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A combined muscle model and wavelet approach to interpreting the surface EMG signals from maximal dynamic knee extensions

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KEYWORDS: Complex Morlet, quadriceps, isovelocity, amplitude, mean frequency

ABSTRACT

This study aimed to identify areas of reduced surface EMG amplitude and changed frequency across the phase space of a maximal dynamic knee extension task. The hypotheses were: (1) amplitude would be lower for eccentric contractions compared to concentric contractions and unaffected by fibre length; and (2) mean frequency would also be lower for eccentric contractions and unaffected by fibre length.

Joint torque and EMG signals from the vastii and rectus femoris were recorded for eight athletic subjects performing maximum knee extensions at thirteen joint velocities spanning $\pm 250° \text{ s}^{-1}$. The instantaneous amplitude and mean frequency were calculated using the continuous wavelet transform time – frequency method, and the fibre dynamics were determined using a muscle model of the knee extensions.

The results indicated: (1) only for the rectus femoris were amplitudes significantly lower for eccentric contractions (p = 0.019), for the vastii amplitudes during eccentric contractions were less than maximal, but this was also the case for concentric contractions due to a significant reduction in amplitude towards knee extension (p = 0.023); and (2) mean frequency increased significantly with decreasing fibre length for all knee extensors and contraction velocities (p = 0.029). Using time – frequency processing of the EMG signals and a muscle model allowed the simultaneous assessment of fibre length, velocity and EMG.

KEYWORDS: Complex Morlet, quadriceps, isovelocity, amplitude, mean frequency

INTRODUCTION

Maximal voluntary movements are important skills in many sports, e.g. sprinting and jumping. During eccentric and slow concentric contractions the *in vivo* maximal voluntary force is lower than the equivalent *in vitro* tetanic force (Westing et al., 1988; Dudley et al., 1990; Westing et al., 1991). Similarly, surface electromyography (SEMG) amplitude is reduced for eccentric contractions, which has been taken to support an involuntary neural inhibition mechanism that protects the body under high force conditions (Edman, 1988; Westing et al., 1991; Seger & Thorstensson, 1994; Aagaard et al., 2000). However, for maximum voluntary concentric contractions SEMG results have shown marked inconsistencies. With increasing shortening velocity SEMG amplitude has been reported to: increase (Westing et al., 1991; Bobbert & Harlaar, 1993; Seger & Thorstensson, 1994; Aagaard et al., 2000; quadriceps); decrease (Behm & Sale, 1996; dorsiflexors); increase then decrease (Komi et al., 2000; elbow flexors); not change (Gerdle et al., 1988; quadriceps); and show muscle specific changes (Cramer et al., 2004; quadriceps). Numerous theories have been proposed to explain these observations. Increases in SEMG amplitude are linked to the eccentric neural inhibition extending to low concentric velocities (Westing et al., 1991; Seger & Thorstensson, 1994); decreases in SEMG amplitude are linked to changes in motor unit recruitment and to antagonist muscle activity (Behm & Sale, 1996); muscle specificity is linked to fibre-type distributions, muscle architecture, and tissue composition (Cramer et al., 2004). Notably, the majority of these studies have considered only joint velocity effects and not any simultaneous angle effects. The SEMG data has been averaged over either the full range of motion (Gerdle et al., 1988; Behm & Sale, 1996; Cramer et al., 2004), or a mid-angle range (Westing et al., 1991; Seger & Thorstensson, 1994; Aagaard et al., 2000). Only Komi et al. (2000) attempted to quantify both velocity and angle effects, by averaging the SEMG data over five successive 22° ranges of motion.

The effect of contraction velocity on the frequency content of SEMG signals during maximal movements has received little attention. No change (Gerdle et al., 1988; Karlsson et al., 2001) or muscle specific changes (Cramer et al., 2004) in the mean and median frequency have been reported for concentric knee extensions. Komi et al. (2000) found that above approximately $200°\ s^{-1}$, the median frequency is significantly lower for eccentric compared to concentric elbow flexions. The SEMG signals from dynamic contractions are non-stationary requiring their spectral properties to be determined using a combined time – frequency method (Karlsson et al., 2000). Traditionally the short time Fourier transform method has been employed (Gerdle et al., 1988; Komi et al., 2000), however more recently the continuous wavelet transform (CWT, described in Qian, 2002) has been used in a number of SEMG studies (Karlsson et al., 2000; Pope et al., 2000; von Tscharner, 2000; Lauer et al. 2005). The CWT is suggested to provide a better time – frequency resolution trade-off for SEMG signals based on the range of their frequency content (Akay, 1998; Karlsson et al., 2001). However, there remains some uncertainty in the application of the CWT method to SEMG signals, e.g. selecting the optimal wavelet shape for the specific resolution requirements of an investigation.

SEMG reflects the net activity of the active motor units within the detection volume of the electrodes and hence the signal characteristics are more strongly related to the local fibre kinematics than to the global joint kinematics. The two measures are not equivalent, e.g. ultrasound measurements have shown that as dorsiflexion shortening velocity is increased from isometric to $240°\ s^{-1}$, a joint angle of 8° (of plantarflexion) corresponds to a tibialis anterior fibre length that increases from 50 to 70 mm (Reeves & Narici, 2003). Hence, previous studies investigating joint velocity effects on SEMG characteristics which averaged their data over a constant joint range of motion, could have been considering different local fibre lengths at each velocity. Furthermore, SEMG signals are affected not only by motor unit recruitment and firing rate, but also by a number of physiological, anatomical and geometrical factors associated with the movement and recording process which increases the complexity of their interpretation (Farina et al., 2004; Beck et al., 2006). Ultimately SEMG cannot measure individual motor unit changes throughout the muscle, it can only reflect the net effect of all the active motor units within the electrode detection volume in the context of a physiologically, anatomically and geometrically dynamic environment. Considering this, it would seem prudent to examine the signal in light of what is simultaneously happening with the dynamics of the environment. If, for a given change in SEMG signal all but one factor, such as neural strategy (which on its own comprises a number of variables), can be kept constant then the SEMG will reflect the changes in the neural strategy alone. Functionally this is a near impossible task, but an alternative and more realistic approach is to measure, or calculate, how the factors characterising the dynamic environment are changing with time. The SEMG data could then be assessed along lines of a constant factor, e.g. estimating the local fibre kinematics and examining the results based on a constant fibre length or velocity.

This study aimed to identify areas of reduced SEMG amplitude and changed frequency across the phase space, i.e. the entire range of possible combinations of states described by fibre velocity and fibre length, of a maximal dynamic knee extension task. The methodology combined: i) maximum voluntary knee extension strength and SEMG measurements on an isovelocity dynamometer; ii) a muscle model to estimate the fibre kinematics beneath the electrodes; and iii) SEMG signal processing using the CWT method with a specific focus on selecting a wavelet shape to give appropriate time and frequency resolution properties in the processed data. It was hypothesized that: (1) amplitude would be lower for eccentric contractions compared to concentric contractions and would be unaffected by fibre length; and (2) similarly, mean frequency would be lower for eccentric contractions compared to concentric contractions and would be unaffected by fibre length.

METHODS

Subjects

Eight male volunteers (age 27.5 ± 5.0 yrs; body mass 81 ± 11 kg; height 1.77 ± 0.05 m; mean $\pm$ SD) gave their informed consent and the study was conducted in accordance with the approval given by Loughborough University Ethical Advisory Committee. All subjects were involved in sprint-based sports which included maximal strength training, and none reported injuries to the knee of their dominant leg or a history of neuromuscular disorders.

Data collection

Maximal effort knee extensions were conducted on an isovelocity dynamometer (Cybex NORM, CSMI, Stoughton, MA, USA). The subjects were seated upright and strapped across the chest and pelvis. The dominant leg was strapped tightly to the unpadded dynamometer crank arm directly above the ankle joint using a moulded plastic shin guard for protection. The chair and crank arm positional settings were adjusted to align the rotational axis of the crank with the centre of the knee joint during near maximal efforts and in each isometric trial the knee angle was measured using a mechanical goniometer.

Subjects commenced with a 10 minute self-selected warm up followed by a number of submaximal knee extensions. The main protocol comprised maximal isometric contractions at crank angles of 105° to 165° in 15° increments where 180° corresponded to full extension of the knee joint. This was followed by maximal dynamic contractions at preset crank velocities starting at 50° s^{-1} and increasing in steps of 50° s^{-1} up to a maximum of 300° s^{-1} using repeated eccentric-concentric cycles as described in Yeadon et al.(2006). An isometric and a dynamic trial were repeated to provide a measure of reliability and fatigue. Finally, maximal isometric knee flexions were conducted at 120°, 135° and 150° to provide a maximum SEMG amplitude for the biceps femoris. A rest interval of at least two minutes between trials was enforced to minimise muscle fatigue.

SEMG signals from the vastus medialis, vastus lateralis, rectus femoris and biceps femoris were recorded using an active bipolar surface electrode system (Biovision, Wehrheim, Germany). The skin was cleaned using isopropyl wipes and rectangular pre-gelled disposable electrodes (35 mm × 24 mm; Ag-AgCl; Bio-logic, Mundelein, IL, USA) were applied to the belly of each muscle parallel to the fibre direction (Delagi & Perotto, 1980; Hermens et al., 1999) with an inter-electrode distance of 24 mm. The grounded electrode was placed on the patella. The SEMG signals were amplified at source using small custom-built amplifiers taped firmly to the skin (gain setting, 3000; input impedance, 10^{12} Ω; bandwidth, 10 – 1000 Hz; and CMRR better than 100 dB).

The dynamometer and SEMG signals were recorded synchronously at 2000 Hz using a 16-bit analog-to-digital card (National Instruments, Austin, TX, USA) and stored for off-line analysis in MATLAB (The MathWorks Inc., Natick, MA, USA).

The dynamometer data were low pass filtered at 8 Hz using a fourth order zero-lag Butterworth filter. To account for differences between the crank and joint kinematics (Herzog, 1988) the crank data were converted to a joint basis using a linear regression of the goniometer angle measurements obtained in the isometric trials. The joint torque and angle data were input to a standard Hill-type muscle model comprising contractile and series elastic elements, to estimate the corresponding fibre force, length and velocity for each of the knee extensors. The force data from each dynamic trial were visually inspected and the isovelocity data from the single maximum concentric and eccentric contractions were selected for further analysis (Figure 1).
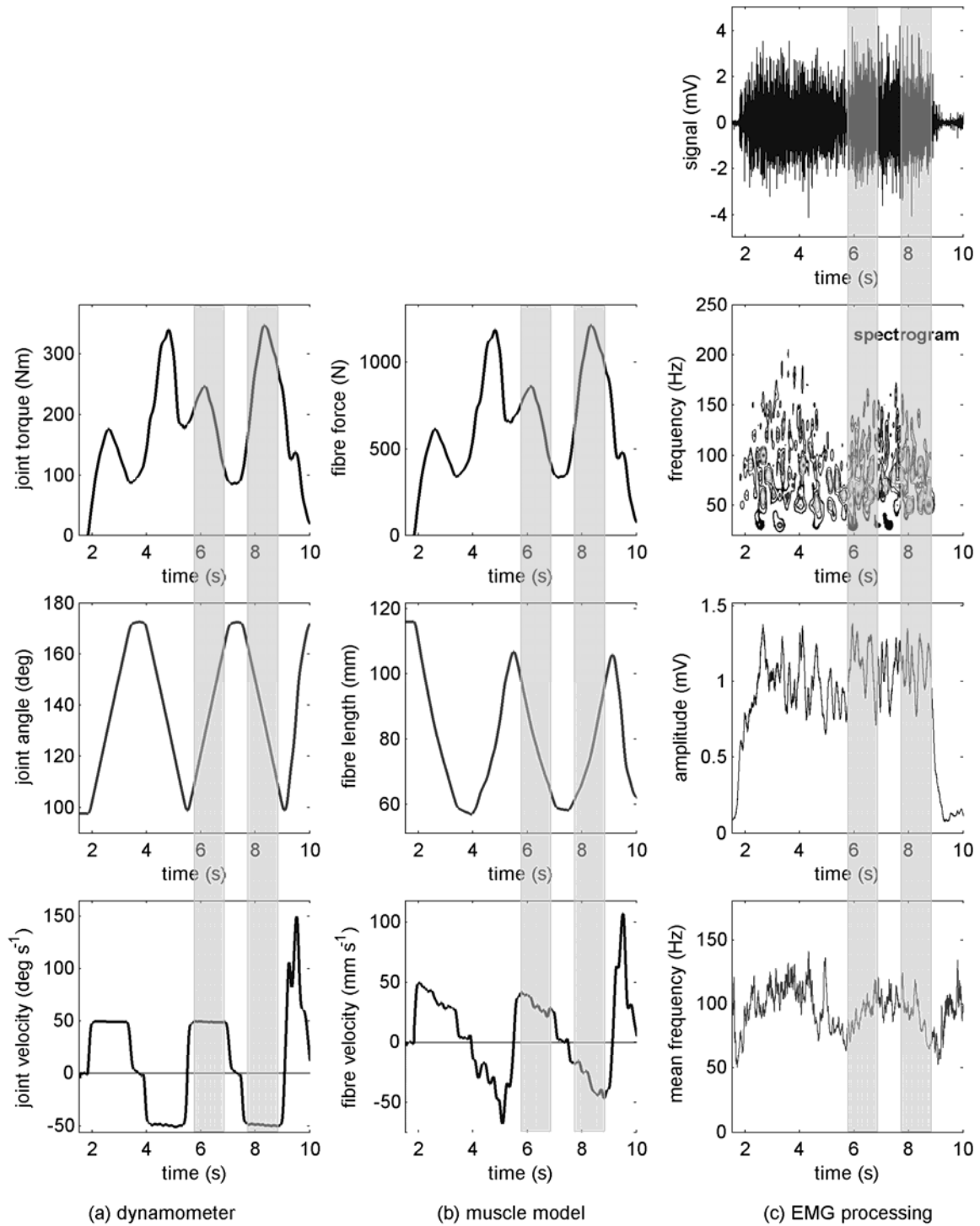
Figure 1. Typical results from a single dynamic trial (preset crank velocity = 50° s⁻¹): (a) dynamometer data on a joint basis; (b) rectus femoris fibre dynamics obtained from the muscle model; and (c) rectus femoris raw SEMG signal, CWT spectrogram, instantaneous amplitude and mean frequency. The shaded regions represent the isovelocity data for the single maximum eccentric and concentric contractions which were selected for further analysis.

CWT analysis

The SEMG signals were bandpass filtered at 10 – 500 Hz using a fourth order zero-lag Butterworth filter. The CWT time – frequency method employing the complex Morlet wavelet (Karlsson et al., 2000; von Tscharner, 2000) was used to determine the instantaneous amplitude and

frequency characteristics of the SEMG signals. The complex Morlet wavelet is described by (MATLAB wavelet toolbox, The MathWorks Inc., Natick, MA, USA):

$$\Psi(t) = \frac{1}{\sqrt{\pi f_B}} \times \exp(2i\pi f_C t) \times \exp\left(\frac{-t^2}{f_B}\right) \tag{1}$$

where $\Psi(t)$ is the mother wavelet function, f_B the bandwidth parameter, and f_C the centre frequency parameter. The time and frequency resolutions at frequency f_i (Δt_i and Δf_i respectively) for this wavelet are given by (Yan & Miyamoto, 2006):

$$\Delta t_i = \frac{f_C \sqrt{f_B}}{2 f_i} \tag{2a}$$

$$\Delta f_i = \frac{f_i}{2\pi} \frac{1}{f_C \sqrt{f_B}} \tag{2b}$$

With a corresponding joint angle resolution ($\Delta \theta_i$):

$$\Delta \theta_i = \omega \, \Delta t_i = \omega \frac{f_C \sqrt{f_B}}{2 f_i} \tag{2c}$$

where ω is the joint angular velocity. Eqs.(2) suggest that the single combined parameter value $f_C \sqrt{f_B}$, the wavelet shape factor, sets the resolution properties of the instantaneous amplitude and mean frequency estimates. Furthermore, for a given value of wavelet shape factor only angle resolution depends on joint velocity; frequency and time resolutions remain fixed across velocities. Eqs.(2) also illustrates the trade-off between time (or angle) and frequency resolution; for lower values of $f_C \sqrt{f_B}$ improved temporal (or angular) resolution is offset by poorer frequency resolution and vice versa.

A suitable value for the wavelet shape factor was investigated by defining angular, temporal and frequency resolutions for the knee extension task (10°, 100 ms and 10 Hz respectively over the frequency range 20 – 250 Hz). These resolutions were selected to enable length and velocity effects to be investigated over the range of experimental conditions. Using the resulting wavelet, the instantaneous amplitude and mean frequency were obtained from the spectrograms determined over the frequency range 10 – 500 Hz using a non-uniform distribution of 112 scales concentrated between 20 and 250 Hz. The instantaneous amplitude was normalised based on the maximum value obtained in all isometric and dynamic trials. The SEMG data corresponding to the selected isovelocity data from the muscle model for the single maximum concentric and eccentric contractions were then identified (Figure 1).

These data processing methods resulted in the maximum force, length, velocity and corresponding instantaneous amplitude and mean frequency data for each of the vastus medialis, vastus lateralis and rectus femoris at thirteen isovelocities (crank setting: 300° s[-1] eccentric through to 300° s[-1] concentric).

Statistical analyses

The results were group averaged and the effects of fibre length and contraction velocity on the amplitude and mean frequency were examined based on their values at three fibre lengths and seven contraction velocities spanning the range of experimental conditions. Two-way (length and eccentric velocity, and length and concentric velocity) within subjects repeated measures ANOVAs were applied to the data, and for significant F-ratios the difference between means was assessed using a Bonferroni post-hoc test. Paired samples t-tests were used to examine the effect of contraction mode (eccentric and concentric) on force, amplitude and mean frequency. A statistical

significance level of p ≤ 0.05 was assumed throughout and all statistical procedures which were conducted using SPSS version 13.0 (SPSS Inc., Chicago, IL, USA).

RESULTS

SEMG amplitudes were only significantly lower for eccentric compared to concentric contractions for the rectus femoris (Figures 2(b) – 3(b); p = 0.019). SEMG amplitudes for the vastii decreased significantly with increasing eccentric velocity (Figure 2(b) – 3(b); p = 0.017) with the magnitude of decrease being 15 – 20% over the velocity range -25 to -100 mm s^{-1}. SEMG amplitudes were not affected by fibre length for eccentric contractions but for the vastii decreased significantly with reducing fibre length (towards knee extension) for concentric contractions (Figures 2(b) – 3(b); p = 0.023). The magnitude of the decrease was 15 – 25% while the fibre shortened from 100 – 70 mm. Concentric velocity only had a significant effect on SEMG amplitude for the rectus femoris and only at mid fibre lengths where SEMG amplitude decreased by around 20% with increasing shortening velocity from 25 – 100 mm s^{-1} (p = 0.041). These results indicate that the effects of contraction mode and fibre length on SEMG amplitude are muscle specific, and in particular differ for the rectus femoris compared to the vastii.

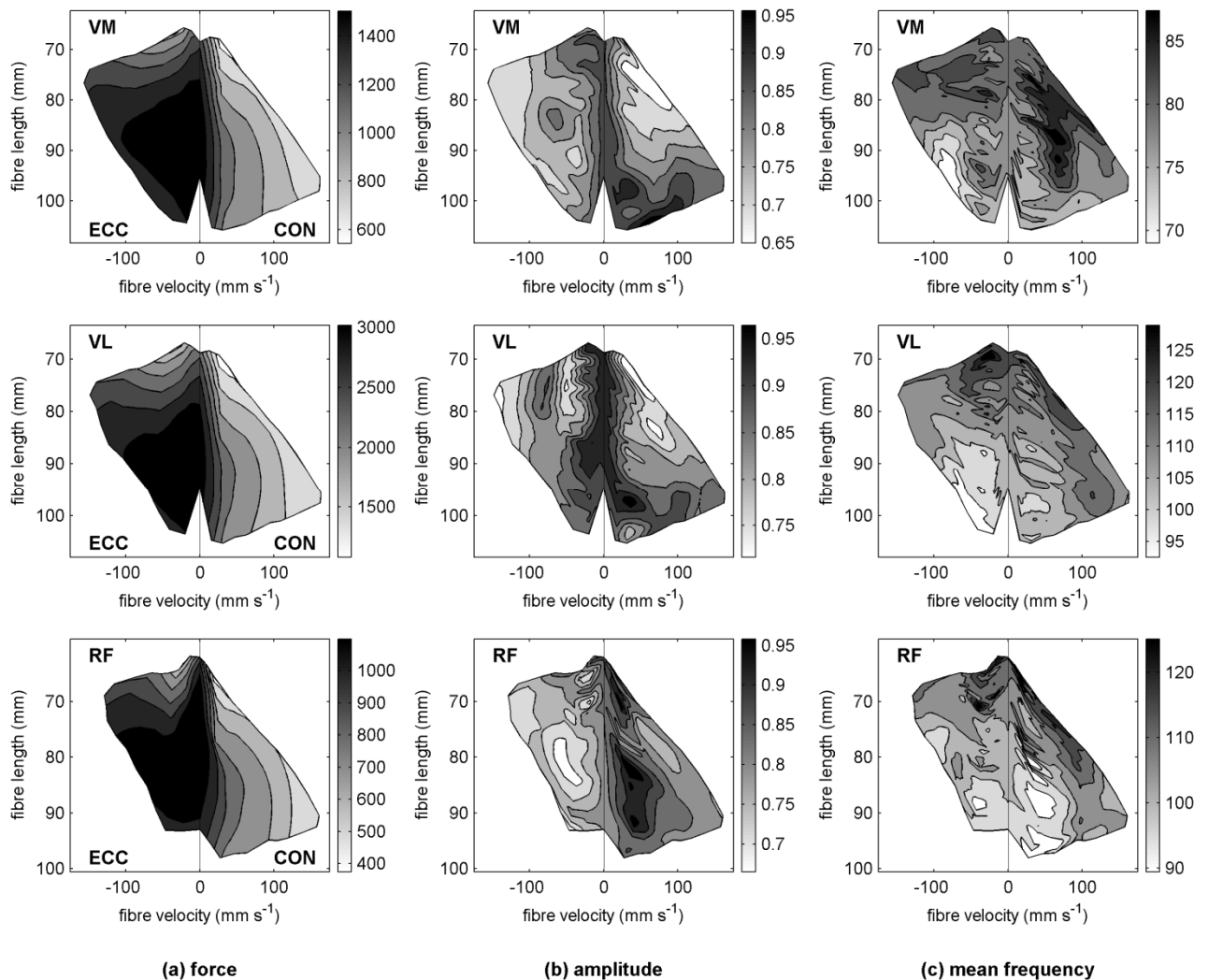Figure 2. Contour graphs showing the group average: (a) fibre force; (b) SEMG amplitude; and (c) mean frequency as a function of fibre length and velocity for the vastus medialis (VM), vastus lateralis (VL) and rectus femoris (RF). The results are limited to the isovelocity range of motion to minimise crank turnaround effects and this decreased from 72° at isometric to 12° at the highest velocity (Table 1).

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Figure 3. Bar charts showing the group average mean ± standard deviation: (a) fibre force; (b) SEMG amplitude; and (c) mean frequency. Based on three fibre lengths (Lshort = black; Lmid = gray; Llong = white) and seven contraction velocities (Hecc, Mecc, Lecc = high, mid and low eccentric; ISOM = isometric; Lcon, Mcon, Hcon = low, mid, high concentric). Regions with statistically significant ($p \leq 0.05$) velocity effects are marked (*V*); and length effects are marked (*L*) as indicated in the text. Note that where no bar exists this indicates that there was no isovelocity data for these conditions, reflecting the tapered shape of the contour plots towards high velocities.
Vastus Medialis (VM) and Vastus Lateralis (VL):

Mean frequency increased significantly with decreasing fibre length for all three knee extensors and for both eccentric and concentric velocities (Figures 2(c) – 3(c); $p = 0.029$). The magnitude of the increase was between 5 – 20% while the fibre shortened from 100 – 70 mm. Mean frequency was not affected by contraction mode or contraction velocity.

Maximum voluntary forces for all three knee extensors were significantly higher for eccentric compared to concentric contractions (Figures 2(a) – 3(a); $p = 0.008$). However, eccentric forces peaked at only 5% above the isometric level, and decreased with increasing lengthening velocity.

Table 1. Summary of the isovelocity dynamometer results (group mean ± standard deviation).

Preset crank velocity ($^\circ$ s^{-1})	-300	-250	-200	-150	-100	-50	0	50	100	150	200	250	300
Measured crank velocity ($^\circ$ s^{-1})	-268	-233	-191	-146	-98	-49	0	50	98	146	192	235	275
Joint velocity ($^\circ$ s^{-1})	-259 ± 13	-225 ± 12	-184 ± 9	-141 ± 8	-95 ± 5	-48 ± 3	0	48 ± 3	95 ± 5	142 ± 8	186 ± 10	227 ± 12	266 ± 12
Isovelocity range of motion ($^\circ$)	12 ± 1	18 ± 4	29 ± 3	41 ± 3	53 ± 2	66 ± 2	72 ± 2	65 ± 2	53 ± 2	41 ± 3	29 ± 3	18 ± 4	12 ± 1
Maximum joint torque (Nm)	256 ± 45	268 ± 53	290 ± 52	311 ± 47	322 ± 59	335 ± 58	321 ± 71	234 ± 46	203 ± 45	185 ± 41	161 ± 41	149 ± 34	139 ± 29

The effect of using muscle kinematics as opposed to joint kinematics to assess the SEMG characteristics is illustrated in Figure 4. These graphs show SEMG amplitude and mean frequency as a function of fibre length and velocity for the rectus femoris onto which white lines of constant joint angle have been plotted. Differences in kinematic behaviour between the muscle fibres and the joint are represented by deviations from the horizontal in the white lines; as observed around isometric and for concentric velocities. A constant joint angle corresponded to a progressively longer fibre length as shortening velocity increased, e.g. at isometric a joint angle of 118° corresponded to a fibre length of 81 mm whilst at 140 mm s^{-1} the same joint angle corresponded to a fibre length of 93 mm.
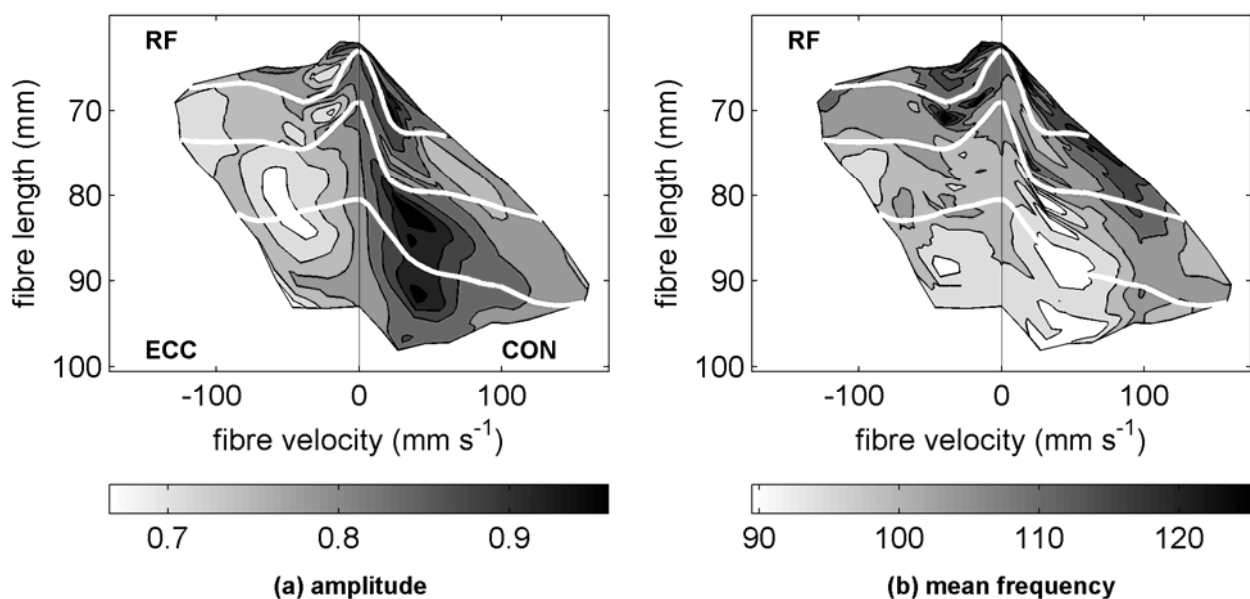

Figure 4. Contour graphs showing the group average: (a) SEMG amplitude; and (b) mean frequency as a function of fibre length and velocity for the rectus femoris (RF) across the isovelocity range of motion. The white lines represent a constant joint angle of: 118° (top line); 130° (middle line); and 142° (bottom line), where 180° is full extension.

Prior to processing of the SEMG signals, a suitable shape for the Complex Morlet wavelet had to be defined. For this general investigation into the effects of fibre length and contraction velocity on the SEMG signal characteristics, a wavelet shape factor of 4.0 was selected to give resolutions in frequency (<10 Hz), time (< 100 ms), and angle (<15°) over the frequency range 20 to 250 Hz and the velocity range 0 to 250° s^{-1}. Note that this did not meet the original resolution specification, but represented the best compromise for the given range of velocities (Table 2 and Figure 5). Individual values for f_B and f_C in eq.(1) were determined using the method described by Yan & Miyamoto (2006) which gave 7.0 and 1.5 respectively. This wavelet design was used to process all the SEMG signals in this study.

The SEMG amplitudes for the biceps femoris were less than 0.15 of their measured maximum voluntary value during all knee extension trials and subjects, suggesting minimal antagonist muscle activity. The repeated trials gave results within 10% of the originals, confirming that fatigue effects were small.

Table 2. Range of suitable wavelet shape factor (WSF) values for different types of investigation and contraction velocities (based on an SEMG signal with frequencies concentrated in the range 20 – 250 Hz).

Investigation type		Time-based	Angle-based	General
Resolution limits	frequency	10 Hz	10 Hz	10 Hz
	time	100 ms	n/a	100 ms
	angle	n/a	10°	10°
Lowest velocity trial	WSF range (optimal WSF)	4.0 (4.0)	4.0 – 8.6 (5.9*)	4.0 (4.0)
Highest velocity trial	WSF range (optimal WSF)	4.0 (4.0)	– (3.6**)	– (4.0**)

* obtained by minimising the sum of the frequency resolution (in Hz) and the angle resolution (in degrees).

** no wavelet shape factor could satisfy the specified resolutions and the optimal value was determined based on the best resolution compromise defined as minimising the frequency range lost at either end of the 20 – 250 Hz spectrum.
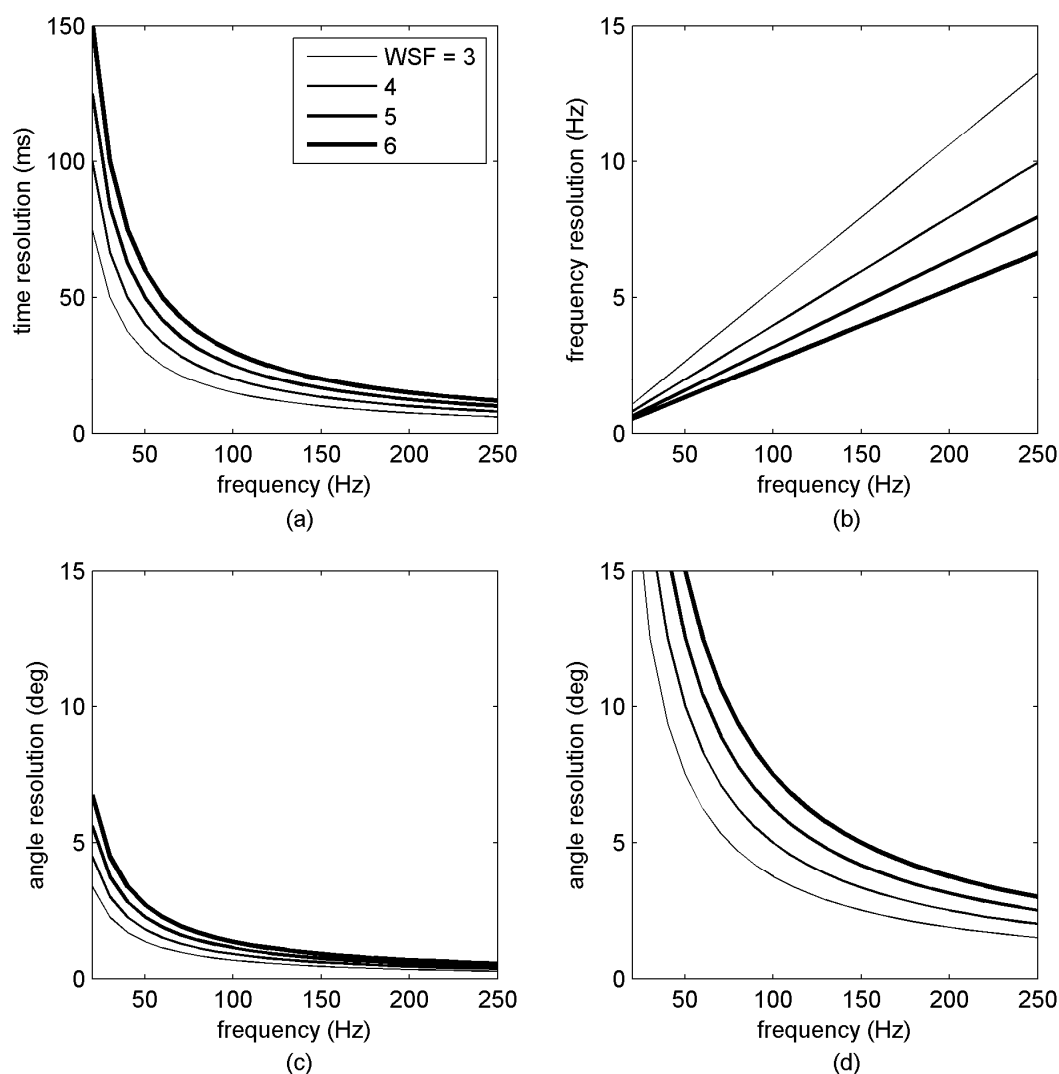
Figure 5. Time, frequency and angle resolutions as a function of frequency for the complex Morlet wavelet with wavelet shape factor values of 3, 4, 5 and 6: (a) time resolution; (b) frequency resolution; (c) angle resolution in the low velocity trial (preset crank velocity = 50° s^{-1}); (d) angle resolution in the high velocity trial (preset crank velocity = 300° s^{-1}).

DISCUSSION

The hypotheses, relating to the effect of fibre velocity and length on the SEMG amplitude and mean frequency, were only partially supported by the results of this study. SEMG amplitudes were lower in eccentric mode for the rectus femoris only, and showed significant reduction towards knee extension during concentric actions for the vastii. Mean frequencies were unaffected by contraction mode but increased significantly towards knee extension for all three knee extensors and all contraction velocities.

Both muscle specificity with regard to the effect of shortening velocity on SEMG amplitude (Cramer et al., 2004), and a significant decrease in SEMG amplitude towards knee extension for the vastii have been observed previously (Aagaard et al., 2000; Croce & Miller, 2006). The latter has been suggested to be a neural response to protect the knee as it approaches full extension. It was not possible to quantify how much the maximum voluntary forces deviated from tetanic in this region of reduced SEMG amplitude near full extension since the forces were relatively low and

therefore any deficit would correspond to small absolute force values (Figure 3(a)). In support of a deficit between maximum voluntary and tetanic force, Westing *et al.* (1990) have reported that knee torque decreases less rapidly towards knee extension when electrical stimulation supplements maximum voluntary. Only the SEMG amplitude of the vastii and not the rectus femoris decreased towards knee extension (Figure 3(b)). This suggests differing roles for the vastii and rectus femoris in controlling knee joint forces and possibly reflects the *in vivo* architecture of the muscles, where the larger series elastic element of the biarticular rectus femoris (Jacobs et al., 1996) make it less suited to such a role.

The trend of increasing mean frequency towards knee extension for all velocities and knee extensors is in agreement (Gerdle et al., 1988; Aagaard et al., 2000; Karlsson et al., 2001) or partial agreement (Cramer et al., 2004) with previous studies. The increase in mean frequency measured in this study did not correlate with the observed deviations between maximum voluntary and tetanic force, nor the areas of changed SEMG amplitude, and is therefore unlikely to reflect neural control but rather the changing architecture or physiology of the system.

Maximum voluntary eccentric forces peaked only marginally above the isometric value and well below the ratio of 1.5 observed for the tetanic stimulation of isolated muscle fibres (Edman et al., 1982). The significant decreases in force and SEMG amplitude with increasing lengthening velocity are in good agreement with previous studies on maximum voluntary knee extensions (Westing et al., 1988, Westing et al., 1991; Seger & Thorstensson, 1994; Aagaard et al., 2000).

The present results generally support previous investigations into the effect of velocity on the force and SEMG characteristics of maximal knee extensions. However, by considering both length and velocity effects on the SEMG characteristics on a muscle fibre basis, the results presented herein may also help to explain the origin of inconsistencies concerning the effect of shortening velocity on SEMG amplitude (Gerdle et al., 1988; Westing et al., 1991; Bobbert & Harlaar, 1993; Seger & Thorstensson, 1994; Aagaard et al., 2000; Cramer et al., 2004). This is illustrated in the contour graphs of Figures 2 and 4 which depict the changes in SEMG amplitude with length and velocity simultaneously. For the rectus femoris (Figure 4) the effect of shortening velocity on the SEMG amplitude displayed a significant dependence on fibre length. At mid-lengths the SEMG amplitude decreased significantly with increasing shortening velocity; however at longer or shorter fibre lengths the SEMG amplitude was unaffected by shortening velocity. For studies that considered only velocity effects the results would depend on the angle range used to average the SEMG data, with a central angle range potentially producing different results to a full angle range. Hence the present methods and results indicate that operating within different regions of the SEMG amplitude – length – velocity contour graph influences the observed effect of shortening velocity.

What is likely to be particularly useful from the current approach, is the ability to look at the relationship between SEMG and force along a line of constant fibre length whilst velocity changes. A constant fibre length corresponds to a constant muscle thickness, an approximately constant pennation angle and an approximately constant thickness of surrounding active muscles in synergistic muscle groups. Using the current model and set up, for a fixed fibre length the electrodes may not necessarily lie above the same part of the muscle throughout the movement task leading to the possibility of innervation and tendon zones affecting the SEMG signals (Farina et al., 2004). Also, the intervening layers may not necessarily have the same thickness and hence signal attenuating properties. However, investigating velocity effects on SEMG characteristics based on a constant joint angle, the fibre length and every factor listed above will vary and affect the SEMG signals, making it very difficult to decipher neural implications from the results. Examining the SEMG amplitude and mean frequency as a function of velocity along a line of constant fibre length controls for more of the factors affecting the signal and is therefore likely to provide a better basis for assessing the underlying neural implications of the results.

It has recently been shown that useful information on neural strategy can be obtained from the amplitude and frequency content of SEMG signals recorded during dynamic contractions (Wakeling, 2007). In this study the SEMG results for the vastii gave significant trends of increasing

mean frequency and decreasing amplitude towards knee extension (fibre shortening) across all concentric velocities. In combination these suggest that the mean frequency increase is more likely to be associated with changing fibre length rather than increased motor unit recruitment, and that any neural involvement in the SEMG amplitude decrease is more likely to be associated with a reduction in firing rate. The SEMG amplitudes were all high (≥ 0.70) which suggests that the majority of motor units were recruited throughout the knee extensions (Enoka & Fuglerand, 2001) which further supports the SEMG amplitude decrease to be associated with changes in firing rate (De Luca, 1997). Similarly, the significant decrease in SEMG amplitude with increasing eccentric velocity is also more likely to be a result of a reduction in firing rate, since the mean frequency is unaffected by lengthening velocity and again the SEMG amplitude values were all high (≥ 0.70).

A simple muscle model was used to estimate the fibre forces and kinematics. However, confidence in the results can be gained from the predicted trends in fibre versus joint kinematics reflecting those reported in the literature from *in vivo* measurements. For example, the model indicated that the same joint angle corresponded to a progressively longer fibre length as shortening velocity increased and a similar trend has been measured in the tibialis anterior by Reeves & Narici (2003). Alternatively, ultrasound could be used to measure the subject-specific changes in muscle architecture and kinematics, and thickness of the intervening layers during the contractions. Surface array electrodes represent a means of confirming that the electrode signals are not affected by tendonous or innervation regions during the contractions. These techniques would provide further understanding on non-neural influences on the SEMG signals and thereby allow greater confidence in inferring the nature of neural contributions.

Time – frequency processing of the SEMG signals allowed the simultaneous effects of length and velocity to be assessed. Once the maximum voluntary force – length – velocity surface for the muscle had been obtained then determining the SEMG over the same phase space with suitable resolutions would seem a logical corollary. Although the CWT has previously been used for processing SEMG signals (Pope et al., 2000; Karlsson et al., 2001; Lauer et al., 2005) little information existed on the choice of wavelet shape. Discriminative SEMG amplitude and frequency data throughout the phase space were only obtained after careful selection of this parameter. Indeed the results indicated that this is an important and not necessarily straightforward selection. For the study of time-based events a single value for the wavelet shape factor can give instantaneous amplitude and mean frequency estimates with the required time and frequency resolution properties independent of contraction velocity. In contrast, for angle-based events a velocity dependent value for the wavelet shape factor provided the optimal resolution properties, since at high velocities angle resolution worsens rapidly, i.e. to much greater than 10° at velocities above 200° s^{-1}. For such applications, one solution would be to run the SEMG signals through the CWT analysis more than once using different wavelet shape factors to investigate different events or narrower frequency bands, however the computational expense of such an approach would be high.

In conclusion, this study supports two previously identified areas of reduced maximum voluntary forces and SEMG amplitudes for maximum voluntary knee extensions. These are for eccentric contractions where forces are lower than tetanic and SEMG amplitudes are less than maximal, and for concentric contractions when the knee approaches full extension SEMG amplitudes of the vastii drop below maximal. Thus, both fibre length and contraction velocity affect the SEMG characteristics of maximal knee extensions and in a muscle specific manner. Notably, the effect of shortening velocity can depend on the region of the graph under investigation which may explain the inconsistent results in previous studies which used different angle ranges to investigate velocity effects. These observations were dependent on using a CWT analysis of the SEMG signals which employs frequency dependent windowing to enable the frequency content to be assessed as a function of time, and a muscle model to allow the SEMG characteristics to be related to the local fibre dynamics. In combination these appear to represent a more reliable method of assessing length and velocity effects on the SEMG characteristics of dynamic contractions

compared to previous studies since the length, velocity and position of the muscle fibres were known at the same time as specific frequency and amplitudes events.

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