Journal of Biomechanics 44 (2011) 1582–1587

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

The utility of an empirically derived co-activation ratio for muscle force prediction through optimization

Rebecca L. Brookham, Erin E. Middlebrook, Tej-jaskirat Grewal, Clark R. Dickerson*

Department of Kinesiology, University of Waterloo, 200 University Avenue West, Waterloo, Canada N2L 3G1

ARTICLE INFO

Article history:
Accepted 21 February 2011

Keywords:
Elbow
Co-activation
Electromyography
Constraint
Optimization

ABSTRACT

Biomechanical optimization models that apply efficiency-based objective functions often underestimate or negate antagonist co-activation. Co-activation assists movement control, joint stabilization and limb stiffness and should be carefully incorporated into models. The purposes of this study were to mathematically describe co-activation relationships between elbow flexors and extensors during isometric exertions at varying intensity levels and postures, and secondly, to apply these co-activation relationships as constraints in an optimization muscle force prediction model of the elbow and assess changes in predictions made while including these constraints. Sixteen individuals performed 72 isometric exertions while holding a load in their right hand. Surface EMG was recorded from elbow flexors and extensors. A co-activation index provided a relative measure of flexor contribution to total activation about the elbow. Parsimonious models of co-activation during flexion and extension exertions were developed and added as constraints to a muscle force prediction model to enforce co-activation. Three different PCSA data sets were used. Elbow co-activation was sensitive to changes in posture and load. During flexion exertions the elbow flexors were activated about 75% MVC (this amount varied according to elbow angle, shoulder flexion and abduction angles, and load). During extension exertions the elbow flexors were activated about 11% MVC (this amount varied according to elbow angle, shoulder flexion angle and load). The larger PCSA values appeared to be more representative of the subject pool. Inclusion of these co-activation constraints improved the model predictions, bringing them closer to the empirically measured activation levels.

1. Introduction

Although inverse dynamic link segment models are often used in biomechanics to predict biophysical magnitudes that are difficult to measure (including moments, joint and muscle forces), difficulties remain in assessing specific tissue loads. It is well known that multiple muscles respond to external loads, which complicates the mathematical allocation of demands amongst these muscles. Specific muscle and joint forces are thus indeterminate since the number of unknown variables (muscle forces and moments) exceeds the number of available mechanical equilibrium equations for most human joints. This indeterminacy problem is often resolved using mathematical optimization (various approaches are described in Herzog, 1996).

Mathematical optimization in biomechanics often involves confounding but necessary simplifying assumptions, including that the body activates specific muscles according to some criterion (e.g. minimum muscle stress or minimum physiological cost) (Crowninshield and Brand, 1981). For such objective functions, antagonistic contraction is counterproductive, as these contractions produce moments that do not contribute to production of the required net joint moments, but increase physiological cost. Thus, those biomechanical optimization models that apply efficiency-based objective functions often underestimate or negate antagonist co-activation (Collins, 1995; Dickerson, 2008; Hughes and Chaffin, 1988; Zajac and Gordon, 1989). However, co-activation patterns have been measured between antagonist and agonist muscles at the elbow (Doheny et al., 2008; Gottlieb, 1998; Praagman et al., 2010; Solomonow et al., 1986), ankle (Granata et al., 2004), trunk (Lee et al., 2006) and knee (Kellis et al., 2003; Kingma et al., 2004). There is consensus in the literature that co-activation occurs for the purposes of limb or end effector motion control, joint or whole body stabilization and limb stiffness (Granata et al., 2004; Latash, 1992; Milner and Cloutier, 1993; Zhang and Eymer, 1997).

Upper limb control and end effector precision depend on effective elbow stability, which makes co-activation especially important at this joint. The elbow is often modeled as a single degree of freedom hinge joint (An et al., 1984; Chaffin et al., 2006; Hughes and An, 1999) despite its known movements

* Corresponding author. Tel.: +1 519 888 4567x37844; fax: +1 519 746 6776.
E-mail address: cdickers@uwaterloo.ca (C.R. Dickerson).

0021-9290/$ - see front matter © 2011 Elsevier Ltd. All rights reserved.
doi:10.1016/j.jbiomech.2011.02.077

in flexion/extension (0–150°) and pronation/supination (75° pronation to −85° supination) (An and Morrey, 1985). Elbow co-activation patterns are influenced by fatigue, movement velocity, eccentric exercise, accuracy and elbow positioning (Bazzucchi et al., 2006; Corcos et al., 2002; Doheny et al., 2008; Gribble et al., 2003; Missenard et al., 2008; Semmler et al., 2007). Praagman et al. (2010) studied elbow muscle load sharing during both flexion/extension and pronation/supination, and found that joint angle (and therefore, moment arm and muscle length) influence EMG amplitude and load sharing between muscles. However, to our knowledge, no studies have explicitly quantified the mathematical relationship of elbow co-activation for specific arm postures and hand loads. While EMG driven models exist to estimate muscle forces and joint loads for the elbow (Soechting and Flanders, 1997; Buchanan et al., 1998), knee (Lloyd and Besier, 2003) and spine (McGill, 1992; Van Dieen and Kingma, 2005), and comparisons have been made investigating the effects of antagonistic co-activation between EMG- and optimization-based models (Van Dieen and Kingma, 2005), there have been limited attempts to merge an a priori co-activation constraint with optimization models that is then independent of empirical data. Our study aims to explore this void.

Hence, this study had two purposes. The first was mathematical description of the empirical co-activation relationship between elbow flexors and extensors. It was hypothesized that co-activation levels would increase as the flexion joint angle decreases (as the elbow moves towards full extension) due to a similar reported trend for reduced task sets in the elbow (Doheny et al., 2008) and knee (Kingma et al., 2004). The second purpose was to assess the influence of including the co-activation relationships as constraints in an existing planar optimization muscle force prediction elbow model. It was hypothesized that enforcing co-activation would result in a more biofidelic model that yields more realistic muscle force predictions.

2. Methodology

2.1. Part I: Defining the co-activation relationship

Participants performed 72 submaximal isometric elbow exertions. Four bipolar surface electrode pairs recorded activity from right elbow flexors and extensors. Prediction equations for the co-activation index (CI) were developed for elbow flexion and extension, using arm posture, hand force and subject characteristics as predictive factors.

2.1.1. Subjects

Sixteen right-hand dominant participants (8 males and 8 females) who were free from chronic (lasting longer than 6 months) or acute (within the past 6 months) shoulder, elbow and/or wrist injury gave informed consent to participate in this study. Participants averaged 24.1 years old (SD 2.4 years), 1.73 m (SD 0.1 m) in stature and 74.4 kg (SD 20.0 kg) in mass with upper arm and forearm lengths of 0.32 m (SD 0.03 m) and 0.27 m (SD 0.01 m), respectively. Mean participant maximum force generation in flexion and extension at 90° elbow flexion was 180 N (SD 68 N) and 175 N (SD 67 N), respectively. Ethics was received from the institutional Office of Research Ethics.

2.1.2. Surface electrodes

The skin over electrode sites was shaved and cleansed with isopropyl alcohol. Disposable bipolar Ag/AgCl surface electrodes (Product #272, Noraxon, USA Inc., Arizona, USA) were affixed using previously published placements on the right side over the biceps brachii, long and lateral heads of triceps brachii (Delagi and Perotto, 1980), and the brachioradialis (Rudroff et al., 2008). A reference electrode was placed over the sternum, just inferior to the suprasternal notch.

2.1.3. Data acquisition

Electromyography (EMG) was sampled using a wireless Noraxon TeleMyo 4200T G2 (Noraxon USA Inc., Arizona, USA) at 3000 Hz. This system had 16-bit resolution on all analog inputs with a band-pass filter from 10–1500 Hz, an input impedance > 100 MΩ, a common mode rejection ratio > 100 dB and a base gain of 500. Data was transferred from the receiver to a personal computer, and analyzed using MATLAB™, 7.10.0 (The Mathworks Inc., USA).

2.1.4. Pre-experimental protocol

Seated participants performed three ramped isometric maximal voluntary contractions (MVCs) of both elbow flexion and extension. MVCs were 6 s in duration, and 2 min of rest were given between MVCs. The peak value from a 500 ms moving window average between the middle 2 s of the MVC trials was calculated and used for normalization purposes to define the maximal contraction for each respective muscle and subject.

2.1.5. Experimental protocol

Each participant performed 72 submaximal isometric exertions that required either right elbow flexion or extension (each of 6 s duration) while holding a load (weighted bottle) in their right hand. Three weighted (1, 2 and 3 kg) bottles were used. Exertions consisted of combinations of 4 elbow angles (0°=full extension, 45°, 90°, 135°), 2 exertion types (elbow flexion and extension); 3 loads (1, 2, 3 kg); and 3 shoulder angles (0° shoulder flexion and 0° shoulder abduction; 0° shoulder flexion and 45° shoulder abduction; 45° shoulder flexion and 0° shoulder abduction). The body was positioned in sitting, prone, supine or prone with 90° trunk flexion postures to accommodate flexion or extension exertions. The wrist and forearm were neutral during all exertions. Postures are outlined in Appendix A. A hand-held manual goniometer was used to confirm postures. The order of exertions was randomized. The approximate duration of the study (from set-up to finish) was 2.5 h for each participant.

2.1.6. EMG processing

EMG was processed in MATLAB™, 7.10.0 (The Mathworks Inc., USA). Raw EMG was full wave rectified and filtered with a low-pass 2nd order single pass Butterworth filter with a cut-off frequency of 3.5 Hz. A 500 ms moving window average (MWA) of linear enveloped data was calculated for the MVC trials. The highest MWA from the two MVC trials (for each elbow flexors and extensors) was used to normalize the 72 experimental trials to respective muscles and participants. The linear enveloped normalized EMG was integrated between 2 and 4 s for each experimental trial and participant using the cumulative trapezoidal integration approach.

2.1.7. Co-activation ratio

The integrated linear enveloped normalized EMG was used in the calculation of a co-activation index (CI), similar to Kellis et al. (2003) (Eq. (1)). CIs were calculated for all 72 exertions for each of the 16 participants, resulting in a total of 1152 CIs:

$$CI = \frac{\int_{t1}^{t3} [EMG_B + EMG_{BR}](t)\ dt}{\int_{t1}^{t3} [EMG_B + EMG_{BR} + EMG_{TL} + EMG_{TG}](t)\ dt} \times 100 \quad (1)$$

where B is the biceps, BR the brachioradialis, TL the triceps (lateral head) and TG the triceps (long head).

The CIs provide a relative measure of flexor contribution to total activation around the elbow. More explicitly, a CI of 0% indicates absence of co-activation (flexors are not activated); a CI of 100% indicates absence of co-activation (extensors are not activated); and a CI of 50% indicates full co-activation (both flexors and extensors are activated in equal amounts).

2.1.8. Statistical analyses

Statistical analyses were performed in JMP 8® (SAS Institute Inc., Cary, NC). Potential predictor variables (arm posture, load and subject anthropometric factors) were analyzed for correlations and redundant variables (variables found to be highly correlated) were removed. A preliminary stepwise multiple linear regression analysis determined which predictor variables should be included in each of the co-activation prediction models. These models (one for extension and one for flexion exertions) were developed using a repeated measures analysis of variance. The CIs were the dependent variables, and the independent variables included shoulder flexion and abduction angles, elbow angle, load and anthropometric factors. Non-linear and sinusoidal models were also developed, but provided no further variance explanation, so the more parsimonious linear models were used.

2.2. Part II: Implementation of co-activation constraints in a model

Co-activation relationships defined from Part I were added as constraints in a biomechanical optimization model. Model muscle force predictions with and without the constraint were compared with each other and experimental electromyographic data.

2.2.1. Model modifications

A 2D inverse dynamics, optimization based, muscle force prediction model of the elbow was created through modification of a model described by An et al. (1984)