J Neurophysiol 110: 1744–1750, 2013.
First published July 31, 2013; doi:10.1152/jn.00296.2013.

Rectification of EMG in low force contractions improves detection of motor unit coherence in the beta-frequency band

Nicholas J. Ward,[1,2] Simon F. Farmer,[1,3] Luc Berthouze,[2,4] and David M. Halliday[5]

[1]*Sobell Department for Motor Neuroscience, Institute of Neurology, University College London, London, United Kingdom;* [2]*Centre for Computational Neuroscience and Robotics, University of Sussex, Brighton, United Kingdom;* [3]*National Hospital for Neurology and Neurosurgery, University College London, London, United Kingdom;* [4]*Institute of Child Health, University College London, London, United Kingdom; and* [5]*Department of Electronics, University of York, York, United Kingdom*

Submitted 26 April 2013; accepted in final form 25 July 2013

Ward NJ, Farmer SF, Berthouze L, Halliday DM. Rectification of EMG in low force contractions improves detection of motor unit coherence in the beta-frequency band. *J Neurophysiol* 110: 1744–1750, 2013. First published July 31, 2013; doi:10.1152/jn.00296.2013.—Rectification of surface EMG before spectral analysis is a well-established preprocessing method used in the detection of motor unit firing patterns. A number of recent studies have called into question the need for rectification before spectral analysis, pointing out that there is no supporting experimental evidence to justify rectification. We present an analysis of 190 records from 13 subjects consisting of simultaneous recordings of paired single motor units and surface EMG from the extensor digitorum longus muscle during middle finger extension against gravity (unloaded condition) and against gravity plus inertial loading (loaded condition). We directly examine the hypothesis that rectified surface EMG is a better predictor of the frequency components of motor unit synchronization than the unrectified (or raw) EMG in the beta-frequency band (15–32 Hz). We use multivariate analysis and estimate the partial coherence between the paired single units using both rectified and unrectified surface EMG as a predictor. We use a residual partial correlation measure to quantify the difference between raw and rectified EMG as predictor and analyze unloaded and loaded conditions separately. The residual correlation for the unloaded condition is 22% with raw EMG and 3.5% with rectified EMG and for the loaded condition it is 5.2% with raw EMG and 1.4% with rectified EMG. We interpret these results as strong supporting experimental evidence in favor of using the preprocessing step of surface EMG rectification before spectral analysis.

partial coherence; EMG rectification; surface EMG; motor unit

SINGLE MOTOR UNITS (MUs) from human muscles activated during contraction can be treated as stochastic point processes (Halliday et al. 1995). Simple level detection and other identification techniques can determine spike times with millisecond accuracy, and from these, time and frequency domain analyses may be used to characterize the statistics of MU firing: the interspike interval histogram, the autocorrelation function, and the autospectral density function. The existence of peaks in the autospectral density indicates departure from the asymptotic value (Bartlett 1963) and is thus indicative of structure within the times of occurrence of the MU spikes. Such peaks typically result from the mean MU firing rate and any modulatory influences on the firing rate. Through recording simultaneous activity in pairs of MUs, further information about the common cortical drive to human motor neurons may be ascertained. In the time domain, MU synchrony can be detected as a peak centered around time zero (Datta and Stephens 1990). In the frequency domain, the cross-spectrum, the coherence, and the phase may be extracted. Such approaches have identified important physiological processes underlying normal and abnormal beta-range (15–32 Hz) central common drive to human MUs (Farmer et al. 1993) and have also been used fruitfully to explore physiological and pathological tremors and other disorders of movement (Elble and Randall 1976; Halliday et al. 1999). These methods can be extended to examine directly the communication between the sensorimotor cortex and the muscle through calculation of corticomuscular coherence (CMC). Various frequencies of CMC have been identified, with the beta-range being especially important during steady muscle contraction [see Salenius and Hari (2003) for a review].

Providing there is adequate signal separation, two surface EMG signals can be used as a substitute to pairs of MU recordings, and numerous studies have identified the same oscillatory drive as shown in pairs of single MUs in simultaneous EMG-EMG recordings (Farmer et al. 2007). In contrast to MUs treated as point processes, the surface EMG is a complex signal resulting from the superposition of a large number of spatially and temporally summated MUs, affected by volume conduction and with both the positive and negative phases of the action potential contributing to the signal (Farina et al. 2004). To access the timing information inherent in the surface EMG signal, filtering and rectification preprocessing steps have been used before calculation of time and frequency domain measures (Elble and Randall 1976). However, recent modeling studies (Neto and Christou 2010; Stegeman et al. 2010) and experimental work (McClelland et al. 2012) have challenged this approach, and the claim has been made that rectification is "inappropriate" and impairs detection of common drive to motor pools (McClelland et al. 2012). It may be argued that simply finding a stronger EMG-EMG coherence or CMC when using an unrectified as opposed to a rectified EMG signal does not mean that rectification "impairs" common drive detection. Rather what is required is a direct measure of the fidelity of unrectified vs. rectified EMG in the detection of the underlying common drive to the MUs.

In this article, we analyze data in which surface EMG and pairs of single MUs were simultaneously recorded (Halliday et al. 1999). Using the frequency domain approach of the autospectral density function, we directly compare the spectra of

Address for reprint requests and other correspondence: D. M. Halliday, Dept. of Electronics, Univ. of York, York, YO10 5DD, UK (e-mail: david.halliday@york.ac.uk).

unrectified and rectified EMGs with those of simultaneous single MU firing. Furthermore, through calculation of the coherence and partial coherence between the MUs with the surface EMG as predictor we are able to ascertain whether the unrectified or the rectified surface EMG signal best represents the common drive to pairs of single MUs. The following sections outline the methods, describe results of the partial coherence analysis, and discuss, with recommendations, the best strategy for EMG preprocessing in frequency domain studies involving the use of surface EMG.

METHODS

The data set analysed in the present study comes from the work of Halliday et al. (1999), where full details of the experimental protocol can be found. Here we present a novel analysis of these data focusing on a partial regression analysis to determine if the raw or rectified EMG is a better predictor of the components of MU synchronization. To achieve this we use estimates of pooled partial coherence (Amjad et al. 1997). The data set consists of simultaneous recordings of bipolar surface EMG, paired MU recordings, and physiological tremor from 13 healthy adult subjects. There were two task conditions, an unloaded condition involving a steady postural contraction of the extensor digitorum communis muscle against gravity, and a loaded condition with increased inertial loading using small weights (5–40 g) added to the extended finger. The surface EMG was band pass filtered (3–500 Hz), and MU firing times were determined using online window discrimination devices that output TTL pulses (sampled with a time step of 1 ms). Full details of the experimental procedure are in the original study (Halliday et al. 1999).

The frequency analysis uses the techniques described in Halliday et al. (1995). MU spike trains are treated as stochastic point-process data allowing individual autospectra and cross-spectra to be calculated. These were estimated using the average periodogram approach described in Halliday et al. (1995). MU and EMG spectra were then pooled over all 106 records in the population of recordings made in the unloaded condition, using a total of 150 min of data from 13 subjects. Similarly, 84 records comprising 125 min of data from 13 subjects were pooled for the loaded condition. The pooling technique is described in Amjad et al. (1997), and EMG spectra were calculated for both unrectified (raw) and rectified (full-wave) EMG.

Once pooled auto- and cross-spectra were calculated, the pooled MU coherence and pooled MU partial coherence were estimated. Two partial coherence estimates were constructed, one using unrectified EMG as predictor the second using rectified EMG as predictor. Coherence measures the degree of linear association between two processes. The coherence between signals x and y is given by

$$\left|R_{xy}(\lambda)\right|^2 = \frac{\left|f_{xy}(\lambda)\right|^2}{f_{xx}(\lambda)f_{yy}(\lambda)}$$

where $f_{xy}(\lambda)$ indicates the cross-spectrum of x and y at frequency λ and $f_{xx}(\lambda)$ indicates the autospectrum of x.

Partial coherence provides a measure of the correlation between two processes, after taking into account any linear interaction between the two processes and a third process (the predictor). The partial cross-spectra between x and y, with z as a predictor, are defined as per Halliday et al. (1995):

$$f_{xy/z}(\lambda) = f_{xy}(\lambda) - \frac{f_{xz}(\lambda)f_{zy}(\lambda)}{f_{zz}(\lambda)}$$

The partial autospectra of x given z as a predictor are defined as per Halliday et al. (1995):

$$f_{xx/z}(\lambda) = f_{xx}(\lambda) - \frac{\left|f_{xz}(\lambda)\right|^2}{f_{zz}(\lambda)}$$

The other partial autospectrum, $f_{yy/z}(\lambda)$, is defined similarly. From the partial spectra, the partial coherence between x and y with z as a predictor, $|R_{xy/z}(\lambda)|^2$, can then be estimated in a similar manner to the ordinary coherence function:

$$\left|R_{xy/z}(\lambda)\right|^2 = \frac{\left|f_{xy/z}(\lambda)\right|^2}{f_{xx/z}(\lambda)f_{yy/z}(\lambda)}$$

Partial coherence, like ordinary coherence, provides a bounded measure of linear association between 0 and 1. The result can be interpreted as the level of residual coupling between x and y at frequency λ after removal of the common effects of process z. For the present data, partial coherence is used to quantify to what degree MU synchronization (coherence) can be accounted for by rectified or unrectified surface EMG. Pooling of partial coherence estimates is achieved by a direct extension of the framework presented in Amjad et al. (1997), where partial auto- and cross-spectral estimates are pooled in the same way as ordinary auto- and cross-spectra. An adjustment to take into account the loss of 1 degree of freedom from using a single linear predictor is required. In the pooled framework (Amjad et al. 1997), the number of segments in each record L is replaced by $L - 1$.

Confidence limits were calculated for both coherence and partial coherence as in Halliday et al. (1995). These are used to guide interpretation of results. If the partial coherence estimate falls below the confidence level at some frequency, this indicates the surface EMG predictor (either rectified or unrectified) accounts for all the MU correlation at that frequency. A partial-coherence estimate where the values are consistently below the confidence limit over the frequency range of interest, 15–32 Hz in this case, is clear evidence that the surface EMG can account for all components of MU synchronization in this frequency band. Partial coherence thus provides a quantitative means of assessing the relative merits of raw vs. rectified EMG in predicting the components of MU synchronization.

To quantify the differences and assess statistical significance in partial coherence estimates with rectified and raw EMG as predictors, we calculate two measures. The first is based on the concept of the residual variance bound (Priestley 1981). We calculate the percentage of residual partial coherence in the beta-frequency band that is left after removal of the linear effects of the EMG signal. This is expressed as a percentage of the original MU coherence in the beta-frequency band. The second measure is a statistical test to determine if there is a significant difference in the magnitudes of the two partial coherence estimates across the beta-frequency range. This uses the difference of coherence test outlined in Rosenberg et al. (1989) to determine if the partial coherence estimates are significantly different at each frequency. To calculate these metrics we use a multitaper spectral approach (Percival and Walden 1993) tuned to cover the beta-frequency range in a single frequency bin. The values used are $NW = 4.5$, $\Delta f = 1/0.512$ Hz, and using $K = (2NW - 1)$ tapers gives an effective bandwidth ~16 Hz. Selecting the nearest Fourier frequency to the center of the beta range, 23.44 Hz in this case, allows the residual correlation and significance test across the beta-frequency range to be undertaken using the values from that single frequency. P values are calculated using the two-sided significance test in Rosenberg et al. (1989). The multitaper analysis is used to calculate the residual variance metrics and perform the statistical tests, all plots use the average periodogram estimates as in Halliday et al. (1999), with $\Delta f = 1/1.024$ Hz.

RESULTS

Pooled spectra of surface EMG and MUs. Pooled autospectral estimates for unrectified EMG, rectified EMG, and single MU firing times are shown in Fig. 1. These are illustrated separately for the unloaded (Fig. 1, *top*) and loaded (Fig. 1, *bottom*) populations. The interpretation is similar in both cases.

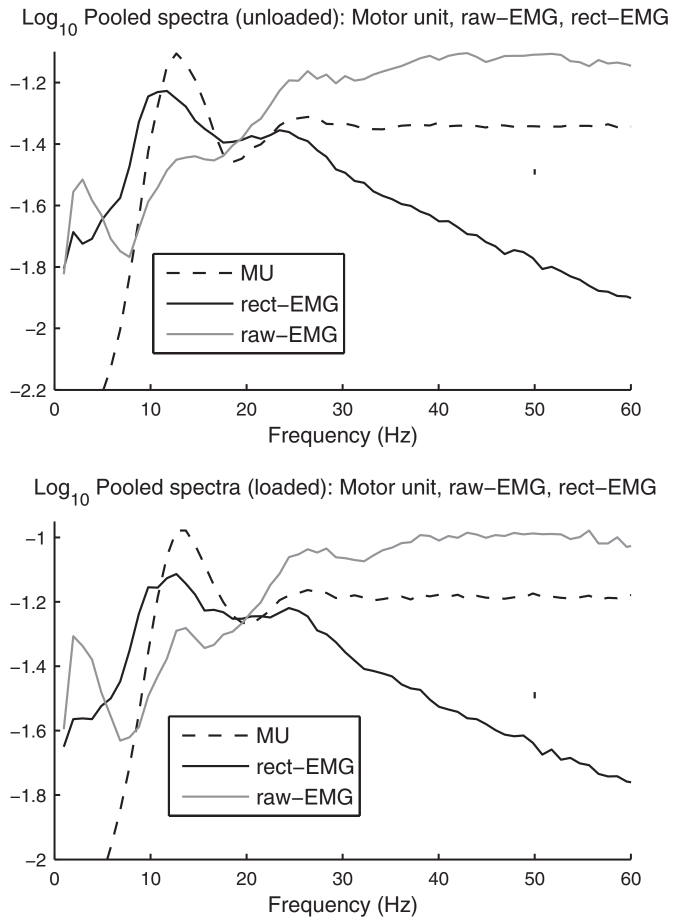

Fig. 1. Estimated pooled autospectra of motor-unit (MU; dashed line), and simultaneously recorded surface EMG, both unrectified and rectified (solid lines), for unloaded population (*top*) and loaded population (*bottom*). MU spectrum estimate is calculated using both single units from each of the 106 records (unloaded) or 84 records (loaded), using a total of 212 (unloaded) or 168 (loaded) separate spike trains. Spectral estimates are plotted on a log scale up to 60 Hz, the MU spectral estimate has been shifted vertically to facilitate comparison with the EMG spectral estimates (see text for details). Small vertical line at *right* represents the magnitude of the confidence band for the pooled spectral estimates, 0.018 dB (unloaded), or 0.02 dB (loaded).

The estimate for the pooled MU spike train is a point-process spectral estimate, and the estimates for the EMG are time-series spectral estimates. While these will have a different interpretation, inclusion of all three on the same plot can be instructive in identifying where the dominant spectral features are located. The mean firing rate for the population of MUs (11.9 spikes/s for unloaded and 12.5 spikes/s for loaded) determines the asymptotic value in the point process spectrum, according to the model presented in Bartlett (1963). For the present data, this level is −2.72 dB for the unloaded population and −2.70 dB for the loaded population on a $\log_{10}$ scale. To allow comparison with the EMG time series data the point process spectrum has been plotted with a constant offset such that the peak value matches the peak value in the autospectral estimate of the unrectified EMG pooled spectrum. This constant offset aids the visual comparison and does not affect interpretation of the point process spectrum. The standard model for interpretation of point-process spectral estimates is a Poisson (or random) sequence of events. The asymptotic value is $\log_{10}(P_N/2\pi)$, where P_N is the mean rate calculated as spikes per time step (1 ms), in our case $P_N = 0.0119$ for the unloaded data and $P_N = 0.0125$ for the loaded data. Departures from the asymptotic or expected value can be taken as evidence of non-Poisson behavior at a particular frequency, i.e., the spike train is nonrandom. For the MU spike train spectral estimate, there are significant departures from Poisson behavior <35 Hz. The main features are dominant spectral peaks ~12 Hz and smaller peaks ~26 Hz seen in both the loaded and unloaded case. The first reflects the periodic firing of the MUs, whereas the second may be the second harmonic or a distinct component indicative of higher frequency modulation of firing rate. There is no evidence of any significant structure in the MU spike train at frequencies >35 Hz. Note that this interpretation is not altered by applying a fixed offset when plotting the MU pooled spectral estimate.

The unrectified surface EMG in both cases contains a broad distribution of power, predominantly concentrated in the 40–60 Hz range. A peak is discernible at the mean MU firing frequency, and a more substantial component is present at ~3 Hz. In contrast, both the rectified surface EMG spectra have a peak at 12 Hz, and a secondary peak at ~24 Hz, with gradually decreasing power at higher frequencies. A comparison of the unloaded and loaded pooled surface EMG spectral estimates with the corresponding pooled single MU spectral estimate shows that most power in the unrectified EMG spectrum is concentrated at frequencies >35 Hz, the region where there is no obvious rhythmic structure in the MU point process spectrum. In contrast, the peak in the rectified EMG spectrum in both unloaded and loaded conditions matches more closely the peaks in the MU spectrum for the same conditions. This suggests that for the low contraction levels in the present study, the rectified EMG signal is better able to capture the rhythmic components of MU firing than its unrectified counterpart (Farina et al. 2013). However, it is well known that spectral peaks are not necessarily an indicator of correlation (Rosenberg et al. 1998; McClelland et al. 2012). Therefore, we also examine MU coherence and partial coherence estimates.

Coherence and partial-coherence estimates. For each record three coherence estimates were calculated, these were the ordinary MU coherence, and two partial MU coherence estimates, one with raw EMG as predictor, the other with rectified EMG as predictor. The predictor that most effectively represents the common drive will have the smallest MU partial coherence. We consider data for the unloaded case first and look at coherence and partial coherence estimates pooled across single subjects and across all subjects. Table 1 summarizes the residual correlation data for each subject, it includes the 11 subjects where more than 1 record was obtained.

A number of observations can be made for the data in Table 1. The first is the variability across subjects. Two of the subjects (*subjects 1* and *11*) have partial coherence estimates with raw EMG as predictor that were larger than the ordinary pairwise coherence estimates. The reason for this unusual result is unclear (but see discussion below); however, it does suggest that the raw EMG is a poor predictor of the frequency content present in the MU correlation for these two subjects. For 10 of the 11 subjects, the residual partial correlation with rectified EMG is numerically smaller than that with rectified EMG as predictor. A worst case scenario view of the data in Table 1 is that the raw EMG will not predict any components of MU correlation (residual correlation of 100%; *subjects 1*, *6*,

Table 1. *Residual partial correlation for unloaded data by subject*

Subject	Records	Raw, %	Rect, %
1	7	>100	16
2	12	83	4
3	16	13	19
4	17	0.57	0.11
5	5	68	47
6	14	76	28
7	7	80	34
8	6	100	36
9	10	82	13
10	5	9	3
11	5	>100	29

Table shows subject ID, number of records, percentage residual partial correlation with raw EMG as predictor, and percentage residual partial correlation with rectified EMG as predictor. Residual partial correlation is expressed as a percentage of the ordinary correlation in the beta-frequency band and was calculated using a multitaper spectral analysis with an effective bandwidth of ~16 Hz (see text for details). A value of 100% indicates that the EMG does not predict any components of motor unit correlation. *Subjects 1* and *11* with an indicated residual correlation of >100% for the raw EMG had partial coherence estimates that were larger than the original ordinary pairwise coherence estimates.

and *11*) and the rectified EMG will predict up to half of the total correlation in the beta band (residual correlation of 47%; *subject 5*). A more balanced view is given below where pooled partial coherence across subjects is considered. A second level statistical analysis applying a Mann-Whitney *U*-test to the two sets of residual correlation values provides evidence in favor of a significant reduction across this subject group with rectified EMG as predictor compared with raw EMG ($P = 0.022$).

To provide a more comprehensive summary across the complete unloaded data set, we estimated pooled coherence and pooled partial coherence over all records from all 13 subjects in the unloaded condition. The results are plotted over the frequency range 10–50 Hz in Fig. 2. The ordinary pooled coherence estimate exhibits clear correlation over the 10–50 Hz range, peaking at 24 Hz (Halliday et al. 1999, their Fig. 3*A*). The horizontal dashed line is the upper 95% confidence limit for all estimates, values below this line are consistent with the hypothesis of zero coherence and zero partial coherence at each frequency.

For both unrectified and rectified predictors there is a marked reduction in magnitude of the partial coherence compared with the ordinary MU coherence estimate. The residual partial correlation with the raw EMG as predictor is 22% of that in the original pooled coherence. Using the rectified EMG as predictor the residual correlation is 3.5%. The difference of pooled coherence test applied to data pooled over all subjects using the multitaper approach outlined above indicates this difference is significant over the beta-frequency range 15–32 Hz ($P = 0.0005$). Like the second level analysis above, this suggests that for this subject group, the rectified EMG is a significantly better predictor of MU correlation in the beta-frequency range than the raw EMG for unloaded finger extension against gravity.

It has recently been suggested that contraction strength can affect prediction of MU correlation using surface EMG (Farina et al. 2013). To consider this, we examine the effects of loading. We perform the same analysis for the 84 records in the loaded condition with small weights attached to the distal

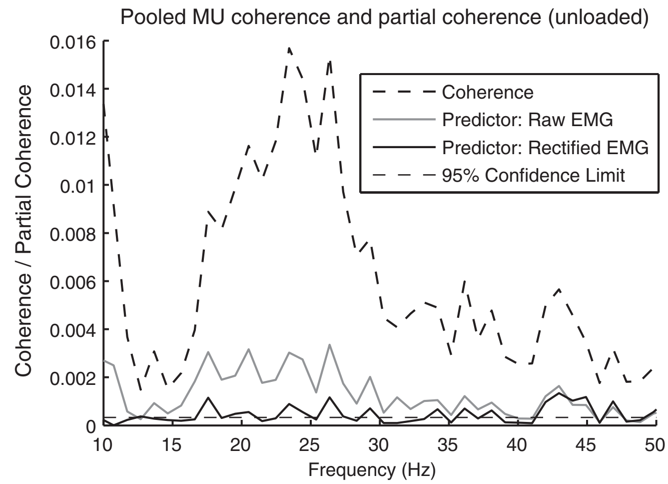

Fig. 2. Estimated pooled coherence and pooled partial coherence between paired MU recordings. Estimates are pooled over 106 records from 13 subjects in the unloaded condition. Ordinary coherence is shown as the dashed line, the partial coherence with unrectified EMG as predictor is the solid grey line, and the partial coherence with the rectified EMG as predictor is the solid black line. Ordinary coherence is identical to that shown in Halliday et al. (1999, their Fig. 3*A*). Dashed horizontal line is the 95% significance level for coherence and partial coherence estimates, based on a NULL hypothesis of zero coherence or zero partial coherence. Estimates are plotted over the frequency range 10–50 Hz. Residual partial correlation with the raw EMG as predictor is 22% of the original MU correlation, with the rectified EMG as predictor the residual correlation is 3.5%. The 2 partial coherence estimates are significantly different over the beta-frequency range (15–32 Hz; $P = 0.0005$; see text for details).

phalanx of the extended middle finger. The results are given in Table 2, broken down by applied load.

The first four entries in Table 2 give a breakdown by load value, and the last two lines summarize the results for the pooled analysis over all unloaded and over all loaded data. Considering the breakdown by load, there appears to be less difference in the residual correlation values in the loaded case (Table 2; first 4 entries) compared with the unloaded case (Table 1). This is supported by a second level comparison, which suggests there is no significant difference between the residual correlation values for rectified and raw EMG with load (Mann-Whitney *U*-test applied to first 4 entries; $P = 0.2$). The last two entries in Table 2 summarize the complete unloaded and complete loaded populations of data; these also include the root mean square (RMS) rectified EMG value in the beta-

Table 2. *Residual partial correlation for all data by load*

Load	Records	Subjects	Raw, %	Rect, %	RMS
5	30	6	18	5	—
10	26	7	5	0.07	—
15	13	6	6.4	6.6	—
>15	15	4	0.9	2.9	—
0	106	13	22	3.5	1.00
5–40	84	13	5.2	1.4	1.17

Table shows load applied to distal end of middle finger (in grams), number of records with this load, number of subjects, percentage residual partial correlation with raw EMG as predictor, percentage residual partial correlation with rectified EMG as predictor, and (for complete unloaded and loaded populations only) root mean square (RMS) value of rectified EMG normalized to the unloaded population. Residual partial correlation is expressed as a percentage of the ordinary correlation in the beta-frequency band and was calculated using the same approach as Table 1. The last 2 rows are for data pooled over the complete unloaded (load 0) and loaded (load 5–40 g) populations.

frequency band. A pooled analysis was repeated for the single set of loaded data using all 84 records with additional inertial loading. The pooled coherence and partial coherence estimates for this combined loaded condition are shown in Fig. 3.

Like the unloaded condition (Fig. 2), there is a marked reduction in the magnitude of the partial coherence compared with the ordinary MU coherence estimate. The residual partial correlation with the raw EMG as predictor is 5.2% of that for the original pooled coherence, for the rectified EMG as predictor the residual correlation is 1.4%. Unlike the unloaded condition, this difference is not significant ($P = 0.19$). As an indicator of the increased MU recruitment resulting from inertial loading, the RMS value of the rectified EMG can be used. Across all records in the loaded condition, the RMS EMG is 17% larger than the same measure across all records in the unloaded condition (Table 2). The RMS EMG allows an assessment to be made of the extent to which amplitude cancellation might be a contributing factor in determining the efficiency of the rectified EMG to predict MU coherence (Farina et al. 2013), see below.

In summary, for the loaded and unloaded conditions, the rectified surface EMG gives a numerically smaller residual partial correlation in the beta-band; however, the improvement in prediction over the raw EMG is not as great as in the unloaded condition.

DISCUSSION

Evidence regarding the appropriateness or otherwise of rectification up to this point has primarily been in the form of simulation studies (Boonstra and Breakspear 2012; Myers et al. 2003; Neto and Christou 2010; Stegeman et al. 2010; Farina et al. 2013). In Stegeman et al. (2010), rectification was found to reduce coherence in a simulated motor pool between the input drive and surface EMG. This study, however, neglects the fact that a single muscle has a range of action potentials shapes, due to the exponential distribution of MU sizes (Fuglevand et al. 1992), instead modeling small and large muscles with identical short and long MU action potential (MUAP) shapes. Boonstra and Breakspear (2012) addressed this shortcoming, showing with modelled EMG that when a heterogeneous distribution of MUAP magnitudes is present, the common oscillatory drive is cancelled out but is recoverable by EMG rectification. Neto and Christou (2010) reconstructed EMG with the same spectral profile as recorded data, but with varying signal-to-noise ratio, concluding that EMG-EMG coherence is impaired by rectification. However, no MU timing information was included in their EMG model, which was based on a sum of sinusoidal signal model. The main aim of EMG spectral analysis is the extraction of information regarding firing patterns in MU activity and the detection of common drive to pools of MUs; for which purpose, it has been suggested that rectification is more appropriate (Boonstra 2010; Halliday and Farmer 2010).

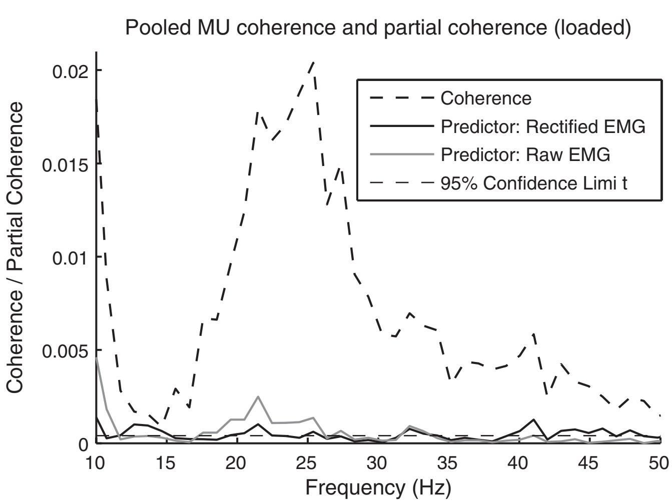

Fig. 3. Estimated pooled coherence and pooled partial coherence between paired MU recordings. Estimates are pooled over 84 records from 13 subjects in the loaded condition. Ordinary coherence is shown as the dashed line, the partial coherence with unrectified EMG as predictor is the solid grey line, and the partial coherence with the rectified EMG as predictor is the solid black line. Ordinary coherence is identical to that shown in Halliday et al. (1999, their Fig. 3*B*). Dashed horizontal line is the 95% significance level for coherence and partial coherence estimates, based on a NULL hypothesis of zero coherence or zero partial coherence. Estimates are plotted over the frequency range 10–50 Hz. Residual partial correlation with the raw EMG as predictor is 5.2% of the original MU correlation, with the rectified EMG as predictor the residual correlation is 1.4%. Reduction in partial coherence with rectified EMG as predictor compared with partial coherence with raw EMG as predictor is not significantly different ($P = 0.19$; see text for details).

Two of the individual subjects exhibited pooled partial coherence estimates with raw EMG that were larger than the ordinary MU coherence estimates in the unloaded case (Table 1, *subjects 1* and *11*). There are a number of factors that could contribute to such an unusual result. A key factor is likely to be the variability in the strength of MU correlation between subjects (see e.g., Halliday et al. 1999, Fig. 2). The residual correlation values in Table 1 do not take account of the magnitude of the MU coherence. In contrast, pooled coherence across subjects (Fig. 2) constructs a weighted sum, where records are weighted inversely proportional to the variance of individual records (Amjad et al. 1997). This provides an estimate of the consistency of any effects across subjects that takes account of the variations in the strength of MU coherence across subjects. This may also in part explain why the residual correlation pooled across subjects (Table 2, *row 5*) are lower than those for individual subjects (Table 1). We interpret the pooled coherence and pooled partial coherence estimates (Figs. 2 and 3) as a more reliable view of the present data. However, in any study it is important to be aware of intersubject variability in parameters of interest (Halliday et al. 1999; Farina et al. 2013).

Recently, it has been suggested that the extent to which raw and rectified EMG reflects the spectrum of common input to motoneurones depends on MU action potential amplitude cancellation (Farina et al. 2013), which increases with increasing MU recruitment. Farina et al. (2013) further suggest that EMG rectification is preferable at low contraction strengths. To get an indication of contraction strength, the RMS value of the rectified EMG can be used. For our data, this is 17% larger for the loaded condition compared with the unloaded condition. This is approximately one-third of the increase seen in the simulated example in Farina et al. (2013, their Fig. 1), which suggests that the contraction strengths in our data are at modest levels and likely to involve low levels of amplitude cancellation. Similarly, there is only a 5% increase in the single MU firing rate in the loaded population over the unloaded population.

Experimental evidence relating to rectification has primarily looked at EEG-EMG or MEG-EMG coherence (Yao et al.

2007; McClelland et al. 2012). Most recently, McClelland et al. (2012) provokingly stated that rectification is an "unnecessary and inappropriate step in the calculation of corticomuscular coherence." This was based on analysis of power spectra from individual records of unrectified and rectified EMG and coherence between a simultaneously recorded EEG. This approach, however, does not provide a firm indication of whether rectified or unrectified EMG more faithfully represents the presynaptic motoneuronal drive. Additionally, the failure to detect significant EEG-rectified EMG coherence at a known common frequency component, the 50-Hz mains artifact, is not as the authors suggest evidence for the inappropriateness of rectification, but in keeping with the interpretation that rectification emphasizes the modulatory drive to the motor pool at the expense of action potential shape and other unwanted high frequency information (Halliday and Farmer 2010).

The key question of whether unrectified or rectified EMG is a better predictor of the frequency components of MU synchronization can only be definitively answered using experimental data consisting of simultaneous recordings of paired MU discharges in which there is synchronization due to common drive and surface EMG from the homonymous muscle. The present study analyses such a data set, and the key result is shown for pooled data in Fig. 2. Use of unrectified EMG as a predictor signal does not remove all components of MU synchronization in the 15- to 32-Hz frequency band. In contrast, the rectified EMG has a partial coherence estimate that fluctuates around the significance level. It is worth noting that while using unrectified EMG as predictor does greatly reduce the magnitude of the partial coherence, in the context of the present debate as to which is the better predictor of MU synchronization, the evidence in Fig. 2 is strongly in favor of the rectified surface EMG. The results verify the improved coherence with rectified EMG observed in previous studies (Boonstra and Breakspear 2012; Myers et al. 2003). Interestingly, the advantage of rectified EMG over raw EMG decreased with increasing inertial loading. Importantly, at the population level our data suggests that raw EMG is unlikely to be a better predictor of MU correlation than rectified EMG (Figs. 2 and 3). The results of partial coherence analysis with increased inertial loading, in which MU recruitment is increased compared with the unloaded condition, is supportive of the suggestion in Farina et al. (2013) that EMG rectification is advantageous at low contraction levels. We note that use of rectification has also been found advantageous in suppressing movement artifacts in vibration studies (Sebika et al. 2013).

A number of previous studies have pointed out that rectification is a nonlinear operation and when used unwisely can distort the frequency components in a signal (Neto and Christou 2010; McClelland et al. 2012). This is true for sinusoidal signals; however, the issue is less clear cut with respect to EMG. As previously discussed in Halliday and Farmer (2010), the main aim of this type of spectral analysis is to characterize the components of MU timing. Taking the single unit spectra as the reference spectra, clear peaks are present around 12–13 and 26 Hz, with Poisson behavior (no significant rhythmic features; see Fig. 1) >35 Hz. The rectified EMG spectrum has peaks around 12 and 25 Hz, which are the main features. Similarly, the unrectified EMG spectrum has evidence of a distinct feature around 12 Hz and possibly around 25 Hz, although this may be part of the broad increase in power that extends above 60 Hz. The maximum power in the unrectified surface EMG is in the region of 40–60 Hz, corresponding to the frequency range where the MU spectrum has no significant rhythmic structure. It has been suggested that this higher frequency range may reflect information due to the shape of MUAPs (Halliday et al. 1995). The alteration of MU action potential shape with fatiguing contractions is known to alter the spectrum of the raw EMG, these changes are often used as a means of quantifying fatigue levels (Hagg 1992). In the present study we are interested in preprocessing approaches that attempt to suppress shape information while maintaining information regarding timing. The conclusion from the experimental data in Fig. 1 is that the rectified EMG spectral estimate is a closer match to the MU spectral estimate than that of the unrectified surface EMG. This is in agreement with the findings of Elble and Randall (1976) who compared MU and rectified EMG spectra, with a focus on the 8- to 12-Hz frequency band.

In conclusion, analysis of a large data set consisting of paired single MUs recordings and surface EMG from the homonymous muscle has demonstrated: *1*) that interpretation of the spectrum of the rectified surface EMG is closer to that of the MU spectrum than that of the unrectified EMG, *2*) that rectified EMG is a better predictor of the components of MU synchronization than the corresponding unrectified EMG at low contraction strengths, and *3*) that increasing contraction strength due to inertial loading lessens but does not reverse the advantage of rectification. These findings support the preprocessing approach of EMG rectification before spectral analysis of data obtained at low to moderate contraction strengths adopted by Elble and Randall (1976) Halliday et al. (1995), and many others.

GRANTS

S. F. Farmer acknowledges funding support from National Institute of Health Research University College London Hospitals Biomedical Research Centre.

DISCLOSURES

No conflicts of interest, financial or otherwise, are declared by the author(s).

AUTHOR CONTRIBUTIONS

Author contributions: D.M.H. and S.F.F. conception and design of research; D.M.H. performed experiments; D.M.H. and N.J.W. analyzed data; D.M.H., N.J.W., S.F.F., and L.B interpreted results of experiments; D.M.H. prepared figures; D.M.H., N.J.W., and S.F.F. drafted manuscript; D.M.H., N.J.W., S.F.F., and L.B edited and revised manuscript; D.M.H., S.F.F., and L.B approved final version of manuscript.

REFERENCES

Amjad AM, Halliday DM, Rosenberg JR, Conway BA. An extended difference of coherence test for comparing and combining several independent coherence estimates: theory and application to the study of motor units and physiological tremor. *J Neurosci Methods* 73: 69–79, 1997.

Bartlett MS. The spectral analysis of point processes. *J Roy Stat Soc B Methods* 25: 264–296, 1963.

Boonstra TW, Breakspear M. Neural mechanisms of intermuscular coherence: implications for the rectification of surface electromyography. *J Neurophysiol* 107: 796–807, 2012.

Boonstra TW. The nature of periodic input to the muscles. *J Neurophysiol* 104: 576, 2010.

Datta AK, Stephens JA. Synchronization of motor unit activity during voluntary contraction in man. *J Physiol* 422: 397–419, 1990.

Elble RJ, Randall JE. Motor-unit activity responsible for 8-to 12-Hz component of human physiological finger tremor. *J Neurophysiol* 39: 370–383, 1976.

Farina D, Merletti R, Enoka RM. The extraction of neural strategies from the surface EMG. *J Appl Physiol* 96: 1486–1495, 2004.

Farina D, Negro F, Jiang N. Identification of common synaptic inputs to motor neurons from the rectified electromyogram. *J Physiol* 591: 2403–18, 2013.

Farmer SF, Bremner FD, Halliday DM, Rosenberg JR, Stephens JA. The frequency content of common synaptic inputs to motoneurones studied during voluntary isometric contraction in man. *J Physiol* 470: 127–155, 1993.

Farmer SF, Gibbs J, Halliday DM, Harrison LM, James LM, Mayston MJ, Stephens JA. Changes in EMG coherence between long and short thumb abductor muscles during human development. *J Physiol* 579: 389–402, 2007.

Fuglevand AJ, Winter DA, Patla AE, Stashuk D. Detection of motor unit action potentials with surface electrodes: influence of electrode size and spacing. *Biol Cybern* 67: 143–153, 1992.

Hagg GM. Interpretation of EMG spectral alterations and alteration indexes at sustained contraction. *J Appl Physiol* 73: 1211–1217, 1992.

Halliday DM, Conway BA, Farmer SF, Rosenberg JR. Load-independent contributions from motor-unit synchronization to human physiological tremor. *J Neurophysiol* 82: 664–675, 1999.

Halliday DM, Farmer SF. On the need for rectification of surface EMG. *J Neurophysiol* 103: 3547, 2010.

Halliday DM, Rosenberg JR, Amjad AM, Breeze P, Conway BA, Farmer SF. A framework for the analysis of mixed time series/point process data-theory and application to the study of physiological tremor, single motor unit discharges and electromyograms. *Pro Biophys Mol Biol* 64: 237–278, 1995.

McClelland VM, Cvetkovic Z, Mills KR. Rectification of the EMG is an unnecessary and inappropriate step in the calculation of Corticomuscular coherence. *J Neurosci Methods* 205: 190, 2012.

Myers LJ, Lowery M, O'malley M, Vaughan CL, Heneghan C, St Clair Gibson A, Harley YX, Sreenivasan R. Rectification and nonlinear preprocessing of EMG signals for cortico-muscular analysis. *J Neurosci Methods* 124: 157–165, 2003.

Neto OP, Christou EA. Rectification of the EMG signal impairs the identification of oscillatory input to the muscle. *J Neurophysiol* 103: 1093–1103, 2010.

Percival DB, Walden AT. *Spectral Analysis for Physical Applications: Multitaper and Conventional Univariate Techniques*. Cambridge, UK: Cambridge Univ. Press, 1983, p. 583.

Priestley MB. *Spectral Analysis and Time Series*. New York: Academic, 1981, p. 890.

Rosenberg JR, Amjad AM, Breeze P, Brillinger DR, Halliday DM. The Fourier approach to the identification of functional coupling between neuronal spike trains. *Pro Biophys Mol Biol* 53: 1–31, 1989.

Rosenberg JR, Halliday DM, Breeze P, Conway BA. Identification of patterns of neuronal connectivity-partial spectra, partial coherence, and neuronal interactions. *J Neurosci Methods* 83: 57–72, 1998.

Salenius S, Hari R. Synchronous cortical oscillatory activity during motor action. *Curr Opin Neurobiol* 13: 678–684, 2003.

Sebika O, Karacanb I, Cidemb M, Türker KS. Rectification of SEMG as a tool to demonstrate synchronous motor unit activity during vibration. *J Electromyogr Kinesiol* 23: 275–284, 2013.

Stegeman DF, van de Ven WJM, van Elswijk GA, Oostenveld R, Kleine BU. The alpha-motoneuron pool as transmitter of rhythmicities in cortical motor drive. *Clin Neurophysiol* 121: 1633–1642, 2010.

Yao B, Salenius S, Yue GH, Brown RW, Liu JZ. Effects of surface EMG rectification on power and coherence analyses: an EEG and MEG study. *J Neurosci Methods* 159: 215–223, 2007.