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Endurance time is joint-specific: A modelling and meta-analysis investigation

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Abstract

Static task intensity–endurance time (ET) relationships (e.g. Rohmert's curve) were first reported decades ago. However, a comprehensive meta-analysis to compare experimentally-observed ETs across bodily regions has not been reported. We performed a systematic literature review of ETs for static contractions, developed joint-specific power and exponential models of the intensity–ET relationships, and compared these models between each joint (ankle, trunk, hand/grip, elbow, knee, and shoulder) and the pooled data (generalised curve). 194 publications were found, representing a total of 369 data points. The power model provided the best fit to the experimental data. Significant intensity-dependent ET differences were predicted between each pair of joints. Overall, the ankle was most fatigue-resistant, followed by the trunk, hand/grip, elbow, knee and finally the shoulder was most fatigable. We conclude ET varies systematically between joints, in some cases with large effect sizes. Thus, a single generalised ET model does not adequately represent fatigue across joints.

Statement of Relevance—Rohmert curves have been used in ergonomic analyses of fatigue, as there are limited tools available to accurately predict force decrements. This study provides updated endurance time–intensity curves using a large meta-analysis of fatigue data. Specific models derived for five distinct joint regions should further increase prediction accuracy.

Keywords

holding time; fatigue; isometric; muscle; references; elbow; knee; shoulder; ankle; trunk; grip

1. Introduction

Determining physical capabilities/limitations has long been the focal point of investigations in sport, exercise, rehabilitation and ergonomics. A critical factor in ergonomic assessment is the identification of potential mechanisms/sources of injury that jeopardise workers' quality of life and the ability to optimise work production. Muscle fatigue is one such process that can be implicated as a potential source for injury; involving high load/short duration tasks or low loads/long duration tasks. Muscle fatigue has been defined as 'any exercise-induced reduction in the ability to exert muscle force or power, regardless of whether or not the task can be

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sustained,' (Bigland-Ritchie and Woods 1984); the 'failure to maintain the required or expected force,' (Edwards 1981); and the 'failure to continue working at a given exercise intensity' (Booth and Thomason 1991). Typically maximum holding or endurance time (ET) is the primary outcome variable used to quantify muscle fatigue development, particularly as a function of static contraction task intensity. Although fatigue resistance has been well-described at the single muscle and/or fibre level (Burke *et al*. 1973), little attention has been given to whether fatigue varies systematically between synergistic muscle groups about anatomical joint axes.

The intensity–ET relationship has been long recognised to be nonlinear: as intensity increases, often standardised to a maximum voluntary contraction, ET decreases in a curvilinear fashion. Accordingly, relatively low task intensities can be sustained for long durations, but ET rapidly decreases to very short intervals at maximum intensity. This relationship is frequently referred to as Rohmert's curve in honor of Walter Rohmert who mathematically modelled a static fatigue curve in the 1960's (Rohmert 1960). Because of this nonlinear relationship, factors potentially influencing ET may be dependent upon task intensity. Thus, between joint comparisons must be considered across a wide range of contraction levels.

Numerous attempts have been made to reproduce or update the classic Rohmert's curve, including several joint-specific models (Rohmert 1960, Monod and Scherrer 1965, Hagberg 1981, Huijgens 1981, Sato *et al*. 1984, Manenica 1986, Sjogaard 1986, Kahn and Monod 1989, Mathiassen and Ahsberg 1999, Rose *et al*. 2000, Garg *et al*. 2002). These models most often consist of either power functions (log–log relationship) or exponential functions (log-linear relationship) between intensity and ET, respectively. Although widely acknowledged, most ET models were based on relatively small sample sizes (n = 5 to 40). In a review of 24 static contraction ET models developed by 12 separate investigators, the upper limb is predicted to exhibit significantly shorter ETs for a given intensity than the trunk or hip (El ahrache *et al*. 2006). This analysis, however, relied on model variance as a surrogate for population variance, rather than using experimentally obtained fatigue data, thus may not truly represent underlying physiological differences. There has yet to be a clear consensus of which static contraction ET model provides the most accurate predictions of fatigue development.

Static contraction endurance limit times are reported at two joints, or torque directions within one joint, in a handful of studies with varying results, albeit rarely with the intent to specifically assess these differences. ET appears to vary between joints in several studies (Petrofsky *et al*. 1976, Ohashi 1993, Zattara-Hartmann *et al*. 1995, Smolander *et al*. 1998, Urbanski *et al*. 1999, Alizadehkhaiyat *et al*. 2007), yet not in others (Clarkson *et al*. 1980, Nagle *et al*. 1988, Deeb *et al.* 1992). In one study, between-joint differences in ET varied across intensities, with no clear trend (Bonde-Petersen *et al*. 1975). Little can be concluded from these findings as they 1) compare only a small subset of possible joint combinations and contraction intensities, 2) are lacking in total number of studies involving multiple joints, and 3) involve relatively small sample sizes. Thus, it is currently not clear whether ET varies systematically across the major joints and/or between muscle antagonists in humans.

Theoretically, ET may depend on several factors, such as variations in fibre type (Burke *et al*. 1973), motor unit distribution/activation (Bigland-Ritchie and Woods 1984), neural activation (Clark *et al*. 2005), task specificity (Hunter *et al*. 2005b), and/or absolute force/muscle cross-sectional area (Hunter and Enoka 2001). However it is not clear that these factors vary systematically between joints. If fatigue-resistance proves to vary between several primary joints of the body, we can then work to better understand the underlying mechanisms responsible for variations in fatigue development and how to minimise its potential negative sequelae.

Despite the plethora of research on static contraction muscle fatigue, this vast array of data has not been systematically analysed to investigate between-joint differences or validate intensity-ET models. The nonlinear dependence of ET on contraction intensity makes traditional meta-analysis techniques in isolation challenging, as ET (and thus effect size) cannot be directly compared across different intensities. Creating joint-specific intensity–ET models based on the available data provides a unique combination of analytic techniques, thereby allowing statistical comparisons between multiple joints across all possible intensities. Thus, the goals of this study were to 1) calculate empirically-derived intensity–ET models which best fit the currently available data; and 2) use these models to make joint-level comparisons of fatigue-resistance. To achieve these goals, a thorough systematic review of the literature was performed to obtain all relevant sustained static contraction ET data. These findings are relevant to ergonomic applications that would benefit from validated static contraction ET models for each major joint for which sufficient fatigue data are available.

2. Methods

2.1. Systematic review of literature

The authors performed a two-stage systematic literature review of the literature to find all relevant data linking static contraction intensity and mean endurance time. The first stage involved searches of the following databases: PubMed (1948–9/9/2009), the Cumulative Index to Nursing and Allied Health Literature (CINAHL; 1937–9/9/2009), Pedro (1929–9/9/2009), Science Direct (1825–9/9/2009), Highwire (1812–9/9/2009), and The Cochrane Library (1993–9/9/2009), and the *Journal of Physiology* online search engine (1948–9/9/2009). A total of 32 search terms/keyword combinations were used to elicit relevant articles, including: endurance, fatigue, strength and fatigue, muscle strength and fatigue, isometric fatigue, muscle fatigue time isometric, muscle fatigue time, endurance isometric, voluntary activation fatigue, aging isometric endurance, fatigue force production; and combinations of the above with specific regions: ankle, knee, trunk, shoulder, elbow, hip, wrist, hand, and grip. The inclusion/exclusion criteria (see below) were then employed to include only studies providing relevant information. The second search strategy involved examining the bibliographies of the studies meeting the inclusion criteria to find additional relevant fatigue studies. The inclusion/exclusion criteria were then applied to this second cohort of potential publications. Both authors reviewed the studies to ensure agreement on inclusion/exclusion criteria as well as the data extracted. All data were checked twice against the original articles to minimise any possible transcription errors.

2.2. Inclusion and exclusion criteria

The inclusion criteria included the following: studies involving healthy, human subjects with a mean reported age between 18–50 years; isometric tasks performed until volitional failure; relative intensity based on maximum voluntary contraction (%MVC); mean maximal endurance time reported; single-joint involvement (per fatigue task); and published in English. Studies were excluded that used: dynamic or intermittent static contractions; electrically stimulated contractions; simultaneous multi-joint testing, functional tasks; a maximum test time limit; or body/limb weight as the primary resistance (e.g. Sorensen test). Endurance times for patient populations were excluded; however data for healthy controls were included if provided. Athletic training status was not used for inclusion/exclusion criteria, as a full range of normal healthy endurance capabilities were desired. Efforts were made to exclude duplicate data from publications that may have reported on separate findings from the same cohort (e.g. controls used for comparison with different patient populations). However, if sample size, mean (SD) age, and endurance times did not match exactly, and authors did not indicate data have been presented in part in prior publications, all eligible studies remained in the final analysis.

2.3. Data analyses

All relevant data were compiled in an Excel database when available including: study information (author, date), sample size, sex (male, female, or mixed), mean age, mean and standard deviation (SD) endurance time (sec), standardised intensity (relative to maximum), joint tested, joint angle, and torque direction (e.g., flexion or extension), if provided. If studies involved multiple task intensities, torque directions, or joints, all conditions meeting the inclusion criteria were recorded. Studies reporting multiple categories of normal, healthy subjects (e.g., male vs. female or endurance-trained vs. power-trained) were averaged (weighted by sample size) for a given intensity, to better represent the overall mean finding for that study (excluding any impaired or patient populations). When relevant data were reported in figure form only, numerical values were extracted using pixel analysis of the plots (Adobe Photoshop, San Jose, CA). Intensities were recorded as values between 0 (0% MVC) and 1 (100% MVC), where 1 represented maximum voluntary intensity.

Power and exponential functions were fit to the entire data set (generalised model), for each of the specific joints (i.e., ankle, trunk, grip, elbow, knee, and shoulder), and for specific joint torque directions (e.g. ankle plantar- and dorsi-flexion) if three or more studies, with 10 or more intensities were reported. All models were fit using sample size as a weighting factor (SPSS, Chicago, IL). Pilot studies using simulated data with random noise added (using Matlab, Mathworks, Natick, MA) revealed that linear, least squares fitting methods using data transformations, e.g. log (intensity) and log (ET) for power functions and log (intensity) for exponential functions, reproduced the original simulated data better than using nonlinear least-squares curve-fitting techniques (e.g. the lsqnonlin function in Matlab). The Coefficients of Determination (R^2 values) were determined using SPSS and used to help determine whether power or exponential models better represented the synthesised fatigue data overall. Best-fit model parameters calculated using SPSS were confirmed using Matlab. Ninety-five percent confidence intervals (95% CI) of the model mean values were calculated for each of the joint-specific and generalised fatigue models (power and exponential functions) for intensities ranging from 0.01 to 1 (10% to 100% maximum) using the ‘polyfit’ and ‘polyconf’ Matlab functions. Thus, the fatigue models were developed using only the weighted mean endurance time data from each study. The experimental data and the respective models were plotted using Sigmaplot (Systat Software Inc., San Jose, CA).

Significant differences between the joint-specific models were determined by their standardised degree of overlap between model 95% CIs, determined for intensities from 0.01 to 1. Standardised overlap was calculated as the absolute overlap between CIs, divided by the average ‘error bar’ length of the two models (Cumming 2009). Following Cumming's convention, positive values indicate overlap between model CIs, negative values indicate separation between CI's (no overlap), and a value of zero indicates CIs just touching. Using 95% CIs, and assuming mean study sample size was 10 or more across studies, significant differences ($p < 0.05$) occur when standardised overlaps are ≤0.59 (partial to no overlap) (Cumming 2009).

Between-joint comparisons of the experimental data were performed using pooled means and standard deviations to calculate the 95% confidence intervals (95% CI) of the between-joint differences and the corresponding effect sizes. These comparisons were considered only at intensities with ET data available for 2 or more joints and with a minimum pooled sample size of 10 subjects per joint. Thus, both mean and variance data were used to assess between-joint differences in addition to the model predictions which rely only on mean data, based on the recommendations of El Ahrache *et al.* (2006). Mean endurance times were calculated (in Excel) as the sum of each reported ET at a given intensity level (for each joint) multiplied by the study sample size, divided by the sum of all study sample sizes at that intensity level (Equation (1)). Pooled standard deviations (SD, Equation (2)) for each intensity (by joint), were calculated as the square root of the sum of sample sizes minus 1 multiplied by the square of the standard deviations, divided by the sum of the sample sizes minus one. Thus, Equation (2) provides a pooled SD, weighted by sample size.

$$\text{mean} \quad \text{ET} = \sum (\text{N} * \text{ET}) / \sum \text{N} \tag{1}$$

$$\text{mean} \quad \text{SD} = \sqrt{\left(\sum (\text{N} - 1) * \text{SD}^2\right) / \sum (\text{N} - 1))} \tag{2}$$

where: N = sample size

Statistical comparisons between two joints were made by determining the pairwise mean differences, pooled standard errors (SE, Equation (3)) and critical t-values (based on sample size) to calculate the 95% CI for the ET differences (Equation (4)) (Portney and Watkins 2000). The pooled SE involved taking the square root of the sum of two, squared pooled standard deviations (for joints A and B), each divided by their pooled sample size (Equation (3)). A 95% CI that does not include zero indicates a significant mean difference between pairs at the p = 0.05 level.

$$\text{SE} = \sqrt{\left(\left(\text{SD}_A^2 / \text{N}_A\right) + \left(\text{SD}_B^2 / \text{N}_B\right)\right)} \tag{3}$$

$$95\% \quad \text{CI} = \text{mean difference} \pm t_{crit} * \text{SE} \tag{4}$$

The effect sizes (Cohen's d) were calculated using the mean differences in ET and pooled SD (Equation (5)).

$$\text{d} = \text{mean difference} / \text{mean} \quad \text{SD} \tag{5}$$

Median and range of effect sizes are reported, with large effect sizes being operationally defined as ≥0.8 (e.g. mean between-joint differences are more than 80% of their pooled SD) (Cohen 1992). Similarly, for those studies reporting ET at more than one torque direction at a joint (e.g. flexion/extension), within-study effect sizes and 95% CIs were determined when possible. Significance was set as alpha = 0.05 for all analyses.

3. Results

3.1. Literature review

The first database search strategy resulted in a total of 17,011 potential publications. Search refinement to include humans and English language only decreased the total number of articles to 12,691. Of these 12,691 articles, 167 met the remaining required inclusion and exclusion criteria. The second strategy searching through cited references yielded an additional 27 publications that met the inclusion and exclusion criteria for a total of 194 studies were included in this meta-analysis. The final numbers of studies and data points, by joint, meeting the inclusion criteria are provided in Table 1. Although not technically a single joint, hand and grip studies involving the first dorsal interosseus (FDI), abductor pollicis brevis (APB), adductor pollicis (ADP), and transverse volar type grip are collectively referred to as ‘hand/grip’ for simplicity. Additionally, all studies involving trunk rotation, flexion, side bending, and extension were compressed and termed 'trunk' for simplicity and due to the number of studies fitting the inclusion criteria. The total sample sizes for each joint ranged from 32 to 875 (Table 1), and mean sample sizes ranged from 10.4 to 22.8 subjects per study.

3.2. Static contraction endurance time models

Empirical fatigue decay models (with 95% CIs) for the entire data set (Figure 1) using both power and exponential functions were calculated using all 369 data points, weighted by sample size. The model coefficients and their respective R^2 values for the general model and each of the six joint models are provided in Table 2. Although both exponential and power models were able to predict a large proportion of the variance in experimental data across all models ($R^2 > 0.67$ and 0.75 for the exponential and power functions, respectively), the power function explained a slightly greater portion of the fatigue data variance in all of the 7 models. Figure 2 (A –F) shows the pooled experimental data and the corresponding power models with their 95% CIs for each joint. Owing to their overall superior fit, only the power models were used for all subsequent joint comparisons.

3.3. Joint comparisons

All of the 15 pairwise joint model comparisons were significant (standardised overlap <0.59 between 95% CIs for joint-specific models) over a region of the intensity range, but the size of the regions were intensity dependent (Table 3, below diagonal). Thirteen were significant for more than 51% of the possible 1–100% MVC range (bold text, Table 3), with the magnitude of the differences varying between joint pairs (see Figures 3 and 4). Although fatigue differences varied with intensity, the ankle was most fatigue-resistant, followed by the trunk, elbow, knee, and finally the shoulder was the most fatigable (Figure 3). The hand/grip model demonstrated a slightly different curvature than the remaining joints (Figure 3), such that it approximated knee, elbow, and trunk models at different intensity levels. As the average of the entire data set, the mean generalised fatigue model fell in the middle, nearest the elbow joint model.

The median effect sizes (Cohen's d) pooled across all studies reporting variance data mirrored the model predictions (Table 3, above diagonal). Large effect sizes (>0.8) were observed across 11 joint pairs, in particular for comparisons with the ankle (the most fatigue-resistant) and the shoulder (least fatigue-resistant). Six representative examples of the model and pooled experimental data means and 95% CIs for each joint pair-wise comparison (n = 15 combinations total) are shown in Figure 4 for brevity. Significant differences are indicated. The general fatigue model was relatively indistinguishable from the elbow model, but varied substantially from the other joint-specific models (Table 3, Figure 3).

Only one joint had sufficient data to compare between torque directions at a single joint. Ankle dorsiflexion and plantarflexion models were not significantly different throughout the intensity range (Figure 5). No other within-joint comparisons were performed due to lack of data available.

4. Discussion

This is the first study systematically to compile investigations of static contractions with accompanying ET data to determine static contraction ET decay models as a function of intensity level; and compare them across joints and torque directions. The primary findings of this investigation are: 1) the power function (log-log relationship) was slightly superior to the exponential function (log – linear relationship) at modelling ET data across all joints; and 2) ET varies significantly between joints (e.g. ankle, trunk, elbow, grip, knee, and shoulder) as a function of contraction intensity as indicated by both the joint-specific models and the statistical between-joint comparisons of the pooled experimental data.

Our results demonstrate that both power and exponential models represent the nonlinear decay of ET for a static contraction relationship with reasonable coefficients of determination (Table 2), but the power function better fit the data across all joints. At the moderate to high intensity ranges both power and exponential curves largely overlap. The similar R^2 values between models, despite the clear disparity at the lowest intensities, are likely a result of the preponderance of data at intensities greater than 25% MVC. This may partially explain why more studies have utilised the power model (Rohmert 1960,Monod and Scherrer 1965,Huijgens 1981,Sato *et al*. 1984,Rohmert *et al*. 1986,Sjogaard 1986) than the exponential model (Manenica 1986,Matthijsse *et al*. 1987,Rose *et al*. 2000). Clearly both functions predict curvilinear relationships between intensity and ET, but the exponential model may under-predict ET at the very low task intensities (see Figure 1).

Previous static contraction decay models have represented both general (no joint specific influences) (Monod and Scherrer 1965, Huijgens 1981, Sjogaard 1986) and joint specific intensity–ET relationships (Rohmert 1960, Mathiassen and Ahsberg 1999, Garg *et al*. 2002). In a review of 24 previously published static task ET models, three regional classes were considered: general fatigue models, upper limb (shoulder, elbow, hand) models, and trunk/hip models (El ahrache *et al*. 2006). The models within a body region were widely heterogeneous, but this may be a result of inter-individual endurance capabilities as each model was based upon relatively small sample sizes (5 to 40). Despite this between-model variability, El ahrache concluded significant differences in fatigue-resistance exist between the trunk/hip and the shoulder/upper extremity regions, consistent with our power ET models.

Similarly, our findings based on pooling data across heterogeneous studies are generally consistent with the handful of studies (i.e. 14 of 194) which tested isometric fatigue at two joints within the same cohort. Eight studies observed large and/or significant differences in ET in line with our model predictions. The shoulder was more fatigable than the trunk (d = 1.8 to 2.0) (Yassierli *et al*. 2007) or grip (d = 1.6) (Alizadehkhaiyat *et al*. 2007). The knee fatigued faster than grip in four of six studies with large effect sizes (Cohen's d): 1.1–1.8 (Smolander *et al*. 1998); 1.7 (Petrofsky and Laymon 2002); 2.1 (Zattara-Hartmann *et al*. 1995); and 2.8 (Urbanski *et al*. 1999). Although effect sizes could not be determined, the ankle fatigued less quickly than the elbow across intensities, with differences ranging 63–100% of reported ETs (Ohashi 1993). Finally the trunk was generally more fatigue resistant than the elbow extensors (d = 1.6–3.0) but not the elbow flexors (d = 0.0–1.3) (Bonde-Petersen *et al*. 1975). However, this study was based on an extremely small sample (N = 3). Only two studies reported ET between-joint differences opposite to our model predictions; the reverse direction was observed with grip fatiguing faster than knee, although it was not significantly different, d = 0.2 (Nagle *et al.* 1988) and 0.9 (Williams 1991). In three studies, no significant difference between joints was observed: knee vs. ankle (Clarkson *et al*. 1980), knee vs. elbow (Deeb *et al*. 1992), and knee vs. grip (Nagle *et al*. 1988). None of the 194 studies included in this analysis investigated fatigue at more than two joints in one cohort, thus no data exists to fully validate our multiple predicted between-joint differences.

Although the models utilise only mean data, the experimental data comparisons using both mean and variance information (see Figure 4) were consistent with the model predictions. This is likely due in part to the relatively large number of studies available reporting ET for a given static contraction. Overall, the within-study two-joint comparisons and the pooled mean and standard deviations support the between-joint differences predicted by the power models based on the full complement of data. Thus, using a large systematically-reviewed dataset allowed for greater joint-level fatigue model fidelity than previously assessed, resulting in six distinct joint region models and accordingly fifteen between-joint model comparisons. We are able to conclude that on average ET varies significantly between joints (e.g. ankle, trunk, elbow, grip, knee, and shoulder) as a function of contraction intensity. Although intensity dependent to some degree, the shoulder is the most rapidly fatigable, followed by the knee, grip and elbow, trunk and the ankle is the most fatigue-resistant (see Figure 1).

It is interesting to note that while the shoulder and knee appear to be more fatigable than the trunk, low back injuries are the most common site of injury in the workplace. This discrepancy may be a result of risk factors other than fatigue (e.g. forceful exertions or awkward postures), may be a result of more work-related tasks affecting the trunk than the shoulder or knee, or may suggest that fatigue is not as critical a risk factor as generally believed. Additional research is needed to better clarify the role fatigue has on musculoskeletal injury.

Muscle composition can vary between muscles; the soleus muscle has a greater distribution of type I fibres (80%) than the gastrocnemius (57% type I) (Gollnick *et al*. 1974). Accordingly, we expected ankle plantarflexion to be more fatigue-resistant than dorsiflexion. However, little to no difference was predicted by the power fatigue curves for ankle plantar- and dorsiflexion. This discrepancy may indicate that the moderate difference in fiber type distribution (e.g. 57% vs. 80%) is less critical than other potential factors, such as activation strategy, pressor response, mechanical advantage, etc., that possibly lead to between-joint differences.

Alternately, the unexpected finding may be a result of between-study heterogeneity. In three studies testing both torque directions, plantarflexion ETs were approximately twice that of dorsiflexion when the knee was flexed (Melbech and Johansen 1973, Ciubotariu *et al*. 2004), while no difference was observed when the knee was fully extended (Shahidi and Mathieu 1995). This suggests that when the gastrocnemius is on slack (flexed knee) the soleus muscle is the primary contributor and thus its muscle properties dominate, but when both muscles are allowed to contribute more equally (extended knee), these differences disappear. Thus, the magnitude of within-joint differences may be smaller than the differences observed between joints, suggesting the underlying mechanisms contributing to between-joint differences are likely more complex than simply muscle composition. Although it is beyond the scope of this meta-analysis, we hypothesize that several factors may partially contribute to between-joint differences, such as differences in muscle mass and intramuscular pressure, muscle or fascicle length, activation strategies and descending motor drive, and/or muscle temperature. For example, larger muscle mass can result in reduced fatigue-resistance during sustained isometric contractions; as suggested by male versus female fatigue investigations at the elbow (Hunter and Enoka 2001). Greater vascular occlusion can occur in larger muscles despite similar relative contraction intensities (Hicks *et al*. 2001). Reduced muscle perfusion may impair energy metabolism and alter local muscle pH (Russ *et al*. 2002, Lanza *et al*. 2006). However this mechanism is not fully understood based on conflicting findings in the literature. For example, reduced endurance was associated with higher handgrip peak force, but not reduced forearm blood flow (Thompson *et al*. 2007). Similarly, quadriceps endurance was strongly correlated to the rate of lactate accumulation, but not to the actual muscle pH (Mannion *et al*. 1995). The between-joint differences predicted by our power fatigue models are only partially consistent with the expected influence of muscle mass. The small ankle dorsiflexors demonstrated on average greater fatigue-resistance than did the larger knee extensors. In contrast, however, the relatively small rotator cuff muscles of the shoulder fatigued more rapidly than the larger knee extensors. Additional mechanisms that may contribute to joint-specific fatigue could include systematic differences in muscle and/or fascicle length (Mademli and Arampatzis 2008), activation strategies which can adapt with training (Hunter and Enoka 2003), firing rate which can differ between muscles (Seki and Narusawa 1996), central versus peripheral fatigue (Bigland-Ritchie *et al*. 1986), muscle temperature variations (Petrofsky and Laymon 2005) and possibly even task-specificity (Hunter *et al*. 2002, Maluf *et al*. 2005). It is not clear what the role of postural support is on fatigue-resistance, as the trapezius muscle at the shoulder and the knee extensors, both involved in postural stability, appear to be readily fatigable. Future studies are needed to better identify the salient factors underlying between-joint fatigue resistance, which may even be unique to each joint pair comparison.

A meta-analytic model to synthesise information provides a method to interpret a large body of literature, however this approach has several limitations and interpretation must be performed with caution. While a comprehensive literature review was performed, it is likely relevant publications were missed. Of those included, various methodologies were reported, including differences in lab environment, investigator feedback/motivation, torque measurement, joint angles tested; and operational definition of fatigue/failure. As the fatigue tasks were inherently dependent upon initial maximum torque measurements, any compromise in maximum effort would subsequently result in underestimates of task intensity. Further, we chose to collapse data from different categories within a single study to a single observation at each target intensity to better investigate the overall joint effect on fatigue. All of these factors likely results in greater heterogeneity or ‘noise’. However, the goals of this study were to investigate the general ET versus intensity relationships across joints in healthy adults, thus using 194 studies, with a total of 369 data points, these heterogeneities may well average out. Lastly, the fatigue studies available for this meta-analysis were largely based on relatively small sample sizes.

Future studies are warranted to better characterise model differences, such as males versus females, young versus old, and those familiar (trained) vs. unfamiliar (untrained) with a particular task. Although these characterisations were beyond the scope of this work, they may have influenced the final models, as the distribution between each potential population category was not necessarily balanced (with the exception of no older adult populations included). For example, of the 126 fatigue data points for the elbow, 62 involved only men, 2 involved solely women, and 62 were mixed, including both men and women. Thus, the resulting fatigue curves are likely to be influenced to a greater extent by men than women. In addition, future efforts may benefit from better characterising individual variations in fatigue and its role in injury. For example, individual heterogeneity may partially explain why some may develop musculoskeletal disorders while others don't. Clearly, improved, joint-specific fatigue models may be beneficial for ergonomic analyses, but may also motivate further research to improve our ability to represent the individual rather than a population.

In summary, we found a large body of literature indicating fatigue is dependent on both contraction intensity and joint and the ET–intensity curve is best fit by a power function. We conclude a single generalised fatigue model does not adequately represent most individual joints. Several between-joint effect sizes were quite large, particularly at low contraction intensities. The ankle was the most fatigue-resistant, followed by the trunk. The elbow and grip exhibited similar mid-range fatigue-resistance. The knee and the shoulder were the most fatigable. These findings provide improved models of ET as a function of contraction intensity, advancing our understanding of joint-specific fatigue development. Notably, the most endurant joints (e.g. trunk) do not necessarily have a lower incidence of injury, suggesting future research is warranted to better clarify this relationship.

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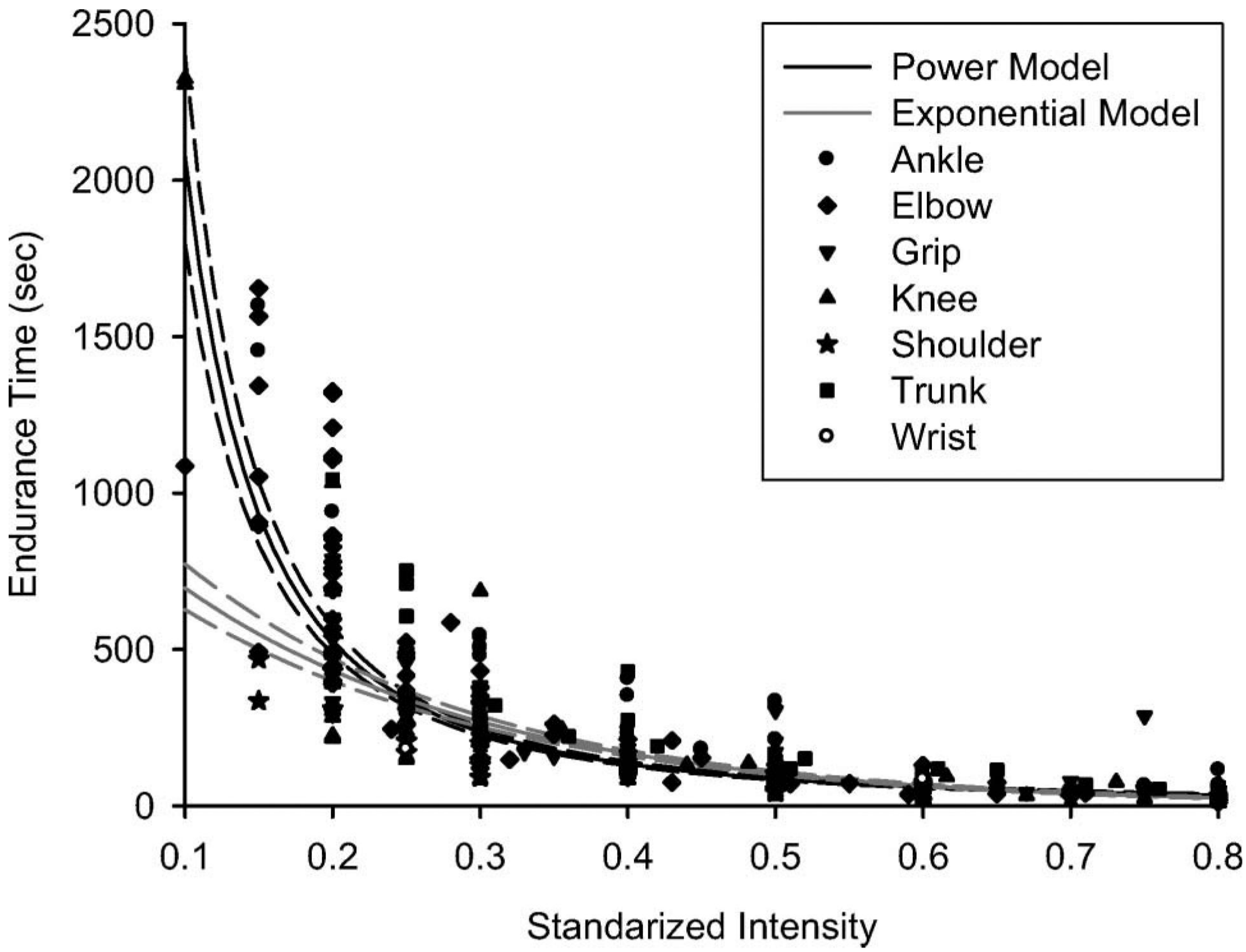

Figure 1.
The general power ($R^2 = 0.81$) and exponential ($R^2 = 0.78$) ET models are shown with their 95% confidence intervals (CIs) overlaid on the full data set (N = 194 studies, 369 task intensities).

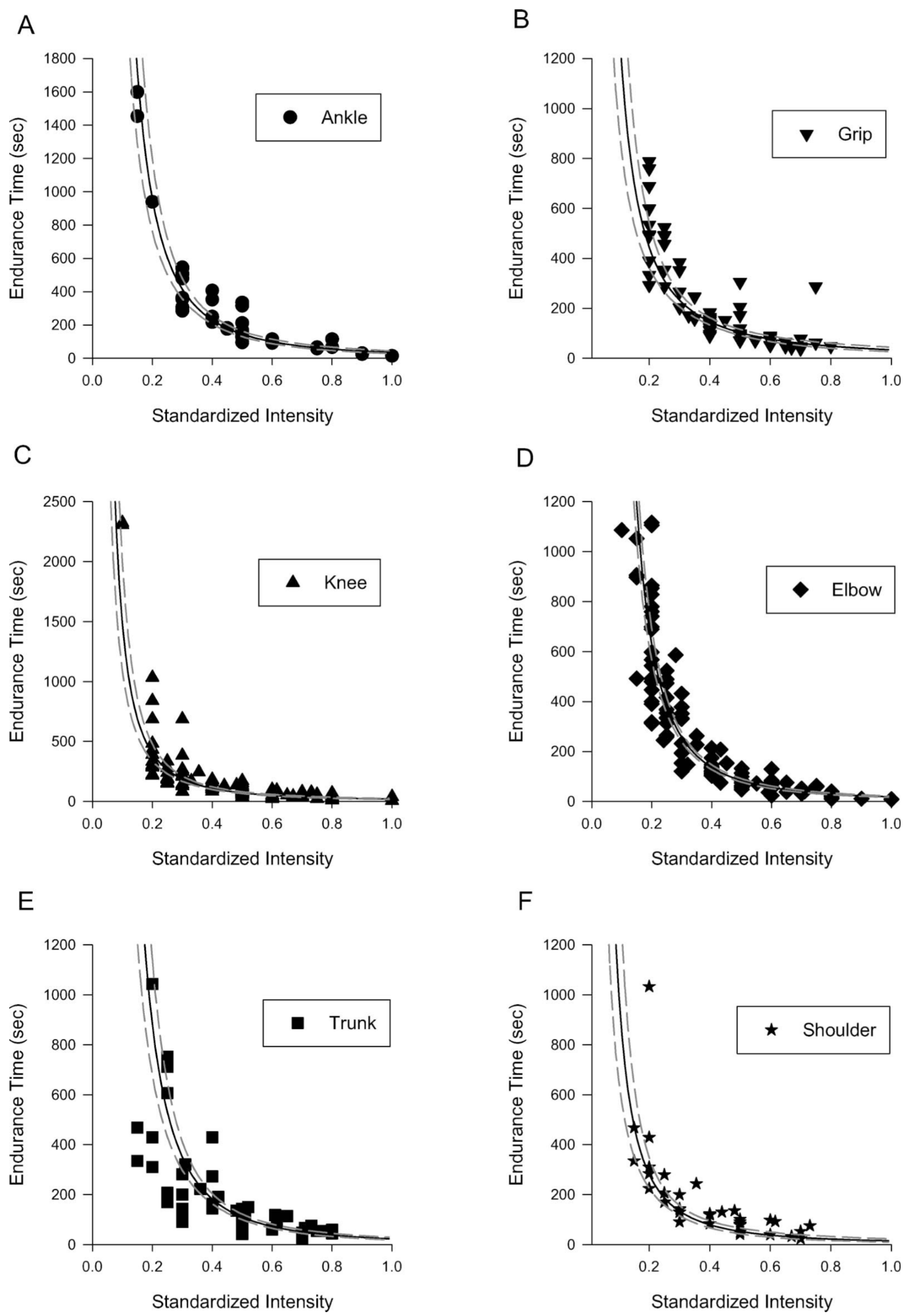

Figure 2.
The joint-specific power models, 95% CIs, and their corresponding experimental data points are shown for the A) Ankle; B) Grip/Hand; C) Knee; D) Elbow; E) Trunk; and F) Shoulder. Each symbol represents the mean endurance time reported for each task intensity. Note the variations in y-axis scaling across panels.

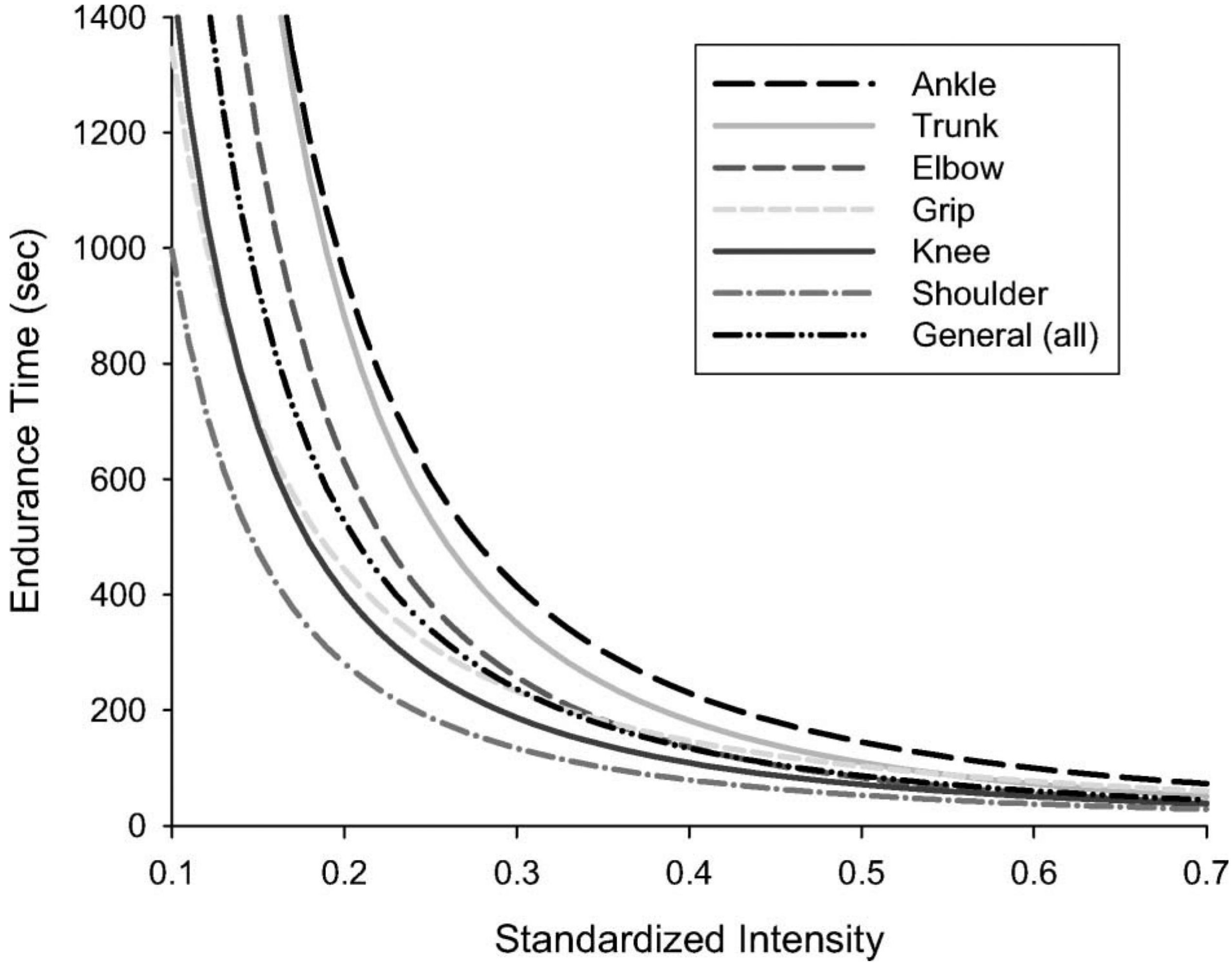

Figure 3.

Joint-specific power fatigue models are plotted to demonstrate relative differences in fatigue resistance (endurance time, ET) as a function of contraction intensity: ankle (solid, dashed); trunk (solid, grey); grip (short-dash, grey); elbow (long-dash, black); knee (solid, black); shoulder (dash-dot, grey). The general model is also shown (dash-dot-dot, black). Note, greater fatigue-resistance is predicted by longer ETs at a given intensity (e.g. ankle versus shoulder).

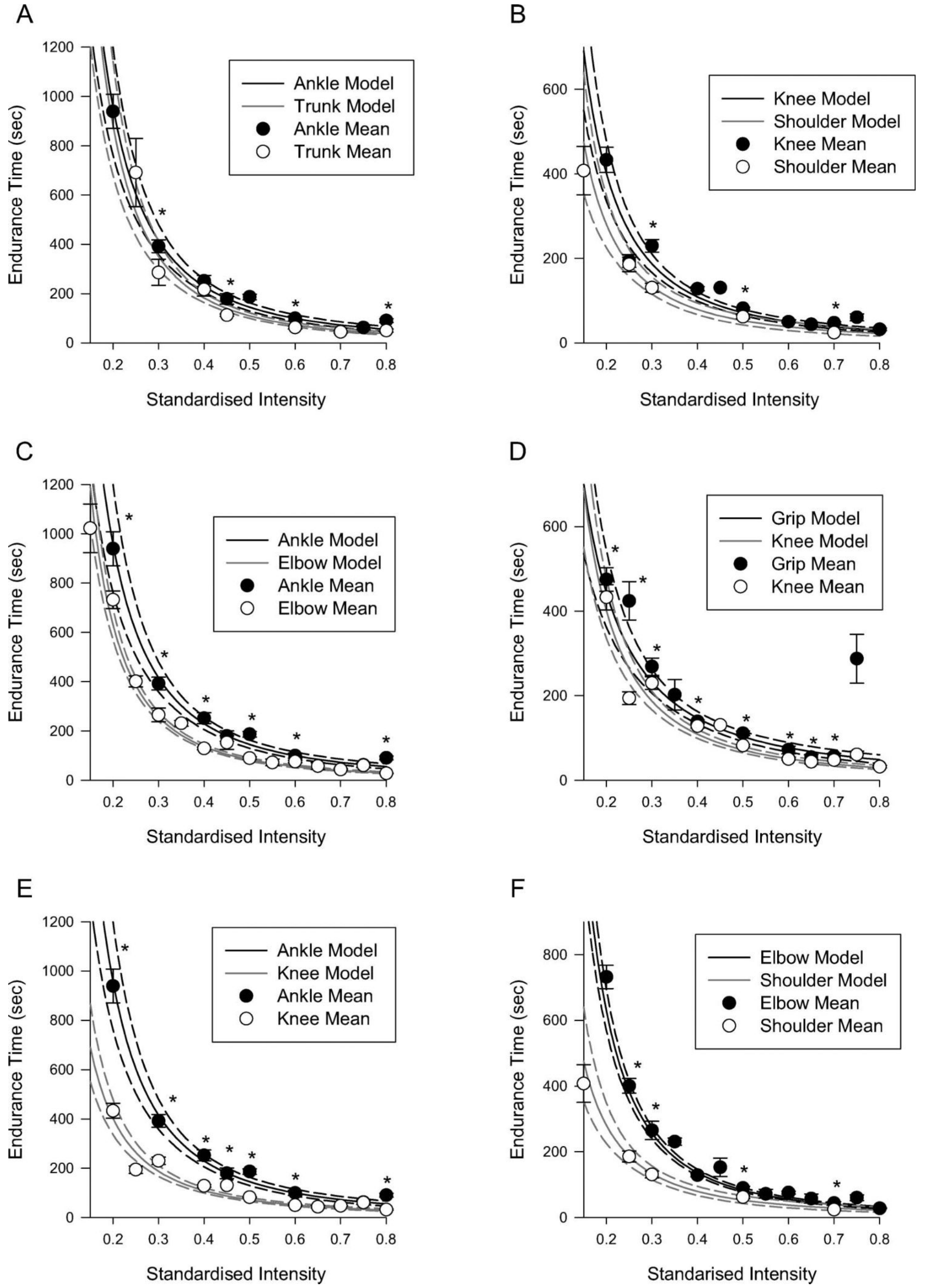

Figure 4.

The mean (95% CIs) power fatigue models for six of the 15 joint pairs demonstrating similar endurance time (ET) predictions: A) Ankle vs. Trunk; B) Knee vs. Shoulder; moderate ET differences: C) Ankle vs. Elbow; D) Grip vs. Knee; and large ET differences: E) Ankle vs. Knee; and F) Elbow vs Shoulder. For each pair, the more fatigue-resistant joint is shown with black circles, the more fatigable joint with open circles. Pooled weighted means (95% CIs) based on reported SD are shown for each joint. Note the varying scales used for ET. *$p < 0.05$ for the experimental data consistent with model.

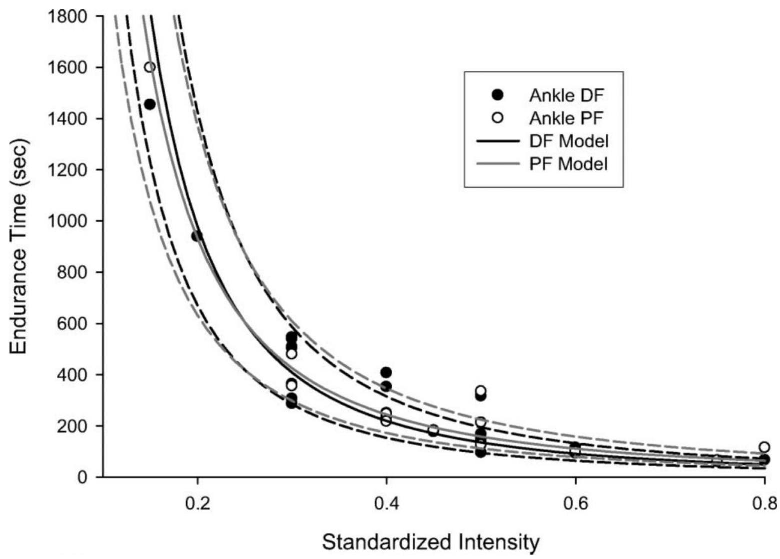

Figure 5.

Within joint comparisons of power fatigue models and the corresponding mean experimental data points are shown for ankle dorsiflexion and plantarflexion. No significant differences were observed between model predictions for ankle torque.

Table 1

Studies included in meta-analyses by joint for ankle, trunk, wrist, hand/grip, shoulder, elbow and knee, listed alphabetically by author.

Joint	Author, Date	N	Intensity (% MVC)	Torque Direction	
Ankle				DF	PF
	(Alway 1991)	24	30	X	
	(Beck *et al.* 2005)	7	30, 40	X	
	(Christie and Kamen 2009)	8	50	X	
	(Ciubotariu *et al.* 2004)	10	50, 80	X	X
	(Clarkson *et al.* 1980)	16	50		X
	(Farina *et al.* 2005)	11	40	X	
	(Houtman *et al.* 2001)	7	30, 60	X	
	(Houtman *et al.* 2002)	8	30	X	
	(Houtman *et al.* 2003)	5	30, 40	X	
	(Hunter *et al.* 2008)	15	20	X	
	(Kent-Braun 1999)	9	100	X	
	(Levenez *et al.* 2005)	12	50	X	
	(Mademli and Arampatzis 2008)	12	40	X	
	(Maton *et al.* 2006)	15	50	X	
	(Matthijsse *et al.* 1987)	8	60		X
	(Melbech and Johansen 1973)	6	50	X	X
	(Ng *et al.* 2000)	11	30	X	
	(Nordez *et al.* 2007)	8	40		X
	(Ohashi 1993)	6	30, 40, 50		X
	(Shahidi and Mathieu 1995)	9	15, 30, 45, 60, 75, 90	X	X
	Totals: 20 studies, 40 data points	N = 207	Range: 15–90		
Elbow				Flex	Ext
	(Bazzucchi *et al.* 2005)	6	30, 50, 80	X	
	(Baudry *et al.* 2009)	20	15	X	
	(Bonde-Petersen *et al.* 1975)	3	25, 30, 35, 40, 50, 60, 70	X	X
	(Calder *et al.* 2008)	10	25	X	
	(Carlson 1969)	47	50, 60, 70, 80	X	
	(Cogiamanian *et al.* 2007)	15	35	X	

Joint	Author, Date	N	Intensity (% MVC)	Torque Direction	
	(Deeb *et al*. 1992)	10	40, 60, 80, 100	X	
	(Dimitrova *et al*. 2009)	6	20, 40, 60, 80, 100	X	
	(Esposito *et al*. 1998)	7	80	X	
	(Fallentin and Jorgensen 1992)	7	10, 40	X	X
	(Felici *et al*. 2001)	6	80	X	
	(Gamet and Maton 1989)	5	25	X	
	(Greiwe *et al*. 1998)	7	50	X	
	(Griffin *et al*. 2001a)	7	20		X
	(Griffin *et al*. 2001b)	7	20		X
	(Hagberg 1981)	9	15, 20, 25, 30, 40, 50	X	
	(Hansen 1967)	19	60	X	
	(Hendrix *et al*. 2009b)	10	30, 45, 60, 75	X	
	(Hermiston and Bonde-Petersen 1975)	3	25, 50, 60, 70	X	
	(Hoeger-Bement *et al*. 2009a)	20	25	X	
	(Hoeger-Bement *et al*. 2009b)	37	25	X	
	(Holtermann *et al*. 2008)	33	25	X	
	(Hunter and Enoka 2001)	14	20	X	
	(Hunter *et al*. 2002)	16	15	X	
	(Hunter and Enoka 2003)	14	20	X	
	(Hunter *et al*. 2003)	24	20	X	
	(Hunter *et al*. 2004a)	31	20	X	
	(Hunter *et al*. 2004b)	27	20	X	
	(Hunter *et al*. 2005a)	8	20	X	
	(Kahn *et al*. 1992)	9	25, 40, 50, 65	X	
	(Kilbom *et al*. 1983)	18	25	X	
	(Klass *et al*. 2008)	11	20	X	
	(Kristev and Kossev 2001)	12	75	X	
	(Krogh-Lund and Jorgensen 1992)	11	15	X	
	(Krogh-Lund 1993)	11	10, 40	X	
	(Krogh-Lund and Jorgensen 1993)	10	30	X	
	(Lloyd 1971)	10	30, 50, 70	X	
	(Lloyd 1972)	30	50	X	

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Joint	Author, Date	N	Intensity (% MVC)	Torque Direction	
	(Lowery *et al.* 2002)	6	30, 50, 80	X	
	(Lowery and O'Malley 2003)	6	80	X	
	(Mamaghani *et al.* 200l)	10	20, 40, 60	X	
	(Mottram *et al.* 2006)	29	15	X	
	(Nicolas *et al.* 2008)	16	40	X	
	(Ohashi 1993)	7	15, 20, 30, 40, 50	X	
	(Orizio *et al.* 1989)	8	20, 40, 60, 80	X	
	(Orizio *et al.* 1992)	13	20, 40, 60, 80	X	
	(Petrofsky and Phillips 1980)	20	25, 40, 55, 70, 90	X	
	(Riley *et al.* 2008)	15	20	X	
	(Rudroff *et al.* 2005)	8	20	X	
	(Rudroff *et al.* 2007a)	20	20	X	
	(Rudroff *et al.* 2007b)	20	20	X	
	(Rudroff *et al.* 2008)	10	20	X	
	(Sacco *et al.* 1999)	10	20	X	
	(Schulte *et al.* 2004)	15	40	X	
	(Semmler *et al.* 1999)	16	15	X	
	(Staudenmann *et al.* 2009)	10	20	X	
	(Watanabe *et al.* 1995)	6	30	X	
	(Yoon *et al.* 2007)	18	20, 80	X	
	(Yoon *et al.* 2008)	15	20, 80	X	
	(Yue *et al.* 1997)	10	20, 65	X	
	Totals: 60 studies, 126 data points	N = 838	Range: 10–100		
Hand/Grip				Grip	Digit
	(Alizadehkhaiyat *et al.* 2007)	6	50	X	
	(Blackwell *et al.* 1999)	18	65	X	
	(Bystrom and Sjogaard 1991)	7	25, 40	X	
	(Chatterjee and Chowdhuri 1991)	40	40	X	
	(Clark *et al.* 2008)	19	20	X	
	(Duchateau *et al.* 2002)	10	25, 50		X
	(Ferguson and Brown 1997)	10	40	X	
	(Fu *et al.* 2002)	5	40	X	

Joint	Author, Date	N	Intensity (% MVC)	Torque Direction	
	(Fuglevand *et al.* 1993)	11	20, 35, 65		X
	(Fuglevand *et al.* 1995)	11	35		X
	(Griffin and Anderson 2008)	10	20		X
	(Huang *et al.* 2007)	14	75		X
	(Hunter *et al.* 2006)	34	20	X	
	(Lind *et al.* 1978)	4	25, 40	X	
	(Lind and Petrofsky 1979)	5	25, 40, 55, 70	X	
	(Longhurst *et al.* 1980)	60	40	X	
	(Louhevaara *et al.* 2000)	21	50	X	
	(Lydakis *et al.* 2008)	15	40	X	
	(Maluf *et al.* 2005)	10	20, 60		X
	(Momen *et al.* 2003)	9	40	X	
	(Momen *et al.* 2004)	9	40	X	
	(Momen *et al.* 2006)	20	40	X	
	(Nagle *et al.* 1988)	10	30	X	
	(Pepin *et al.* 1996)	25	30	X	
	(Petrofsky and Lind 1975)	73	40	X	
	(Petrofsky *et al.* 1975)	61	40	X	
	(Petrofsky *et al.* 1976)	5	40	X	
	(Petrofsky and Laymon 2002)	37	40	X	
	(Petrofsky and Lee 2006)	12	40	X	
	(Saito *et al.* 2009)	9	33	X	
	(Seals 1993)	12	20, 40, 60	X	
	(Smolander *et al.* 1998)	10	20, 40, 60	X	
	(Tharion 2006)	8	50		X
	(Thompson *et al.* 2007)	38	20, 50	X	
	(Torres *et al.* 1983)	18	50	X	
	(Urbanski *et al.* 1999)	10	67	X	
	(Walamies and Turjanmaa 1993)	40	50	X	
	(West *et al.* 1995)	14	30, 50, 70	X	
	(Williams 1991)	6	70	X	
	(Zattara-Hartmann *et al.* 1995)	6	80		X

Joint	Author, Date	N	Intensity (% MVC)	Torque Direction	
	(Zijdewind *et al.* 1995)	12	50		X
	Totals = 40 studies, 58 data points	N = 754	Range: 20–80		
Knee				Flex	Ext
	(Agre and Rodriquez 1990)	41	40		X
	(Bigard *et al.* 2001)	11	25, 70		X
	(Christensen and Fuglsang-Frederiksen 1988)	16	20		X
	(Clark *et al.* 2005)	22	25		X
	(Clarkson *et al.* 1980)	16	50		X
	(Cox and Cafarelli 1999)	10	30		X
	(Crenshaw *et al.* 1997)	11	25, 70		X
	(Deeb *et al.* 1992)	10	40, 60, 80, 100		X
	(Dias da Silva and Goncalves 2006)	9	20, 30, 40, 50		X
	(Easton *et al.* 2007)	10	30		X
	(Ebenbichler *et al.* 1998a)	18	30, 50, 70		X
	(Ebenbichler *et al.* 1998b)	18	30, 50, 70		X
	(Gerdle and Karlsson 1994)	14	10, 25, 70		X
	(Gerdle *et al.* 1997)	20	25, 70		X
	(Grabiner *et al.* 1991)	9	30, 60		X
	(Grabljevec *et al.* 2005)	15	40		X
	(Greiwe *et al.* 1998)	7	50		X
	(Hakkinen and Komi 1986)	21	60		X
	(Hakkinen and Myllyla 1990)	24	60		X
	(Henriksson *et al.* 1986)	13	65		X
	(Hendrix *et al.* 2009a)	9	30, 45, 60, 75		X
	(Hulten *et al.* 1975)	19	50		X
	(Johnson 1982)	15	100		X
	(Karlsson and Ollander 1972)	3	10, 25, 50, 75		X
	(Karlsson *et al.* 1975)	3	30, 50, 80		X
	(Kuroda *et al.* 1970)	6	25, 50, 75, 100		X
	(Larsson and Karlsson 1978)	28	50		X
	(Mademli *et al.* 2008)	12	25		X

Joint	Author, Date	N	Intensity (% MVC)	Torque Direction		
	(Maisetti *et al*. 2002a)	14	50			X
	(Maisetti *et al*. 2002b)	9	50			X
	(Masuda *et al*. 1999)	19	50			X
	(Mathur *et al*. 2005)	22	20, 80			X
	(Maughan *et al*. 1986)	50	20, 50, 80			X
	(Maughan 1988)	8	60			X
	(Mika *et al*. 2007)	10	50			X
	(Nagle *et al*. 1988)	10	30			X
	(Ng *et al*. 1994)	17	30, 50			X
	(Petrofsky and Laymon 2002)	37	40			X
	(Place *et al*. 2005)	11	20			X
	(Place *et al*. 2006)	13	40			X
	(Place *et al*. 2007)	13	40			X
	(Ray and Mark 1993)	8	30			X
	(Ray *et al*. 1998)	10	30			X
	(Rochette *et al*. 2003)	10	20			X
	(Rodriquez and Agre 1991)	41	40			X
	(Rodriquez *et al*. 1993)	7	20, 40, 80			X
	(Smolander *et al*. 1998)	10	20, 40, 60			X
	(Ullrich *et al*. 2007)	12	25			X
	(Ullrich and Bruggemann 2008)	20	20, 40			X
	(Ulmer *et al*. 1989)	30	50			X
	(Urbanski *et al*. 1999)	10	67			X
	(Vangelakoudi *et al*. 2007)	16	44			X
	(Williams 1991)	6	70			X
	(Yamada *et al*. 2002)	14	20, 60			X
	(Zattara-Hartmann *et al*. 1995)	6	80			X
	(Zech *et al*. 2008)	32	50			X
	Totals = 56 studies, 93 data points	N = 875	Range: 10–100			
Shoulder				Flex	Elev	Abd
	(Alizadehkhaiyat *et al*. 2007)	6	50			X
	(Bertmaring *et al*. 2008)	10	15	X		

Joint	Author, Date	N	Intensity (% MVC)	Torque Direction			
	(Brox *et al.* 1997)	19	25				X
	(Gerdle *et al.* 1993)	12	50		X		
	(Hermans and Spaepen 1997)	10	20		X		
	(Kleine *et al.* 2000)	11	50			X	
	(Madeleine and Farina 2008)	12	20			X	
	(Roe *et al.* 2006)	26	25				X
	(Sokk *et al.* 2007)	10	30		X		
	(Takala and Viikari-Juntura 1991)	10	30			X	
	(Troiano *et al.* 2008)	14	50			X	
	(Yassierli *et al.* 2007)	24	30, 50, 70				X
	(Yassierli and Nussbaum 2008)	12	15, 30				X
	Totals = 13 studies, 17 data points	N = 176	Range: 15–70				
Trunk				Flex	Ext	Lat Bend	Rot
	(Bonde-Petersen *et al.* 1975)	3	30, 35, 40, 50, 60, 70, 75		X		
	(Brauer *et al.* 1986)	4	20, 25, 40, 65		X		
	(Clark *et al.* 2003)	20	50		X		
	(Jorgensen and Nicolaisen 1986)	76	60		X		
	(Kankaanpaa *et al.* 1998)	15	50		X		
	(Kondraske *et al.* 1987)	13	50	X	X		
	(Kumar *et al.* 2006)	12	40		X		
	(Ng *et al.* 2002)	12	80				X
	(Ng *et al.* 2003)	23	80				X
	(Nicolaisen and Jorgensen 1985)	32	60		X		
	(O'Brien and Potvin 1997)	22	40				X
	(Potvin and O'Brien 1998)	11	52			X	
	(Sparto *et al.* 2000)	16	70		X		
	(van Dieen *et al.* 1993)	9	25, 40		X		
	(van Dieen and Heijblom 1996)	10	50, 80		X		
	(van Dieen *et al.* 1998)	5	25, 50, 75		X		
	(Yassierli *et al.* 2007)	24	30, 50, 70		X		
	Totals = 17 studies, 33 data points	N = 307	Range: 20–80				
Wrist					Flex		Ext

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Joint	Author, Date	N	Intensity (% MVC)	Torque Direction	
	(Bendahan *et al.* 1996)	6	60	X	
	(Roe *et al.* 2006)	26	25		X
	Totals = 2 studies, 2 data points	N = 32	Range: 25–60		

DF = dorsiflexion; PF = plantarflexion; Flex = flexion; Ext = extension; Grip = transverse volar grip; Digit includes flexion, adduction, and abduction of the hand digits: 1st Dorsal Interosseus (FDI); Adductor Pollicis Brevis (APB); Abductor Pollicis (ADP).

Note: The sum of all joint studies is greater than those meeting inclusion criteria (N = 194) due to 15 studies reporting multiple joints.

Table 2

Power ($Time = bo*(MVC)^{b1}$) and exponential ($Time = bo*exp^{(MVC*b1)}$) model coefficients by joint, where intensity (% MVC) values are between 0.0 and 1.0; time is in seconds.

Model	b_0	b_1	R^2
Power: $ET = bo*(MVC)^{b1}$			
General	21.92	–1.98	0.814
Ankle	34.71	–2.06	0.884
Trunk	22.69	–2.27	0.885
Elbow	17.98	–2.21	0.915
Grip	33.55	–1.61	0.748
Knee	19.38	–1.88	0.789
Shoulder	14.86	–1.83	0.897
Exponential: $ET = bo*exp^{(MVC*b1)}$			
General	1122.32	–4.76	0.784
Ankle	1674.44	–4.51	0.881
Trunk	1165.09	–4.51	0.819
Elbow	1744.7	–5.48	0.892
Grip	808.15	–4.01	0.671
Knee	761.01	–4.38	0.772
Shoulder	685.46	–4.97	0.877

ET = Endurance time (sec).

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Table 3

Model significant differences by intensity (% maximum) below diagonal; Median (range) effect sizes (Cohen's d) above diagonal.

	General	Ankle	Trunk	Hand/Grip	Elbow	Knee	Shoulder
Ankle	**1–100%**		**1.2 (0.4 to 2.3)**	**1.5 (–3.5 to 2.8)**	**1.0 (0.1 to 7.1)**	**2.7 (0.1 to 4.1)**	**2.9 (2.8 to 4.2)**
Trunk	**1–68%**	**34–100%**		0.2 (–0.7 to 1.6)	0.2 (–1.4 to 1.8)	0.6 (–0.6 to 4.0)	**2.1 (1.6 to 3.4)**
Grip	**42–100%**	**1–65%**	1–44%		0.2 (–0.9 to 3.5)	0.7 (0.2 to 3.5)	**2.2 (1.5 to 3.0)**
Elbow	1–28%	**12–100%**	**12–93%**	**44–100%**		0.5 (–0.4 to 1.6)	**1.4 (1.0 to 2.4)**
Knee	**10–69%**	**1–100%**	**1–81%**	**28–100%**	**1–52%**		**1.4 (0.1 to 2.4)**
Shoulder	**3–93%**	**1–100%**	**1–94%**	**16–100%**	**1–72%**	15–62%	

Note: Above diagonal bold text indicates large median effect sizes (≥0.8); below diagonal bold text indicates significant pair wise differences across a majority of the intensity range (≥51%).