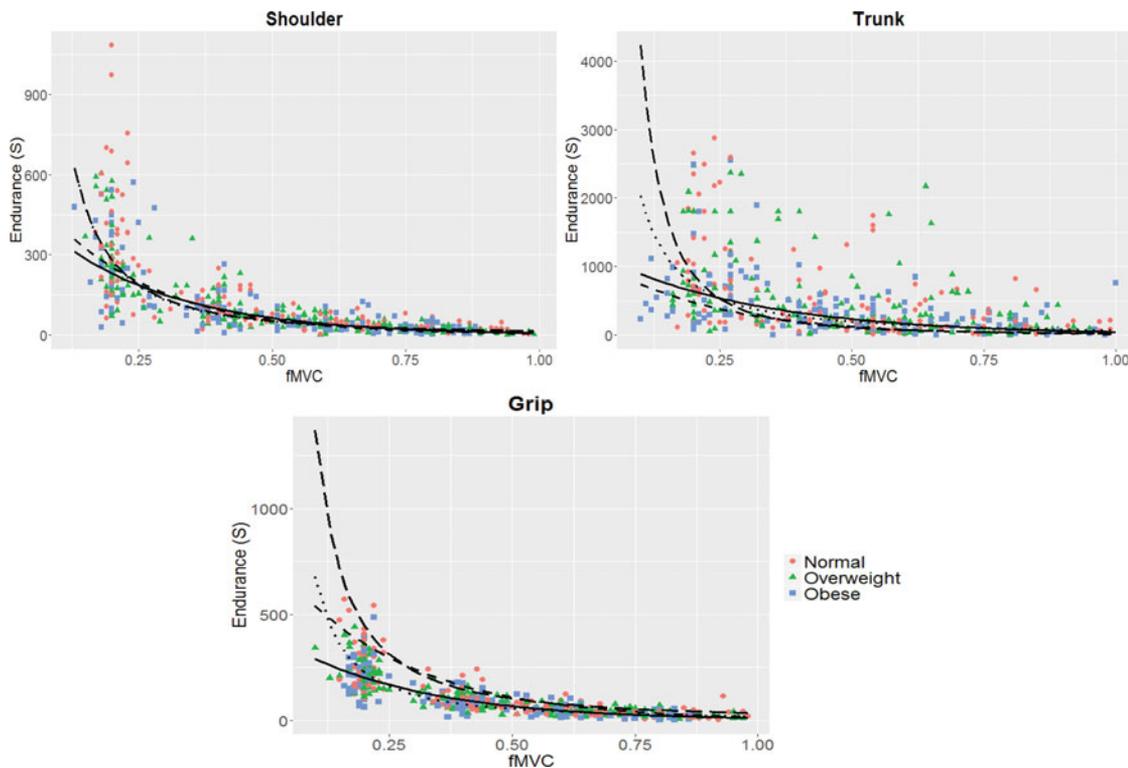
Figure 2. Models of fMVC-Endurance. Solid line: new model, exponential function; Dashed line: FLA model, exponential function; Dotted line: new model, power function; Long-dashed line: FLA model, power function.

fit for the new model. Higher coefficients of determination were observed for the exponential functions compared to the power functions (plotted in Figure 2).

Discussion

Compared to the experimental results of Ma et al.,[27] using the same theoretical model, a 42% faster fatigue rate was observed for the current sample during a shoulder task, indicating that the current theoretical model may not be a good predictor of static endurance time. Even general fatigue prediction models, based on previous experimental models, overestimated grip endurance time when obesity effect was not considered in the analysis. The results of this study support the hypothesis that the current models are not predictive of endurance time with obesity. Revised force-endurance exponential models were provided based on the current proportional sample including obese, overweight, and normal weight subjects, which can allow for more accurate predictions of upper extremity and trunk endurance time.

Previously, there was an attempt to show the effect of personal factors on localized fatigue models. Based on the idea that fatigue rate depends on muscle fiber type distribution, Zhang et al.[33] designed a pushing experiment including 77 workers (38 males and 39 females) to assess the Ma et al.[26] model for gender differences and to compare the results with static experimental model outputs for the shoulder muscle. Results indicated a higher fatigue resistance ability of females compared to male workers corresponding to a greater force decline. This was predicted due to a relatively lower cross-sectional area of fast-twitch muscle fibers for females. The effect of BMI on the fatigue rate was also tested and no significant effect was found, keeping gender as the only significant personal factor in this experiment. However, their sample only included normal weight and overweight subjects.

To our knowledge, this is the first study of localized fatigue modeling considering obesity. In this case, the log-linear relationship better represented the joint-specific nonlinear decay in the new proposed model (higher R^2) compared to the power function. The power function overestimated endurance time especially for fMVC < 0.25 (Figure 2). The observed R^2 values in this study, especially for the trunk extension task, were relatively low due to a large and heterogeneous sample of subjects which includes ranges of BMI from 18–47.5 kg/m^2 with potentially different muscle functional capacity. Previous experimental models with high reported goodness-of-fit had either very small or homogeneous samples. Trunk extension endurance had the greatest variability compared to the shoulder flexion and hand grip tasks. Similarly, a higher variability for back/hip models observed compared to shoulder and elbow models has

been reported previously.[26] This has been attributed to the higher complexity of the musculoskeletal structure and muscle contributions at the back/hip.[26] Also, back/hip was found to be the most fatigue-resistant body part in this study compared to the shoulder and hand grip. A similar pattern was reported previously for empirical models that found the trunk as the joint with the second longest endurance following only the ankle.[21]

Previous models of handgrip failed to accurately predict endurance time in this study. On average, the model overestimated endurance by ∼67 sec with the summary of the previous grip models suggesting a reduced grip endurance in accordance to a shift toward higher obese individuals in the general population. In the literature, a 32% shorter isometric handgrip endurance was reported with obesity.[31] Likewise, a negative effect of BMI on sustained hand grip endurance time was observed in another study.[34] The reduced hand grip endurance in the general population is critical in ergonomic practice where muscle fatigue is estimated based on the existing models to design work-rest schedules in order to prevent or minimize fatigue-related injuries.

While this study presents a comparison to two of the main occupationally relevant localized fatigue models, it is important to acknowledge that other models have been proposed in the literature and are not considered in this analysis.[35,36] These particular models were selected based on their common application. In addition, the findings of this article are limited to endurance during static exertions with a sample of young to middle-aged, healthy individuals who were only recreationally active. Endurance time has been shown to be affected by age, strength, and health status, and the effects of these conditions are not captured in the current sample. Moreover, including individuals with a wider range of BMI increased the variability of endurance times as BMI significantly reduced endurance time during shoulder flexion and trunk exertion tasks at lower intensities.[32] While the models were fit through a range of fMVC values, the design of the experimental tasks and force normalization approach resulted in few data points at the end of the range. Inferences at the low and high ends of relative loading should be approached with caution. Other uncontrolled factors (e.g., the time of day for participation) might have affected the fatigability and caused discrepancies between maximum contractions across sessions. For both the hand grip and shoulder flexion tasks, on average, the coefficient of variation of the MVCs across days was < 10%, whereas for the trunk extension task it was ∼15%.[37] The reliability was consistent across obesity levels. These discrepancies affected the target load and, in some cases, may have led to longer endurance times at low workload

intensities. Future work is needed to explore the effects of other factors and their interactions with obesity level.

Conclusions

This was the first empirical fatigue modeling study to include a large sample with a distribution of normal weight, overweight, and obese individuals in proportion with the general population. In this sample, significantly higher fatigue rates were found for the shoulder flexion and hand grip tasks. For the former task, there was a 42% higher fatigue rate compared to previous experimental results, indicating the importance of considering obesity in fatigue modeling for the shoulder. For the latter task, a lack of fit of the previous models with the current data was observed. In general, there was a shift of the predictive model toward lower endurance times. This paper presents new force-MET models based on the empirical data for estimation of endurance time for ergonomic application.

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