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Rectification is required to extract oscillatory envelope modulation from surface electromyographic signals

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Dakin CJ, Dalton BH, Luu BL, Blouin JS. Rectification is required to extract oscillatory envelope modulation from surface electromyographic signals. *J Neurophysiol* 112: 1685–1691, 2014. First published July 2, 2014; doi:10.1152/jn.00296.2014.—Rectification of surface electromyographic (EMG) recordings prior to their correlation with other signals is a widely used form of preprocessing. Recently this practice has come into question, elevating the subject of EMG rectification to a topic of much debate. Proponents for rectifying suggest it accentuates the EMG spike timing information, whereas opponents indicate it is unnecessary and its nonlinear distortion of data is potentially destructive. Here we examine the necessity of rectification on the extraction of muscle responses, but for the first time using a known oscillatory input to the muscle in the form of electrical vestibular stimulation. Participants were exposed to sinusoidal vestibular stimuli while surface and intramuscular EMG were recorded from the left medial gastrocnemius. We compared the unrectified and rectified surface EMG to single motor units to determine which method best identified stimulus-EMG coherence and phase at the single-motor unit level. Surface EMG modulation at the stimulus frequency was obvious in the unrectified surface EMG. However, this modulation was not identified by the fast Fourier transform, and therefore stimulus coherence with the unrectified EMG signal failed to capture this covariance. Both the rectified surface EMG and single motor units displayed significant coherence over the entire stimulus bandwidth (1–20 Hz). Furthermore, the stimulus-phase relationship for the rectified EMG and motor units shared a moderate correlation ($r = 0.56$). These data indicate that rectification of surface EMG is a necessary step to extract EMG envelope modulation due to motor unit entrainment to a known stimulus.

rectification; oscillatory input; vestibular stimuli; motor unit; EMG preprocessing

NEURAL OSCILLATIONS are hallmark features of our functioning brain and are thought to subserve functions such as encoding new information or the maintenance of a sensorimotor or cognitive state (see for review Engel and Fries 2010; Klimesch 1999). In the sensorimotor system, oscillatory brain activity, captured with electroencephalography or magnetoencephalography, is correlated with ongoing muscle activity over a variety of bandwidths (see, e.g., Gross et al. 2000; Halliday et al. 2000; Halliday and Rosenberg 1999). Typically, electromyogram (EMG) signals are full-wave rectified prior to their correlation with other signals, and the reasoning for this preprocessing step remains unclear (Boonstra 2010). Similarly, coupling between electrical vestibular stimuli applied to the mastoids and lower limb muscle activity during standing balance (Dakin et al.

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2007, 2010, 2011) or locomotion (Blouin et al. 2012; Dakin et al. 2013) can be identified with correlation measures in the time and frequency domains. Vestibulo-muscular coherence was observed over a bandwidth of 0–20 Hz in lower limb muscles, but the results were also dependent on full-wave rectification of the surface EMG signal. Recently, EMG rectification has come into question, stimulating great interest and debate (Boonstra 2010; Farina et al. 2013; Halliday and Farmer 2010; McClelland et al. 2014). Proponents for rectifying EMG to identify neural modulation of muscle activity suggest that the preprocessing step accentuates EMG spike timing information (Halliday and Farmer 2010; Myers et al. 2003; Yao et al. 2007), whereas opponents argue it is an unnecessary and potentially inappropriate preprocessing step because of its nonlinear distortion of the unprocessed data (McClelland et al. 2012, 2014; Neto and Christou 2010).

Farina et al. (2013) justified the rectification of EMG signals for cortico-muscular or muscle-muscle coherence in contexts where EMG amplitude cancellation is minimal, i.e., during low-level isometric contractions. Despite both modeling and experimental support for EMG rectification under specific conditions, there are still concerns regarding this preprocessing step (Farina et al. 2013; McClelland et al. 2014). A major limitation with experimental estimates of coupling within or between electroencephalographic, magnetoencephalographic, or EMG recordings is that the inputs/outputs are unspecified and consequently cannot be controlled to explore factors influencing the observed cortico-muscular correlations. Here we characterized the influence of EMG rectification on estimates of neural oscillations by exposing subjects to defined sinusoidal electrical vestibular inputs (1–20 Hz) known to linearly induce phase-locked modulation in the surface EMG signal (Dakin et al. 2011). We compared the coherence and phase relationships between the unrectified and rectified EMG with those obtained from indwelling single-motor unit (MU) recordings from the same subjects. Single MUs were used to quantify motor neuron behavior because of the one-to-one relationship between the motor neuron and MU action potentials (Liddell and Sherrington 1925). Also, individual MU spikes are converted to a binary code prior to correlation with the input signal. Consequently, signal rectification is unnecessary (only the timing information remains) and the resulting phase relationship is related closely to the input stimulus-motor neuron phase relationship. The vestibular stimulus-single MU phase relationship will therefore provide an estimate of motor neuron modulation to a controlled vestibular input. This will allow comparison of the vestibular stimulus-single MU correlation

estimates with those obtained from the unrectified and rectified surface EMG. We hypothesized that if rectification is necessary to extract stimulus-induced low-frequency modulation in the motor neuron pool, correlation (i.e., coherence) between vestibular stimuli and unrectified surface EMG will be absent. In addition, we expected that motor neuron modulation to the vestibular input estimated from the rectified surface EMG will produce coherence and phase estimates that are similar to those estimated from single MUs.

METHODS

This study was collected over two separate experiments. The first experiment examined whether or not rectifying surface EMG is a necessary preprocessing step in identifying correlations between a controlled input (vestibular stimuli) and the output (medial gastrocnemius EMG). The data from *experiment 1* were adapted from Dakin et al. (2011) but reanalyzed here to provide data encompassing the entire effective bandwidth of the electrical vestibular stimulus (0–20 Hz). The second experiment was collected at only four stimulus frequencies spanning the vestibular stimulus's bandwidth. The latter experiment aimed to validate the correlations observed in *experiment 1* by comparing coherence and phase estimates between the stimulus and surface EMG to the coherence and phase estimates obtained from single MUs (representing motor neuron behavior). Although oscillatory muscle activity may correlate with the input stimulus (with or without EMG rectification), it is unclear that any correlation identified with surface EMG faithfully represents the oscillatory characteristics of the motor neuron pool. Therefore, single-MU recordings were taken as the gold standard measurement of motor neuron oscillatory activity, independent of any preprocessing technique.

Participants

Five healthy participants (4 men, 1 women) between the ages of 20 and 32 yr participated in the first experiment, and four healthy men (2 from the first experiment) between the ages of 22 and 37 yr participated in the second. All participants had no known neurological disorders or injuries. The procedures were explained verbally and in writing, with the subjects giving informed written consent prior to testing. All procedures conformed to the Declaration of Helsinki and were approved by the University of British Columbia's Clinical Research Ethics Board.

Vestibular Stimuli

Carbon rubber electrodes (9 cm^2), set up binaurally (anode right/cathode left; anode left/cathode right), were coated with Spectra 360 electrode gel (Parker Laboratories, Fairfield, NJ) and secured over the mastoid processes with an elastic headband (*experiment 1*) or Duropore tape (3M, St. Paul, MN) (*experiment 2*).

In *experiment 1*, vestibular stimuli were generated with LabVIEW software (National Instruments, Austin, TX) on a personal computer and sent via a multifunction data acquisition board (PXI-6289, National Instruments) to a constant-current isolation unit (model 2200 Analog Stimulus Isolator, A-M Systems, Carlsborg, WA). Sinusoidal vestibular stimuli were delivered at integer frequencies from 1 to 20 Hz, in a single 90-s trial (1 frequency per trial) and with peak amplitudes of $\pm 2\text{ mA}$.

During *experiment 2*, electrical vestibular stimuli were generated on a PC with Spike2 software (version 6.10, Cambridge Electronic Design, Cambridge, UK) and sent through a multifunction data acquisition board (Micro 1401 mkII, Cambridge Electronic Design) to an isolated constant-current biphasic nerve stimulator (DS5, Digitimer, Welwyn Garden City, UK). Sinusoidal vestibular stimuli were delivered at four frequencies (2, 8, 14, and 20 Hz), with a peak

amplitude of $\pm 4\text{ mA}$. Stimulation at each frequency lasted 120 s. On the basis of pilot testing we chose to provide the sinusoidal stimuli at these settings to ensure responses in single MUs for all frequencies.

Electromyography

Experiment 1. Surface EMG was collected from the left medial gastrocnemius (33-mm interelectrode distance), amplified ($\times 5,000$ – $20,000$; Grass P511, Grass-Telefactor, West Warwick, RI) and band-pass filtered (30–1,000 Hz) prior to being digitized along with the sinusoidal vestibular stimuli at 8,192 Hz. Data were sampled to allow the fast Fourier transform (FFT) to be conducted on windows of a power of 2 (8,192). This allowed the estimation of EMG power at integer frequencies in correspondence with the vestibular stimuli (also delivered at integer frequencies). A common ground electrode was positioned over the left lateral malleolus. Surface EMG was filtered off-line in MATLAB (MathWorks, Natick, MA) with a 5th-order zero-phase shift Butterworth high-pass filter at 30 Hz to remove any residual stimulus artifacts. This filtering should not affect the spike timing information because the majority of the frequency content contained within the surface EMG signal is greater than 20 Hz (Kwatny et al. 1970) and spike timing information occurs at a higher frequency than the filtered frequencies (Potvin and Brown 2004; Riley et al. 2008).

Experiment 2. Surface and intramuscular EMG were collected from the left medial gastrocnemius with a surface electrode placement similar to that in *experiment 1*. Surface EMG signals were band-pass filtered (50–1,000 Hz), preamplified ($\times 100$), amplified ($\times 2$; NeuroLog, Digitimer), and sampled at 2,049 Hz along with the sinusoidal vestibular stimuli. Similar to *experiment 1*, the surface EMG signals were also filtered off-line in MATLAB with a 5th-order zero-phase shift Butterworth high-pass filter at 30 Hz. Intramuscular EMGs were recorded from the medial gastrocnemius with custom-made insulated fine-wire electrodes wound together (Stablohm 800A, California Fine Wire, Grover Beach, CA). The tips of the recording electrodes were hooked and cut at the recording end to expose only the cross section of the wires (0.05 mm). No extra insulation was removed from the wires in order to record distinct MU action potentials. While the participant was seated comfortably, three pairs of hooked-tip fine wires were inserted via 25-gauge hypodermic needles (Becton-Dickinson, Franklin Lakes, NJ) into the muscle belly. After insertion the needles were withdrawn, leaving the electrode pairs embedded in the medial gastrocnemius. Multiple fine-wire electrode pairs were used to maximize the yield of MU action potential trains recorded through the entire experimental session (4 different stimulation frequencies). A common ground electrode was positioned over the left lateral malleolus. Intramuscular EMG signals were band-pass filtered (50–15,000 Hz), preamplified ($\times 100$ – $1,000$), amplified ($\times 2$ – 5), and sampled at 16,393 Hz. All EMG and electrical stimulation signals were digitized with a 16-bit analog-to-digital converter (Micro 1401 mkII, Cambridge Electronic Design) and stored to a PC.

Testing Protocol

Each participant stood with his or her arms relaxed at the sides and feet positioned with the medial malleoli ~ 2 – 3 cm apart. The head was turned to the right, and gaze was maintained on a target to maintain Reid's plane parallel to the floor. By maintaining this head position, the postural response to the vestibular stimuli was aligned primarily with the anterior-posterior directions (Lund and Broberg 1983), along the line of action of the plantar flexors (we recorded from the left medial gastrocnemius). To maintain this position throughout the protocol, the placement of the feet was marked and participants were asked to focus on a target to their right. In the first experiment, participants were provided 20 sinusoidal stimuli spanning 1–20 Hz, with each frequency provided as a single 90-s trial. In the second experiment, participants were exposed to twelve 120-s trials of sinu-

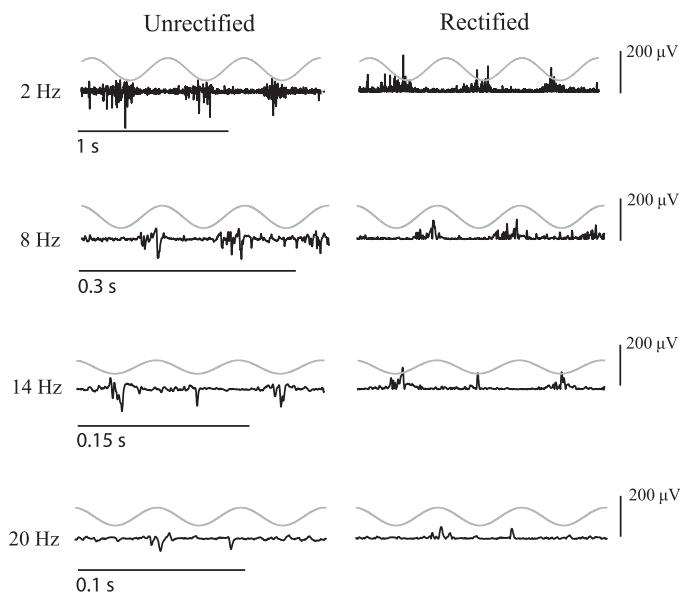


Fig. 1. Unrectified and rectified surface EMG from a representative subject for 4 different stimulation frequencies corresponding to the frequencies used during the single-motor unit (MU) recordings. Each trace displays the modulation in EMG amplitude with the stimulus at each frequency and the effect of rectification. Gray sinusoids are the change in stimulus current causing EMG modulation. Note that the timescale (*x*-axis) decreases with increasing frequency.

soidal vestibular stimulation, which consisted of three trials at the four stimulation frequencies (2, 8, 14, and 20 Hz) delivered in a random order. Rest periods were given after every four trials to avoid fatigue.

Data Analysis

Experiment 1. First, power spectra were calculated for both unrectified and rectified EMG to determine whether muscle oscillations at the stimulation frequencies are captured by the FFT. Second, coherence and phase functions were estimated between the input (vestibular stimulus) and the unrectified and rectified EMG signals to determine whether or not rectification is necessary to identify the coupling between the vestibular input stimulus and EMG response. Coherence and phase were calculated with a MATLAB script based on the methods described by Rosenberg and colleagues (Halliday et al. 1995; Rosenberg et al. 1989). Windows of 8,192 data points were used to provide a frequency resolution of 1 Hz. By using windows of this size, coherence estimates occur at integer frequencies corresponding to the stimuli delivered at integer frequencies. Coherence was deemed significant in each subject if it was greater than a 0.95 confidence limit (0.09) based on the number of disjoint segments (90 segments) used in estimation. If rectification is necessary to identify coupling between the stimulus and lower limb EMG, then correlation between the vestibular stimuli and unrectified EMG will not result in significant coherence.

Experiment 2. To determine whether or not the unrectified and rectified surface EMG can provide information regarding motor neuron modulation by the input vestibular stimulus, a total of eight single MUs (2 per subject) were discriminated. The MU recordings were analyzed with a template matching algorithm (Spike2 version 6.10, Cambridge Electronic Design). Discrete MU action potentials were identified by comparing the MU waveform shape with respect to the temporal and spatial characteristics of the template. An experienced investigator manually inspected all MU action potentials to determine whether an action potential belonged to a specific MU based on the wire electrode pair from which the MU was recorded, the action potential shape, and differences in spike timing and MU firing rate.

MU spike times were used to estimate coherence and phase functions between the sinusoidal vestibular input stimuli and MU action potential trains. Coherence and phase estimates were also calculated on the unrectified and rectified surface EMG from the same participants to determine whether vestibular-evoked modulations of motor neurons estimated with unrectified or rectified EMG signals reflect the neural oscillations occurring at the single-MU level. Coherence and phase estimates were derived in a manner similar to the methods of *experiment 1* (window size: 2,048 points). These data were also analyzed to provide a frequency resolution of ~1 Hz corresponding to the integer frequencies of the sinusoidal vestibular stimuli. Coherence and phase estimates were computed for all trials and averaged over the three trials for each stimulation frequency.

For statistical purposes, vestibulo-muscular coherences estimated with the two preprocessing methods for the surface EMG (unrectified and rectified) and single-MU recordings were compared with a two-way analysis of variance (ANOVA) (EMG type \times stimulation frequency) after application of a Fisher *z*-transformation (Rosenberg et al. 1989). Because stimulus-EMG coherence estimates were not significant for the unrectified surface EMG at any frequency tested, the phase estimates obtained from the unrectified EMG signals were not interpretable (Halliday and Rosenberg 1999) and were excluded from the two-way ANOVA and further statistical analyses (see RESULTS). A two-way ANOVA (EMG type \times stimulation frequency) was also used to compare the estimated phase values for the rectified surface EMG and single MUs. When a significant main effect or interaction occurred, a Tukey's honestly significant difference (HSD) post hoc analysis was performed. The α level was set at $P \leq 0.05$. A Pearson correlation coefficient (r) was computed to assess the relationship between the phase estimated with the rectified surface EMG and single MUs.

RESULTS

At low frequencies (<20 Hz), vestibular stimuli induced visible modulations in EMG activity (Fig. 1). The strength of these oscillations varied across trials and participants and was modulated closely by the vestibular stimuli. In all subjects, the FFT of the unrectified EMG did not show any peaks at the stimulus frequency across the range of frequencies used in this study. After rectification of the surface EMG, clear power at each stimulus frequency was observed (Fig. 2).

Since surface EMG data prominently covaried with the vestibular stimulus (with a fixed time lag), it is expected that

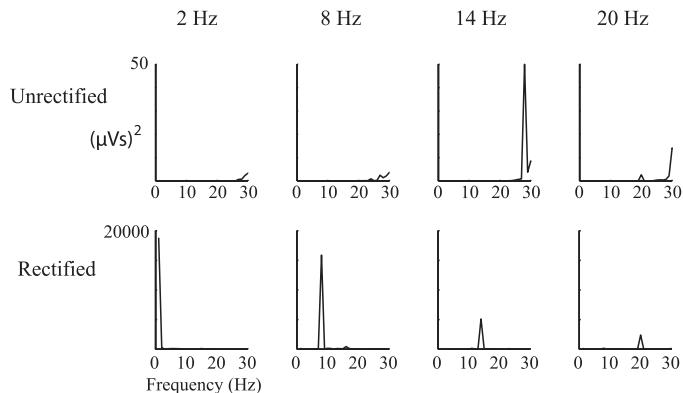


Fig. 2. Power spectra for the surface EMG data displayed in Fig. 1 from 1 representative subject. Top: modulation in the amplitude of unrectified EMG at the 4 stimulus frequencies is not captured by the fast Fourier transform (FFT). Rectification prior to FFT provides a representation of EMG envelope modulation. The power spectrum of the unrectified EMG for the 14-Hz input vestibular stimulus revealed power at a harmonic (28 Hz) of the input frequency. Note the expanded *y*-axis scale for the power spectra of the unrectified EMG (top) compared with those of the rectified EMG (bottom).

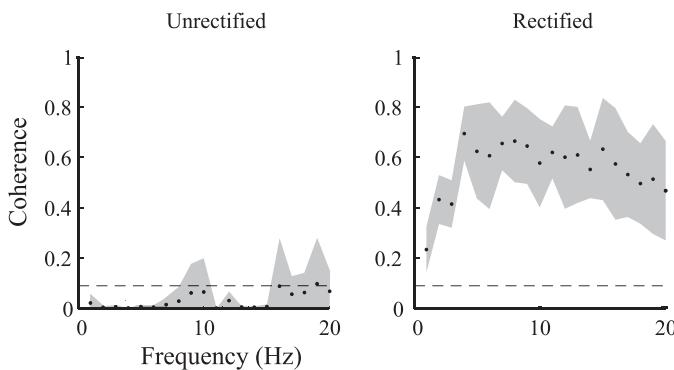


Fig. 3. Peak coherence between the stimulus and the unrectified and rectified EMG across all subjects ($n = 5$). Stimulus-EMG coherence was not significant with the unrectified data, whereas the rectified EMG data exhibited significant coherence with vestibular stimuli presented at all frequencies (1–20 Hz). Black dots indicate mean peak coherence; gray shading is SD of coherence amplitude. Dashed line indicates 95% confidence limit.

stimulus and EMG signals should correlate at the stimulus frequency. Again, only the rectified EMG signals demonstrated a significant correlation with the vestibular input signals at the specific input frequencies tested (Fig. 3).

Single-Motor Unit Activity (Experiment 2)

Single-MU firing rates were modulated by all frequencies of vestibular stimulation (Fig. 4). At higher stimulation frequencies (14 and 20 Hz), single MUs from the medial gastrocnemius did not fire at every oscillation of the vestibular input, but stimulus-EMG coherence was significant for all frequencies (Fig. 5A). Similar to experiment 1, stimulus-EMG coherence was observed at all frequencies for the rectified surface EMG but not for the unrectified EMG (Fig. 5A). Since coherence between the stimulus and unrectified surface EMG was not significant, its phase was not interpretable (Halliday and Rosenberg 1999). Thus the coherence amplitude and phase estimates for the unrectified EMG were not compared statistically with the single MUs and are excluded from Fig. 5B.

For coherence amplitude, there was a main effect for stimulation frequency ($F_{3,30} = 15.91, P < 0.01$) and EMG type ($F_{1,10} = 32.02, P < 0.01$) with an interaction ($F_{3,30} = 3.518, P < 0.05$). The largest coherence for both stimulus-rectified EMG and stimulus-single MU data was observed for the 8-Hz stimuli. For the 8-Hz vestibular stimulus 35% of the variability in MU firing rate covaried with the vestibular input, but this value decreased for the other stimulation frequencies tested, ranging from 10% to 25% (Fig. 5A). Furthermore, the lowest stimulus-single MU coherence was observed at the highest frequency (20 Hz), corresponding to the decrease in coherence at 20 Hz observed in surface EMG. Coherence amplitude was 38–78% greater for the rectified surface EMG than the single MUs at all corresponding frequencies. For the phase estimates there was a significant main effect for stimulation frequency ($F_{3,30} = 11.04, P < 0.01$) with no effect for EMG type ($F_{1,10} = 0.046, P = 0.84$) or interaction ($F_{3,30} = 0.85, P = 0.48$). The phase between the input vestibular stimuli and EMG responses (rectified and single MUs) decreased as the stimulation frequency increased, and this relationship was not statistically different between the EMG recording types (Fig. 5B). Furthermore, a moderate correlation was present between stimulus-MU and stimulus-rectified EMG estimated phase ($r = 0.56, P < 0.01$).

Thus, at the stimulation frequencies examined, the estimated phase relationship between the vestibular stimulus and modulation in MU activity was captured by the rectified surface EMG (Fig. 5B).

DISCUSSION

Sinusoidal vestibular stimulation was used to modulate lower limb motor neuron activity at a series of defined frequencies ranging from 1 to 20 Hz. With the controlled input stimuli, we investigated whether rectification is a necessary preprocessing step in order to extract low-frequency oscillatory signals from surface EMG in lower limb motor neurons. Indeed, rectification is required, as the stimulus did not cohere with the unrectified EMG data for the frequencies tested. Additionally, we found that the phase relationship estimated with rectified EMG was correlated with that obtained with single MUs. Overall, these results indicate that rectification is a necessary step in identifying muscular oscillations in multi-unit surface EMG recordings.

Oscillatory input onto the motor neuron pool, depending on its strength, can modulate motor neuron output at the specific input frequency. Weak oscillatory input will cause small shifts in the timing of the action potentials, while strong oscillatory input may result in complete phase locking of motor neuron firing pattern to the stimulus. The single-MU data presented here indicate that, depending on the stimulus frequency, there

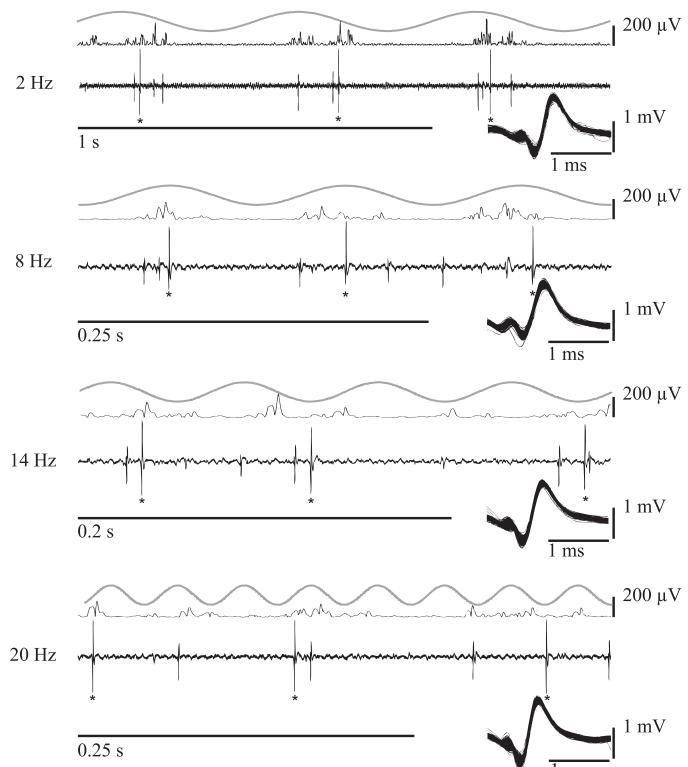


Fig. 4. Rectified EMG and single-MU data for 4 different stimulation frequencies (experiment 2). For each frequency top trace represents the change in stimulus current amplitude (gray sinusoids) causing EMG modulation, middle trace depicts the rectified surface EMG, and bottom trace represents unprocessed single-MU action potentials. Asterisks indicate MU action potentials included in the MU overlay at bottom right. The same MU was decomposed and overlaid for all frequencies. Timescales were chosen for illustrative purposes.

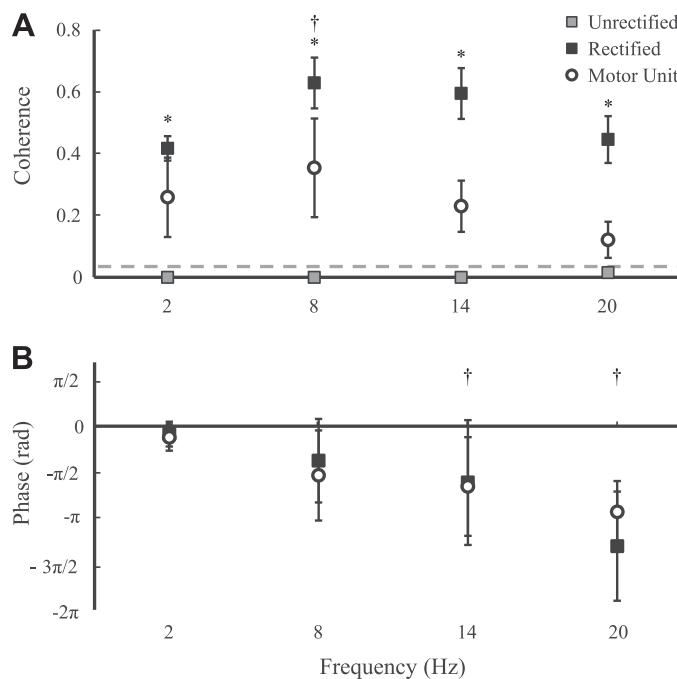


Fig. 5. *A:* coherence amplitude at each stimulation frequency for the different EMG preprocessing methods and single MUs. Single MUs cohered at the stimulus frequency over the frequency range tested in a pattern similar to the coherence estimated with the rectified surface EMG data. Stimulus-unrectified EMG coherence remained near zero at all frequencies. Coherence amplitude was highest at 8 Hz for both the rectified EMG and single MUs (${}^{\dagger}P < 0.05$), and coherence amplitude was greater for the rectified surface EMG than for the single MUs ($*P < 0.05$). Data are presented as means \pm SDs. Dashed line represents 95% confidence interval. *B:* phase-frequency functions are presented for single MUs and the rectified EMG data only because coherence between the stimulus and unrectified surface EMG was not significant, and therefore its phase was uninterpretable. The rectified EMG exhibited a phase-frequency relationship that was not statistically different from that of single MUs ($P = 0.84$) in that the estimated phase decreased as stimulation frequency increased for both rectified EMG and the single MUs. \dagger Phase is statistically different from the value estimated for the 2-Hz stimulus ($P < 0.05$). Data are presented as means \pm SDs.

is a weak to moderate entrainment of motor neurons to sinusoidal vestibular electrical stimulation. For lower stimulation frequencies (2 and 8 Hz) 25–35% of the variability in MU firing rate covaries with the vestibular input, but this value decreases to 10–20% for the 14- and 20-Hz stimuli (Fig. 5A). This covariation is visually obvious in the raw data as single MUs fired at every oscillation of the vestibular input for the 2- and 8-Hz stimuli but skipped a number of oscillations at the higher frequencies (Fig. 4). One of the factors contributing to this observation is the low frequency of MU firing rate in medial gastrocnemius during standing balance (4–14 Hz; Héroux et al. 2014). Ultimately, spike timing information is crucial to assess the oscillatory influence of an input upon the motor neuron. The techniques used here were well suited to identifying how a known input influences motor neurons because we controlled the input (sinusoidal electrical vestibular stimulus at a defined frequency) and recorded motor neuron output indirectly via discrete single-MU action potential recordings. The one-to-one relationship between motor neuron and muscle fiber action potentials simplifies this inference (Liddell and Sherrington 1925).

Most experiments, however, are not performed under these ideal conditions. Here we replicated the major experimental

challenge of identifying motor neuron oscillations to a known vestibular input with surface EMG. Surface EMG recordings are the most common way to assess muscular oscillations but do not provide a direct assessment of motor neuron spike timing because of a global recording area and the superposition of multiple MU action potentials. Multiunit EMG preprocessing techniques such as rectification seek to enhance spike timing information to provide an indirect estimate of motor neuron spike timing. Our results show that preprocessing the surface EMG signal via rectification is necessary to characterize muscular oscillations induced by an electrical vestibular stimulus. Two important features of the oscillations estimated from the rectified surface EMG must be discussed. First, the strength of the observed vestibular input-motor neuron coupling is increased when using rectified surface EMG. This is likely because surface EMG signals reflect the contribution from a summation of multiple active MUs under the recording electrodes. Consequently, it is expected that the estimated coherence would be larger for multiunit recordings (Negro and Farina 2011). Second, the phase relationship with the input stimulus estimated from rectified surface EMG is correlated with that computed from single MUs ($r = 0.56$). One possible explanation for this moderate correlation stems from the larger variability in phase estimation for input stimuli at 14 and 20 Hz. Coherences for these stimuli were lower, paralleling similar observations with surface EMG (Dakin et al. 2007) and suggesting weaker oscillatory influence on the motor neuron pool. Because of this weaker coherence, phase estimates from single MUs and rectified surface EMG at these higher frequencies must be interpreted cautiously.

The recording of single MUs also includes waveform information that can obscure timing information in certain contexts. During low-level contractions where superposition of action potential spikes is minimal, waveform information is useful to identify and sort single MUs. Once MU classification is complete, this waveform information is typically discarded and only the timing information retained (Rieke et al. 1999; Wood et al. 2004). Oscillations in muscle activity estimated with this MU identification technique are comparable to those estimated with rectified surface EMG. Separating the timing and waveform information contained within the surface EMG is quite difficult, and much of the current debate is focused on whether rectifying the EMG signal obscures this critical timing information. Our data suggest strongly that rectification retains the important motor neuron spike timing information and should be adopted for future studies quantifying vestibular-induced muscular oscillations in similar contexts. In the present data, the separation of timing and waveform information in the surface EMG may be simplified by the relatively strong phase locking and low-level muscle activity. These factors result in a prominent modulation in EMG amplitude at the given stimulus frequency. The envelope of the EMG activity therefore provides timing information about the motor neuron pool while reducing the influence of the waveform shape upon subsequent analysis. Although rectification distorts the waveform data nonlinearly, it also acts as a crude enveloping procedure facilitating the identification of low-frequency EMG oscillatory activity with a FFT. Correspondingly, the results from using other standard enveloping procedures, such as the instantaneous amplitude of the Hilbert transform (Myers et al. 2003), are very similar to rectification (Dakin et al. 2012). Impor-

tantly, in our study rectification maintained the MU timing information as shown by the similar coherence spectra and input-output phase relationships estimated with rectified surface EMG and single MUs.

Should Rectification Be Used as a General Principle?

Whether or not surface EMG data need to be rectified seems to be a matter of context. Recently, Farina et al. (2013) demonstrated for low-level contractions of intrinsic hand muscles that rectification of the surface EMG may enhance timing information. However, as the muscle activity amplitude increases with a concomitant rise in amplitude cancellation of MU action potentials, the benefit of multiunit EMG rectification is lost and may even be detrimental to identifying oscillatory input to the muscle. In the present study, participants were asked to stand freely, which generally requires low levels of muscle activity in the triceps surae (Héroux et al. 2014; Joseph and Nightingale 1952; Luu 2010). This low-level contraction, combined with low-frequency input stimuli (below 20 Hz), is likely well suited for EMG rectification because of the separation between independent MU spikes (or clusters of spikes) leading to prominent EMG amplitude modulation (see surface EMG from Figs. 1 and 4). At higher contraction intensities rectification may be beneficial, but only if EMG amplitude modulation at the stimulus frequency is still prominent. Rectification, however, may be limited when trying to correlate weak or higher-frequency oscillatory inputs with multiunit EMG. In this context, the subtle timing between spikes becomes important and EMG envelope modulation at the frequency of oscillation may be less apparent or absent. Although not specifically addressed here, indirect evidence supports EMG rectification for the detection of muscular oscillations elicited by a vestibular stimulus at frequencies up to 75 Hz (Forbes et al. 2013). However, the usefulness of rectification in the contexts of higher-intensity contractions or higher stimulus frequencies requires further investigation. Finally, in specific contexts where the shape of the EMG waveform is important, as in studies examining action potential waveform changes with fatigue, the nonlinear distortion of waveform information may be undesirable and therefore should be avoided. Additionally, in certain frequency and time-frequency analysis, EMG rectification may not be recommended.

Filtering Considerations

An important consideration regarding the methods of this study is the role of surface EMG filtering. Band-pass filters of the EMG signal between 20 and 500 Hz are commonly used before digitally sampling the data. This preprocessing step is performed to remove low-frequency artifacts from cable motion or electrical signals from the heart (Clancy et al. 2002; Redfern et al. 1993) and high-frequency noise from the EMG signal (Clancy et al. 2002). Without these filters, electrical signals from sources other than electrical muscle activity can contaminate the EMG data. Since the timing of muscle action potentials is crucial, low-frequency signals may interfere with the estimation of input-output correlations. Indeed, high-pass filtering may further enhance spike timing information in surface EMG signals (Potvin and Brown 2004; Riley et al. 2008). To remove these potentially confounding signals, we

digitally high-pass filtered the surface EMG signals at 30 Hz, using a 5th-order zero-phase shift Butterworth filter. The high-pass filter did not influence the correlation measures obtained with the rectified surface EMG but did remove correlations with the unrectified EMG data. This was despite the obvious firings of multiunits at the vestibular stimulus frequency observed in the filtered unrectified EMG (Fig. 1). Here we propose that the absence of correlation in the unrectified EMG is due to the limited ability of the FFT to represent the low-frequency envelope modulation present in the unrectified EMG signals.

Conclusions

Much debate exists over whether or not multiunit EMG data should be rectified prior to further analysis, particularly to identify muscular oscillations. Depending on the type of muscle contraction, rectification may be beneficial or detrimental. Here we demonstrated that surface EMG rectification is necessary in order to identify motor neuron oscillations in responses to a defined vestibular electrical stimulus during standing balance. We conclude that EMG rectification is a necessary step in the detection of coupling between vestibular stimuli and ongoing muscle activity.

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DISCLOSURES

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

AUTHOR CONTRIBUTIONS

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

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