Quantifying synergist activation patterns during maximal plantarflexion using an orthogonal expansion approach

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Abstract

This study sought to determine how well a pattern-recognition approach based on orthogonal-expansion theory can quantify patterns of activation amplitudes that have been recorded by electromyography (EMG) from synergist muscles during isokinetic plantarflexion. Raw surface EMG data were recorded from six muscle sites – four over the agonist Triceps Surae (Lateral and Medial Gastrocnemius, a lateral and a medial site over the Soleus) and two over the antagonist Tibialis Anterior (upper and lower sites) muscles – in ten healthy subjects as they performed three maximal plantarflexor efforts against an isokinetic dynamometer at each of three angular-velocity settings (90 trials). After the root-mean-square amplitude had been calculated for each EMG recording and had been normalized to the amplitude measured during a maximal voluntary isometric contraction, the mean normalized amplitudes were calculated for each muscle site, for the 90 trials. The differences among the muscles and the variability in the normalized amplitudes indicated that the sample mean did not characterize the patterns of relative activation amplitudes among the six muscle sites for all subjects, trials and conditions. An eigenvector decomposition of the normalized data yielded a set of vectors that represents the principal patterns of the activation amplitudes. The principal pattern reflected by eigenvector 1 was 0.57, 0.49, 0.47, 0.45, 0.12 and 0.06 corresponding to the Lateral and Medial Gastrocnemius, the lateral and medial Soleus sites, and the upper and lower sites on the Tibialis Anterior, respectively. The percent trace for eigenvector 1 was 94%; however,
two or three eigenvectors were needed to characterize the patterns for some trials. Since 99\% of the variance was accounted for by three eigenvectors, the approach was effective in reducing the data while maintaining the salient features in the synergistic patterns of activation amplitudes during isokinetic plantarflexion. © 1998 Elsevier Science B.V. All rights reserved.

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1. Introduction

Assessing the level of coactivity (i.e. simultaneous recruitment) of synergistic muscles performing specific tasks using electromyography (EMG) is important in the study of human movement, whether addressing clinical, ergonomic, or sport related problems. Although there are many definitions of muscle synergism (Basmajian and DeLuca, 1985; Sirin and Patla, 1987), in this study, both agonist and antagonist muscles have been included since both affect the net moment of force necessary to produce movement about a joint or provide stabilization to a joint (Falconer and Winter, 1985; Hubley-Kozey et al., 1994; Psek and Cafarelli, 1993). The need to record EMG from more than one agonist, from more than one site on large muscles and from the antagonist muscles is supported by several studies. Activation amplitudes have been shown to differ among individual muscles in agonist groups (Gravel et al., 1987; Hof and van den Berg, 1977; Mayniak et al., 1991; Sirin and Patla, 1987) and differences in the relative activation amplitudes measured from different sites within the same muscle have also been reported (Jongen et al., 1989; Zuylen et al., 1988) for various test conditions. Coactivation amplitudes recorded from antagonist muscles of healthy subjects have been shown to be quite variable for a variety of muscle groups and test conditions (Falconer and Winter, 1985; Hebert et al., 1991; Hof et al., 1987; Osternig et al., 1984; Psek and Cafarelli, 1993; Richardson and Bullock, 1986). These studies demonstrate the need to record from multiple sites to study synergistic muscle activation patterns and provided the impetus to examine pattern-recognition techniques to quantify EMG patterns. To be effective, a pattern-recognition technique must reduce the amount of data needed to characterize a pattern and maintain the salient features from the measured pattern, (i.e. the error between the measured patterns and the reduced patterns is minimal). Pattern-recognition based on orthogonal-
expansion theory has been applied to quantify phasic/temporal EMG patterns during gait (Patla, 1986; Shiavi and Griffin, 1981; Wootten et al., 1990), but not to determine patterns of relative EMG amplitudes from synergistic muscles.

The purpose of this study was to determine how well a pattern-recognition approach based on orthogonal-expansion theory (Gerbrands, 1981) can quantify patterns of activation amplitudes that have been recorded by surface EMG from multiple sites over synergistic muscles during maximal, concentric isokinetic plantarflexion. Plantarflexion was chosen as the study movement for several reasons. First, the majority of the plantarflexor torque is produced by the three components of the Triceps Surae (TS) muscle group with the Tibialis Anterior (TA) muscle responsible for the majority of the antagonist dorsiflexion torque; all are accessible using surface EMG. Secondly, the three TS muscles produce plantarflexor torques, but they have different fibre type proportions (Edgerton et al., 1975), different cross-sectional areas (Woittiez et al., 1983), different functions i.e. include one and two-joint muscles (Gravel et al., 1987), and are innervated by different branches of the Tibial nerve (Kimura, 1989). Thirdly, the level of TA coactivation for normal healthy subjects, during stepping and walking movements was shown to be highly variable (Hof et al., 1987), with significantly higher levels of TA coactivation reported for untrained versus trained subjects during isokinetic plantarflexion (Amiridis and Morlon, 1995). Finally, the Soleus muscle has a large cross-sectional area (Woittiez et al., 1983) and the TA muscle is relatively long (Wickiewicz et al., 1984) providing examples of muscles that may require more than one recording site to accurately reflect the activation amplitude of the entire muscle. It was anticipated that a range of activation patterns would be recorded among subjects, trials and conditions. Subsequently more than one pattern would be needed to account for the variance in the relative EMG amplitudes.

2. Methods

2.1. Subjects

Ten healthy subjects (2 men and 8 women, mean age = 30 ± 7.5 years, height = 166 ± 4.7 cm. and mass = 63 ± 12.8 kg) participated in the study after giving their informed consent in writing, in accordance with the Ethical Guidelines of Dalhousie University’s Faculty of Health Professions. Subjects
had no neuromuscular or musculoskeletal problems that would affect their ability to produce a plantarflexor torque.

2.2. Procedures

2.2.1. Electrode placement

Amplifier specifications and procedures for the EMG data acquisition were in accordance with the ISEK standards document (Winter et al., 1980), as well as more recent publications (Ortengren, 1996; Redfern, 1992, Winter, 1996). The skin above each muscle site was shaved to remove excess hair and rubbed with alcohol to reduce skin impedance (the input impedance of the amplifiers was 100 MΩ). The recommended ratio of the skin–electrode impedance to the input impedance of the amplifier is less than 1% (Winter, 1996), and our ratio was less than 0.05% (upon attachment the skin–electrode impedance was less than 50 KΩ). Since skin–electrode impedance decreases and stabilizes over time (Redfern, 1992), the effect on the signal amplitude was negligible and below the resolution of the measuring system. Meditrace silver/silver chloride pellet surface electrodes (ECE 1801, 10mm, Graphics Control) were placed in a bipolar configuration (collar to collar), in line with each muscle’s fibre orientation, over six sites. Electrodes for the Lateral Gastrocnemius muscle (LGA) were placed consistent with the lead line and position described by Zipp (1982) and the Medial Gastrocnemius muscle (MGA) site was consistent with Ericson et al. (1985). Electrodes for the lateral site on the Soleus muscle (LSO) were placed consistent with Ericson et al. (1985) and the medial site on the Soleus muscle (MSO) was at the same level as the lateral site with care taken for each pair of electrodes to ensure that they were attached to the centre of the areas defined, respectively, by the Achilles tendon (identified by palpation), the inferior border of the LGA, and the lateral border of the Soleus muscle, and by the Achilles tendon, the inferior border of the MGA, and the medial border of the Soleus muscle. Electrodes for the upper site on the Tibialis Anterior (UTA) muscle were placed consistent with the lead line and position defined by Zipp (1982); those for the lower site on the TA (LTA) were placed 2 cm distal to the lower UTA electrode along the lead line (with minor adjustments to accommodate anatomical differences between subjects). Therefore, the LTA was approximately at the site described by Ericson et al. (1985). The ground electrodes were attached to the tibia and the lateral epicondyle of the femur. The electrode sites were validated by means of isolating movements associated with each muscle (Winter, 1996). The electrode surface area and the inter-electrode distance,
were within the recommended areas and distances (Soderberg, 1992). Therefore, the pick up area for the electrode placements used in this study were within the surface area of the muscle over which the electrodes were placed (Winter, 1996; Fuglevand et al., 1992), minimizing the potential for cross talk from adjacent muscles.

2.2.2. Trials

No EMG recordings were taken for at least 10 min after the electrodes were attached to allow the skin–electrode impedance to stabilize. Then, while the subjects lay supine, completely relaxing their muscles, EMG signals were recorded to establish a baseline and noise level for the subject and system. The root-mean-square (RMS) noise for the total system was less than 5 µV for each EMG channel (i.e. approximately one A-to-D unit). The subjects’ right foot was then secured in the Cybex dorsiflexor/plantar flexor footplate, with their knee flexed to 160° (180° being full extension), as determined by a standard goniometer. After the subjects had warmed up by performing 3–5 submaximal plantar- and dorsiflexion movements at the three test velocity settings on the dynamometer (30°/s, 90°/s and 150°/s), they performed two maximal voluntary isometric contractions (MVICs) of both the plantarflexor and the dorsiflexor muscles (ankle angle was 5° plantarflexion, i.e. 0° was neutral). The contractions were maintained for 3 s, and the data were collected for the middle 250 ms of the MVIC. The MVIC trials were performed in random order, with the subjects being given a 2 min rest between contractions. They were repeated if the peak torque measured in the two trials differed by more than 10%.

The test trials consisted of three plantarflexion and three dorsiflexion contractions at each of three angular-velocity settings on the Cybex – 30°/s, 90°/s and 150°/s – with the order of the test trials being randomly assigned. The starting angle was standardized, and the total range of motion for the dynamic plantarflexion contactions was approximately 50° (i.e. 0° dorsiflexion to 50° plantarflexion). The subjects were asked to produce a maximal effort and were given verbal feedback on their performance; they were allowed to rest at least 2 min between trials.

2.3. Data acquisition

Each data channel was calibrated before each testing session. The torque channel from a Cybex II isokinetic dynamometer (Lumex, New York) was calibrated by hanging known masses a measured distance from the axis of ro-
The potentiometer was calibrated by using a carpenter’s level to identify 0° and 90°. The six EMG channels (bandpass 10–500 Hz, CMRR = 90 dB, input impedance 100 MΩ) were calibrated using a 166 Hz sine wave that had a peak-to-peak amplitude of 1 mV. The gain was calculated for each EMG channel, to calculate the voltage at the skin–electrode interface. For each trial, the eight channels of data were digitized at 1000 samples per second, using a Tecmar Labmaster (12-bit resolution (±10 V), Scientific Solutions, Solon, OH) interfaced with an IBM 8086 Personal computer (Boca Raton, FL). The resolution for the EMG channels was between 2.5 and 5 μV at the electrode level. The sampling rate of 1000 Hz was two times the upper cut off frequency of the EMG amplifiers (Bandpass 10–500 Hz), the latter is consistent with the recommended upper cut off for amplifiers used for surface EMG (Ortengren, 1996). Winter (1996) suggested that the sampling rate should be four times the upper frequency of interest. The majority of the frequency content of the surface EMG signal is less than 300 Hz (Basmajian and DeLuca, 1985; Redfern, 1992), therefore the majority of the signal content (i.e., below 250 Hz) was within the recommended rate. The data were stored on floppy disk for off-line processing.

2.4. Data processing

The eight channels of data were processed on an IBM 486 personal computer. FORTRAN programs first corrected for baseline bias and then converted values to millivolts for the six EMG channels. The data were visually windowed from the beginning of movement to the end of movement, using the angular-displacement curve obtained from the Cybex for each trial. The average torque (Nm) and the root-mean-square (RMS) amplitudes for the six EMG sites over the entire window were calculated, using numerical recipe algorithms (Press et al., 1992). The RMS amplitudes, which reflect the power in the EMG signal (Basmajian and DeLuca, 1985) were normalized to the standard maximal isometric contraction (Lawrence and DeLuca, 1983) to quantify the relative activation levels for each muscle site during each contraction, using

\[
\text{RMS}_{N/j} = \frac{\text{RMS}_{TR/ij}}{\text{RMS}_{MVIC,j}} \times 100, \tag{1}
\]

where \(\text{RMS}_{TR/ij}\) is RMS amplitude for muscle \(j\) for trial \(i\), \(\text{RMS}_{MVIC,j}\) is the highest RMS amplitude between the two MVIC trials for muscle \(j\) and...
RMS_{Nji} is the normalized percentage data for muscle j, trial i. RMS_{Nji} were used in the data-reduction analysis.

Further data processing was performed on a Stellar GS1000 computer (Stardent Computer, Concord, MA). The transform to reduce the pattern space of the RMS_{Nji} data for each muscle-contraction trial was applied as follows. The processed EMG data set consisted of m muscle sites and n discrete realizations of the process. The vector x_i, dimensioned m x 1, represents the RMS_{Nji} for trial i. Thus, we have n vectors x_i such that [X] = [x_1, x_2, ..., x_n], where [X] is an m x n data matrix for which m = 6 electrode sites and n = 90 (the number of trials per subject times the number of subjects in the sample).

First, a cross product matrix, [C_c] of the columns of [X] was calculated.

\[
[C_c] = \frac{1}{n-1}[X][X]^t.
\]

The transform of the vectors x_i defined as, y_i = [T]^t x_i, produces a vector of coefficients, y_i, for each trial for each subject (see Gerbrands, 1981). The transform matrix [T] was calculated from an eigenvalue and eigenvector analysis: [C_c] = [T][A][T]^t. FORTRAN programs, including NAG library routines (Numerical Algorithms Group), were used to calculate the eigenvalues and eigenvectors of a real symmetric matrix using Householder’s reduction and the QL algorithm. [A] is the diagonal matrix of eigenvalues, \lambda_i, ordered in descending order of magnitude to minimize the mean-square error. [T] is the transform matrix, dimension m x m, of orthonormal eigenvectors EV_i. The required number k of EV_i's depends on the total information contained in them. An estimate of the error of truncation by k eigenvectors was obtained from the average error e_k, expressed as a percentage by considering the signal voltage associated with the average error,

\[
\% \text{tr}[C_c] = \left( \sum_{i=1}^{k} \lambda_i \right) \times \frac{100}{\text{tr}[C_c]},
\]

where tr[C_c] is the trace of the cross product matrix; eigenvalues, \lambda_i, are in descending order of magnitude. The percent trace was used to determine the number of eigenvectors required for accurate reconstruction of the measured values. The expansion was truncated to k terms, where k < m. The vectors \hat{x_i} were reconstructed using the equation

\[
\hat{x_i} = [T_r] y_{i(r)},
\]

where \hat{x_i} is the reconstructed vector of RMS_{Nji} for trial i, [T_r] is the reduced transform matrix, and y_{i(r)} is the reduced coefficient vector of dimension k for
trial $i$. Thus, $\hat{x}_i$ is the linear combination of $k$ EVs weighted by $k$ coefficients $y_i$. Since expansion theory determines the linear combination of orthonormal basis functions that minimize the mean square error, the magnitude of $y_i$s provides a weighting factor for the contribution of the associated $EV_i$s to the overall synergistic pattern of activation across the six muscle sites. Relative errors (Rel) were calculated to evaluate the accuracy of the reconstruction by dividing the root-mean-square error by root-mean-square signal and multiplying by 100. Worst-case relative errors were examined. The reconstruction was then applied to the RMS values recorded using the same procedures from one test subject who was not part of the original sample to determine whether the $EV_i$s could accurately reconstruct their pattern of activation amplitudes. The same error analysis as above was performed.

2.5. Statistical analysis

To address the purpose of this study, only the plantarflexion trials were analysed. Whether the torque and the RMS EMG amplitude for each muscle differed significantly between the two MVIC trials was tested by means of paired $t$-tests ($\alpha = 0.05$) for dorsiflexion and plantarflexion, separately. Three one-factor repeated-measures analysis of variance (ANOVA) tested whether there were statistically significant differences in the average torque among the three test trials at each angular velocity. A Bonferroni correction ($\alpha = 0.0167$) was used to test which pairwise differences were significant for the significant $F$ ratios from the ANOVA. Means and standard deviations were calculated for each coefficient ($y$), for RMS, and for the relative error of the reconstructions based on $1$–$k$ patterns. Pearson Product Moment correlations were calculated to determine the relationship between the coefficients ($y$) and the $RMS_{Nji}$. Statistical significance was determined for each correlation ($p \leq 0.05$), and the associated $r^2$ for each correlation was calculated. All statistics were calculated using Excel (Ver. 5.0). The reconstructed patterns for the sample and for individual trials were graphed, so that the effects of including from 1 to $k$ $EV_i$s could be assessed qualitatively.

3. Results

The mean torques for the dorsiflexor MVIC trials were 35 (±12.5) Nm and 36 (±13.0) Nm for trial 1 and 2, respectively; those for the plantarflexor
MVIC trials were 103 (±20.5) Nm and 102 (±18.8) Nm. There were no statistically significant \( (p \geq 0.05) \) differences between trials for either dorsiflexion or plantarflexion. When the RMS amplitudes for the six muscle sites were individually compared, there were no significant differences between the two MVIC trials, either for dorsiflexion or for plantarflexion.

The mean torques were not significantly \( (p \geq 0.05) \) different among the three trials for 90°/s (26, 25 and 23 Nm) and 150°/s (18, 17 and 19 Nm), but Trial 2 (40 Nm) was significantly different \( (p \leq 0.05) \) from Trial 1 (50 Nm) – although not from Trial 3 (46 Nm) – at 30°/s. The total angular displacement over which the movement took place was 45° for the 30°/s trials and 49° for the two faster-velocity trials.

Fig. 1 shows the RMS\(_{N_{ij}}\) mean (SD) for the 90 trials for each of the six muscle sites. The mean RMS\(_{N_{ij}}\) recorded from all four sites on the TS muscles were above 100% of the MVIC, with the mean and the variability being highest for the LGA site (127% ± 62). The mean RMS\(_{N_{ij}}\) from the two sites on the TA muscle were less than 30% MVIC.

The pattern recognition results are in Table 1. The cumulative percent trace of matrix \( [C_c] \) was 99% for three EV\(_1\)s; therefore, less than 1% of the pattern information was contained in EV\(_4\)–EV\(_6\). Given this low expected error, the highest number of eigenvectors evaluated to reconstruct the EMG patterns was \( k = 3 \). The percent trace for EV\(_1\) shows that the pattern depicted by EV\(_1\) was the dominant pattern (i.e. LGA having the highest weighting and MSO the lowest for TS, and UTA having twice the weighting of LTA). Since the EVs are not correlated (i.e. orthogonal), the pattern of weightings for each muscle site are different, as the vectors in Table 1 illustrate. The large SD for the coefficients \( y \) (Table 1) shows that the three EV\(_1\)s differed in their contribution to the overall pattern between trials, conditions and subjects.

Fig. 1 graphically illustrates the reconstructed patterns using the sample mean \( y \) for each EV\(_i\) from one to three, compared to the mean measured pattern of activation for the sample. The sample means were reconstructed with minimal error using EV\(_1\). Fig. 2(a) and (b), however, illustrate that EV\(_1\) did not accurately reconstruct the relative patterns of activation for the two trials depicted, and two or three eigenvectors were needed to reconstruct the patterns of activation among the synergists. The two trials were chosen to illustrate the range of patterns that were recorded from healthy subjects under consistent test conditions.

The coefficients \( y_i \) were significantly correlated with some of the relative activation amplitudes measured from the six muscle sites (Table 2). Coefficient \( y_1 \) had a significant correlation \( (p \leq 0.001) \) with the RMS\(_{N_{ij}}\) amplitudes.
from all four TS sites, and the best $r^2$ was 0.79 for the LGA. Although the RMS$_{Nji}$ from the UTA was significantly correlated ($p \leq 0.05$) with $y_1$ and $y_3$ and the RMS$_{Nji}$ from the LTA was significantly correlated ($p \leq 0.05$) with $y_1$, the associated $r^2$ values illustrate that less than 10% of the variance in the data was explained by these relationships.

The correlation between $y_1$ and the sum of the TS ($\sum$ TS) RMS$_{Nji}$ amplitudes ($r = 0.999$) indicates that 99% of the variance was explained by this linear relationship. The deviations in the activation amplitudes of the individual
muscle sites from the principal pattern, however, were characterized by the
ratios of $y_2$ and $y_3$ to $y_1$ ($R_1 = y_2/y_1$, and $R_2 = y_3/y_1$). The high correlation
($r = -0.97$ in Table 3) between $R_1$ and the ratio of the LGA to MSO ampli-
tude (the two sites with the highest weighting in $\text{EV}_2$) indicates that $R_1$
explains 94% of the variance associated with the deviation of LGA to MSO
ratio from the principal pattern. Similarly, the correlation ($r = 0.91$) between
$R_2$ and the ratio of MGA to LSO (the two sites with the highest weighting in $\text{EV}_3$)
indicates that $R_2$ explains 80% of the variance from the principal pat-
ttern for MGA to LSO ratio. Small ratios indicate minimal variance from the
principal pattern, whereas a large ratio indicates that the variance from the
principal pattern was explained by the pattern associated with the numerator
of the ratio.

Two examples of how $y_1$, $R_1$ and $R_2$ quantify two very different patterns
are depicted in Fig. 2(a) and (b). The $\sum TS$ in Fig. 2(a) is only slightly higher
than the sample mean, whereas that in Fig. 2(b) is much higher; the associ-
ated $y_1$ for Fig. 2(a) is similar to the sample mean, and much higher for
the trial in Fig. 2(b). The patterns of relative activation among the sites dif-
fered greatly between the two trials, and they both differed from the sample
mean. The $\text{RMS}_{N_1}$ amplitudes in Fig. 2(a) had higher relative amplitudes re-
corded from the two lateral sites than from the two medial sites. The associ-
ated $R_1$ (0.04) was small, indicating that $\text{EV}_2$ contributed very little to the
overall pattern, consequently the LGA/MSO ratio was similar to the sample
mean. The large negative $R_2$ ratio ($-0.25$) occurred because MGA/LSO ratio

Table 1
Cumulative percent trace (%) for the three eigenvectors with the highest associated eigen values

<table>
<thead>
<tr>
<th>Eigenvectors</th>
<th>$\text{EV}_1$</th>
<th>$\text{EV}_2$</th>
<th>$\text{EV}_3$</th>
</tr>
</thead>
<tbody>
<tr>
<td>%trace</td>
<td>94.4</td>
<td>97.0</td>
<td>98.7</td>
</tr>
<tr>
<td>Sites</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>LGA</td>
<td>0.5655</td>
<td>-0.7796</td>
<td>0.0669</td>
</tr>
<tr>
<td>MGA</td>
<td>0.4893</td>
<td>0.1925</td>
<td>0.5832</td>
</tr>
<tr>
<td>LSO</td>
<td>0.4701</td>
<td>0.1691</td>
<td>-0.7823</td>
</tr>
<tr>
<td>MSO</td>
<td>0.4486</td>
<td>0.5709</td>
<td>0.1439</td>
</tr>
<tr>
<td>UTA</td>
<td>0.1203</td>
<td>0.0606</td>
<td>-0.1458</td>
</tr>
<tr>
<td>LTA</td>
<td>0.0642</td>
<td>0.0342</td>
<td>-0.0381</td>
</tr>
<tr>
<td>$y_1$</td>
<td>233.3</td>
<td>5.3</td>
<td>-0.9</td>
</tr>
<tr>
<td>SD</td>
<td>74.9</td>
<td>40.7</td>
<td>33.2</td>
</tr>
</tbody>
</table>
Fig. 2. Measured RMSN and reconstructed patterns for: (a) a 90°/s trial for subject 8; $y_1$ is 245.6, $y_2$ is 8.6 and $y_3$ is 60.7, $R_1$ is 0.04 and $R_2$ is 0.25; (b) 150°/s trial for subject 2; $y_1$ is 327, $y_2$ is 88 and $y_3$ is 27, $R_1$ is 0.27 and $R_2$ is 0.08. The measured RMSN is in the left column and the reconstructed patterns based on including one, two and three eigenvectors are illustrated in the three columns to the right.
was very different from the principal pattern i.e. in this trial the LSO was much higher than the MGA. There was a large increase in the UTA compared to the LTA, also characteristic of EV3. It is clear that EV3 improved the accuracy of the reconstructed pattern for the trial in Fig. 2(a).

In addition to having higher-than-mean relative amplitude for the \( \sum \)TS sites, the pattern in Fig. 2(b) had higher relative amplitudes recorded from the two medial TS sites than from the two lateral sites. The large positive \( R_1 \) (0.27) associated with EV2 occurred because the LGA/MSO ratio was

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Table 2
Pearson product moment correlations between the y coefficients and the muscle activation amplitudes (\( RMS_N \)) and associated \( r^2 \) values

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Coefficients</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>( y_1 )</td>
</tr>
<tr>
<td></td>
<td>( r )</td>
</tr>
<tr>
<td>LGA</td>
<td>0.89***</td>
</tr>
<tr>
<td>MGA</td>
<td>0.83***</td>
</tr>
<tr>
<td>LSO</td>
<td>0.75***</td>
</tr>
<tr>
<td>MSO</td>
<td>0.70***</td>
</tr>
<tr>
<td>UTA</td>
<td>0.25*</td>
</tr>
<tr>
<td>LTA</td>
<td>0.21*</td>
</tr>
<tr>
<td>( \sum )TS</td>
<td>0.999***</td>
</tr>
<tr>
<td>( \sum )TA</td>
<td>0.09</td>
</tr>
</tbody>
</table>

\* \( p \leq 0.05 \).
\** \( p \leq 0.01 \).
\*** \( p \leq 0.001 \).

---

Table 3
Pearson product moment correlation between the ratios and the muscle activation ratios and the associated \( r^2 \) values

<table>
<thead>
<tr>
<th>Muscle ratios</th>
<th>Ratios for coefficients</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>( R_1 )</td>
</tr>
<tr>
<td></td>
<td>( r )</td>
</tr>
<tr>
<td>LGA/MSO</td>
<td>-0.97***</td>
</tr>
<tr>
<td>MGA/LSO</td>
<td>0.16</td>
</tr>
<tr>
<td>UTA/LTA</td>
<td>0.13</td>
</tr>
</tbody>
</table>

\*** \( p \leq 0.001 \).
much lower than the principal pattern. Adding \( \mathbf{EV}_3 \) did not substantially improve the accuracy of the reconstructed pattern, and the associated \( R^2 \) (0.08) was small. \( \mathbf{EV}_1 \) and \( \mathbf{EV}_2 \) weight the UTA two times the LTA, which was consistent with the measured pattern in Fig. 2(b).

The relative reconstruction errors for the entire sample of trials – 5.0 (±3.1), 3.1 (±2.3) and 1.6% (±1.1), based on including 1, 2 and 3 \( \mathbf{EV}_i \)s, respectively – were consistent with the expected error estimated by the percent trace. The measured and reconstructed amplitudes for the worst- and best-case errors from this sample are shown in Table 4. The worst-case relative error based on including three \( \mathbf{EV}_i \)s (Table 4) illustrates that the main site contributing to this large error was the LTA. The best reconstruction based on three \( \mathbf{EV}_i \)s illustrated that all sites but the LTA were within 1% MVIC. The results for the test subject are also included in Table 4. The average reconstruction error was less than 4% with the largest deviation at the UTA site. The corresponding \( y_i \)s were 179.5, 15.7 and −22.8 for \( y_1 \), \( y_2 \), and \( y_3 \), respectively. \( R_1 \) was 0.09 and \( R_2 \) was −0.13.

4. Discussion

This study sought to determine how well a pattern-recognition approach based on orthogonal-expansion theory quantified patterns of activation amplitudes recorded by surface EMG from synergist muscles during isokinetic plantarflexion. The normalization procedure was standardized and the results confirm that a consistent effort was achieved between the two MVIC trials for dorsiflexion and for plantarflexion. The dynamic contractions were also stan-
standardized among trials, subjects and velocities and the results support a consistent maximal dynamic effort. The focus was then to quantify the pattern of relative levels of activation among the six electrode sites and to determine whether the pattern was consistent among trials, subjects and conditions.

The large variability for the sample means and the patterns depicted in Fig. 2(a) and (b) and Table 4 illustrate that healthy subjects performing a standardized movement produced a wide range of relative activation patterns. The RMSN\textsubscript{ij} amplitudes from the four TS sites during the dynamic contractions show that the EMG amplitudes recorded for some trials were greater than and for some trials were less than 100\% MVIC; this is consistent with previous reports of differences in relative amplitudes of activation recorded from surface EMG between isometric and dynamic contractions (Theeuwen et al., 1994). Theeuwen et al. (1994) concluded that the differences in direction-dependent EMG activity of a muscle between isometric and movement tasks were the result of changes in the recruitment thresholds of motor units between isometric and dynamic contractions. Our results also corroborate differences reported in the relative EMG activity between one and two-joint muscles (Theeuwen et al., 1994), among individual TS muscles performing plantarflexion tasks for various test conditions (Gravel et al., 1987; Hof and van den Berg, 1977; Sirin and Patla, 1987) and among recording sites from within the same muscle under varying conditions (Buchanan et al., 1986; Jongen et al., 1989; Zuylen et al., 1988). These amplitude differences have been related to trade offs, coactivation, task specificity and inhomogeneous motor unit recruitment (Caldwell and van Leemputte, 1991; Gravel et al., 1987; Hebert et al., 1991; Jongen et al., 1989; Sirin and Patla, 1987; Zuylen et al., 1988). As well, the TA co-activation amplitudes in this study were consistent with amplitudes reported for healthy trained and untrained subjects during isokinetic plantarflexion exercises (Amiridis and Morlon, 1995) and the variability among subjects was consistent with that reported for stepping and walking (Hof et al., 1987; Falconer and Winter, 1985). Therefore, in order to describe or test relationships among the six sites, complex statistical analysis are needed if the data cannot be reduced.

Although the variability in the data supports the need to record EMG from the six sites used in this study, the pattern-recognition analysis shows that one principal pattern, (EV\textsubscript{1}), explained 94\% of the variance in the relative levels of activation. Overall three patterns (99\% variance explained), rather than the measures from six recording sites, were needed to characterize the syner-
gistic patterns which included trial to trial differences, differences among velocities and differences among subjects. The high correlation between the \( \sum \) TS activation amplitudes and the \( y_1 \) (Table 2) indicates that one coefficient, \( (y_1) \), rather than four measures, can be used to evaluate the activation amplitude of the TS muscle group. The correlations between \( R_1 \) and the ratio of the LGA/MSO and between \( R_2 \) and the ratio of MGA/LSO (Table 3), illustrate that the two ratios can be used to evaluate the relative activation amplitudes among the four TS muscles rather than comparing the main effects and interactions among the four individual measures. While none of the individual coefficients or ratios were highly correlated with the TA activation amplitudes, combining the three eigenvectors accurately reconstructed the TA patterns for most trials.

The accuracy of the EVs to maintain the salient patterns of synergistic activation for this sample were confirmed through the error analysis since the estimated (%trace) and the mean of the actual errors (relative errors) were small. The worst-case-error (Table 4) occurred because a high percent MVIC was recorded from the LTA site compared to the UTA site (i.e. LTA was 50% MVIC). The high level of activation from the LTA site was inconsistent with the other nine subjects in the original sample, however, the test subject had slightly higher EMG amplitudes recorded from the LTA site compared to the UTA site. Increasing the sample size from which the eigenvectors are calculated is needed to determine whether the accuracy for the TA reconstructions could be improved. The day to day variability for the percent trace and the coefficients also should be addressed before applying the EVs calculated in this study to quantify synergistic patterns during isokinetic plantarflexor testing. The low errors for the test subject were an encouraging result, however, more test subjects are needed to determine how well the patterns characterize the measured patterns from other subjects and test conditions.

In conclusion, healthy subjects used different relative percentages of their MVIC recorded from sites over agonist and antagonist muscles to perform a simple, controlled, maximal-effort, plantarflexion task at three angular velocities. The orthogonal-expansion analysis reduced the data from six electrode sites to three principal patterns representing 99% of the variance in the measured EMG patterns. Therefore the principal patterns characterized the salient features in the synergistic pattern of activation. Each pattern weighted the individual muscle amplitudes differently, and the associated coefficients provided a weighting factor for each pattern. The total amplitude of activation for the four TS muscle sites was highly correlated to one coefficient and the relative differences among TS sites were highly
correlated with two ratios. The results illustrate that the approach has merit for reducing patterns of activation amplitudes measured via surface EMG measures.

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References


