Joint Stiffness Fluctuations Are Potentially Associated with Postural Sway

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Abstract

Joint stiffness can be induced by simultaneous contraction of muscles on opposite sides of the joint. When the electrical oscillations of these antagonistic muscles are in-phase, fluctuations in joint stiffness can occur, which can pose a threat to those with a neuromuscular deficit, as they will not be able to adequately control these fluctuations. In an attempt to develop rehabilitation programs to train these individuals on how to control these fluctuations, electrical activity of the muscles in the lower leg, and postural sway were measured. It was hypothesized that the amplitude coherence of low-frequency, in-phase oscillations of antagonistic muscles was positively correlated with postural sway. Data collection occurred while subjects were standing in the tandem stance, a position of induced coactivation of the antagonistic muscles of the lower leg. Electrical activity of the right tibialis anterior, medial gastrocnemius, and soleus were measured and analyzed against center-of-pressure power, a measure of postural sway. EMG measurements, such as amplitude coherence and power spectral density, were only analyzed at low-frequencies (0-3 Hz) at which the oscillations were in-phase. Data analysis yielded no significant correlations between amplitude coherence and center-of-pressure power, and no significant correlations between power spectral density and center-of-pressure power. Future modifications to this study can incorporate visual feedback in order for subjects to be able to modulate their muscle activity to hit a specific target of co-activation, and COP sway will be measured simultaneously.

Introduction

Neuromuscular physiology is highly focused on those individuals with neural deficits that experience a loss of muscular control, and whether this malfunction occurs within the bounds of the neural system or the muscular system. Because the two bodily systems interact heavily, it is difficult to know where to focus rehabilitation techniques in order to help these individuals regain motor control. A large problem for victims of neuromuscular impairment is their difficulty in maintain postural control, as their ability to activate the leg muscles has been compromised. In healthy individuals, this activation is accomplished by intended contraction (low-frequency in-phase oscillations between agonist and antagonistic muscles, referred to as "common drive"). Because muscles on either side of the bone are being contracted in-phase, a certain level of joint stiffness is created that maintains joint and postural stability. Healthy individuals are mostly able to control any imbalances that may occur through unequal gravitation forces and subsequent reciprocal activation of the antagonist muscles of the leg, contributing to a push-pull mechanism and sway in the anterior posterior direction. However, in positions in which mechanical imbalances are particularly prevalent, the in-phase neural oscillations may contribute to sway due to the phasic nature of the co-contraction. This phasic activation of the antagonist muscles can cause fluctuations in joint stiffness, which can contribute to the sway already caused by mechanical imbalances.

However, in those individuals with neurological impairment, there is greater effort expended for normal levels of spinal cord activation (10), which may perturb the system greater and cause more fluctuations in joint stiffness. In addition, those who cannot maintain muscular control will have more difficulty maintaining balance should there be an unexpected decrease in joint stiffness and subsequent decrease in joint and postural stability. Because these possibilities can pose a dangerous threat to the lives of those with neuromuscular impairment, my study will examine the possible positive correlation between joint stiffness and sway. Previous researchers have examined the level of EMG activation of the muscles of healthy individuals under neuromuscular fatigue (which represents a model similar to neuromuscular impairment), and have found that the magnitude and phase of the common drive components greatly affect stability of the muscles (10). However, my study will induce co-activation of the muscles, and therefore joint stiffness, in healthy individuals, in order to examine sway in a position of high mechanical imbalance. Under these conditions, the co-activation of antagonistic muscles necessary to maintain balance should induce a high level of common drive, and more phasic oscillations. I am hypothesizing that the coherence of the EMG signals of antagonistic muscles will be positively correlated with postural sway, measured by center of pressure velocity. If this hypothesis is supported, it is suggested that the similar shapes of these low-frequency oscillations contribute to postural instability. This information can be used to find techniques, such as biofeedback, to consciously control or inhibit the in-phase component, especially in a setting such as a neuromuscular rehabilitation

Literature Review

Mechanical imbalances in body posture can occur in positions in which it is necessary to maintain balance on a narrow base, such as the tandem position. These forces can cause extremely slow mechanically-driven oscillations (< 0.5 Hz) that can occur due to mechanical imbalances in body posture. The mechanical oscillations can cause reciprocal activation of the antagonistic muscles to maintain balance, and can contribute to a push-pull mechanism of the tibialis anterior and soleus. Neural oscillations (0.5-3 Hz) originating in supraspinal locations have previously been referred to as “common drive” because the phasic signals from the brain are sent to an entire muscle group to maintain tonicity, including all of the agonist and antagonist muscles. This phenomenon occurs in the upper limbs as well (1). These “common drive signals” have always been known as those necessary in order to maintain muscular control, because contraction is occurring between antagonistic muscles on either side of a bone. However, it has been found that, in positions of high co-activation between the antagonistic muscles, common drive signals may actually contribute to higher degrees of unsteadiness. It is speculated that this is due to the fact that co-activation of antagonistic muscles creates high joint stiffness, and therefore, any fluctuations in joint stiffness would result in the limb swaying back and forth (1) because the speculated sway caused by joint stiffness fluctuations can contribute to the sway caused by mechanical imbalances. If common drive has the ability to be “controlled” by the brain consciously, then this could help to decrease the postural sway caused in positions of high co-activation of the muscles (1). Therefore, the current study is examining the correlation between parameters of EMG activity of antagonistic lower leg muscles, and postural sway.

It has been known that victims of neuromuscular disorders, such as stroke victims, must exert greater effort for spinal cord activation, which may greater perturb the system of joint stiffness. In addition, the fluctuations in joint stiffness can be a major problem if there were to be a sudden decrease in joint stiffness and tonicity at the instant the patient bears weight on one leg during walking. One of the speculated interventions in order to find if these oscillations can be controlled includes bouts of co-contraction and individual contraction practice; this entails actively practicing contraction of antagonistic muscles simultaneously and individually in the hopes of training the neuromuscular system to be under more conscious control (1). It was found that bouts of practice did not improve one's ability to perform steady co-contraction well; however, these practice periods took place over the span of an hour, a relatively short period of time (1). In addition, these experiments were performed with antagonistic muscles in the arm, instead of the leg. It has been suggested that practicing co-contraction through exercise can have benefits for stroke victims, due to the ability to increase cognitive control and to reduce joint stiffness (2).

By being able to analyze the characteristics of the low-frequency oscillations that constitute "common drive", one will be able to statistically analyze the characteristics of the signals that cause sway in positions of high co-activation of the antagonist pairs of the lower leg. These pairs include the tibialis anterior, the medial head of the gastrocnemius, and the soleus (4). There have been studies in which parameters of the EMG signals between these muscles are analyzed during quiet standing (7), and during sitting (4); however, the waves have not been analyzed while the subject is trying to maintain a position of high co-activation of the antagonistic leg muscles. Therefore, in the current study, the subject should stand in a position of high co-activation, the tandem position in which the toe of the right foot is touching the heel of the left foot, and try to maintain this position for 30 seconds.

Balance is quantified through force measurements recorded by a force plate and analyzed as center-of-pressure parameters such as center-of-pressure velocity and rms distance from the mean (3, 5, 6). Studies conducted measuring center-of-pressure deviations, and therefore postural unsteadiness, in male athletes concluded that postural control improved after practice of steady co-contraction (6). Even though this is a non-clinical application of the hypothesis, perhaps it can be applied to the necessity of stroke victims to practice co-contraction in order to control low-frequency oscillations. These center-of-pressure deviation measurements can be measured in both the anterior-posterior and medial-lateral directions (3). Because the set of antagonistic muscles being studied are on the anterior and posterior of the leg, it is speculated that these will contribute to anterior-posterior sway, not necessarily medial-lateral sway.

In addition, postural variability as measured by center-of-pressure deviations has been extensively studied as a function of aging (5, 7). In a study examining groups of young and elderly people in both eyes-open and eyes-closed conditions found that there were significant differences in both the center of pressure velocity between the two age groups, and between the two eye conditions (5). In another study measuring the same muscle activity and postural sway, a higher level of postural sway is seen during time periods of activation of the tibialis-anterior, and therefore during time periods of co-contraction, but only in the elderly (7). These findings could be due to the elderly losing neuromuscular control with aging, and having high levels of joint stiffness overall, which would lead to difficulties performing actions involving joint movement. This high joint stiffness, also increased in positions of co-activation of antagonist muscles would further decrease the ability of the elderly to control movements (7). This conclusion implying causation is purely speculative and is not supported by the results of the study, as the study is purely correlational (7). The current study, while not excluding any age groups, perhaps should attempt to increase the subject population to include a large age range, in order to eliminate any potential bias in the results. The mentioned previous study (5) includes a detailed explanation of the data analysis techniques used for center-of-pressure calculations, so this has been used as a model for the data analysis of the current study.

EMG characteristics that are being analyzed in the current study include amplitude coherence, phase coherence, and cross-correlation between 2 muscles in a pair, and power spectral density of the waveforms of the individual muscles (1). Each of these parameters represent different characteristics of the similarities between waveforms of different muscle pairs during standing in tandem position for 30 seconds. It was found that cross-correlation between the biceps brachii and triceps brachii was increased during co-contraction (10). However, it is not known whether this increase in cross-correlation also correlated with an increase in instability of the limb.

Muscle pairs being analyzed in the current study are the tibialis anterior-medial gastrocnemius (antagonistic pair), the tibialis anterior-soleus (antagonistic pair), and the soleus-medial gastrocnemius (synergist pair). In a study in which force plate measurements were taken during sitting subjects attempting to adjust their position on the force plate to move their center-of-pressure to hit a target, visual feedback of their own center-of-pressure in relation to the target was shown during the training period. The visual feedback increased the accuracy, timing, and smoothness of movements in the training period, but not in the post-training period (4). Because this study has found that visual feedback information improves control during the training period (4), the current study may be able to find the same effects, but in reaction to visual feedback of electromyography signals. Visual feedback shown would include the percentage of maximal volume contraction shown by electromyography signals. If the subject were able to see and aim to hit a certain level of co-contraction, it may influence their ability to maintain specific levels of activation of each muscle in order to balance in the tandem position with the least amount of postural sway.

Because the target audience for which rehabilitation techniques should be improved based on the results of these studies is the stroke victim population, current devices and techniques used by stroke victims should also be considered. One of these devices is an ankle-foot orthotic (AFO), which induces ankle stiffness (9). This is due to the fact that in this situation, there is no fluctuation in joint stiffness; therefore, there is no decrease in joint stiffness. After EMG waveforms correlated with sway give more insight into how sway is correlated with specific levels of joint stiffness, future work can include development of a self-adjusting AFO, in which the stiffness provided by the orthotic is adjusted while the patient is walking in order to only compensate for the stiffness that the patient cannot produce themselves.

By determining which aspects of the EMG wave pairs are coherent with each other, and are also positively correlated with postural instability, the current study will be able to speculate about the characteristics of the electrical signal that contribute to sway, and can therefore speculate about improved rehabilitation techniques for stroke victims. Based on the results of the current study, future steps can be taken to introduce bouts of practice that take place over weeks or months in stroke rehabilitation, and the specific nature of this practice will be based on the knowledge gained by the current study. It has been shown that it is possible to increase muscular strength without increasing the activity of low-frequency oscillations (12); therefore, implementing practice in stroke rehabilitation should allow patients to regain muscular control.

Methods and Materials

Experimental Set-Up

There were four subjects that participated in the study, although 2 of the subjects completed the study more than one time. Therefore, a total of 6 sets of data were collected. All subjects were healthy young adults with no history of neurological disorder or neuromuscular deficit. The ages of the subjects ranged from 17-21 years of age; 3 subjects were female and one subject was male. All subjects performed the same experimental protocol. Experimental set-up first included instructing the subjects about the tandem stance in which they would be standing in. The subjects were instructed to place their left foot directly in front of their right foot, with the heel of their left foot touching the toes of their right foot. No practice of the stance was done; the subjects were only instructed how to assume the stance. The tandem stance can be seen in Figure 1 to the right.

Figure 1. Tandem stance.

Electromyography set-up was then completed. Bipolar electrodes were attached to the skin over the belly of the tibialis anterior, medial gastrocnemius, and soleus muscles of the right leg. Prior to the electrode attachment, the skin was prepared by gently exfoliating it and cleaning it with alcohol. Each electrode had an interelectrode difference of 18 mm (Z03 EMG; Motion Lab Systems, Baton Rouge, LA). EMG signals were acquired at a rate of 1.9841 kHz, or about 1,9841 samples per second. An AMTI AccuGait force plate was also part of the experimental set-up. Force and moment data in three dimensions was acquired from the force plate at a rate of 140 samples per second using AMTI's data acquisition software. A trigger system was built by another undergraduate student in order for data collection from the force plate and the EMG electrodes to begin simultaneously and be synced in time.

Experimental Protocol

Prior to beginning each trial, the subject was instructed to stand on the force plate, and the force plate was then tared; this was done so that the subject's weight did not factor into the force plate data. The subject was then instructed to assume the tandem stance as previously instructed. Once the subject was comfortable and steady in the tandem stance, the trigger was flipped to begin both EMG and force plate recording. Data was recorded for 30 seconds, and the trigger was then flipped again to signal the end of EMG recording. Force plate recording was set up to automatically end after 30 seconds. The subject then stepped off the plate while the data was saved in both text files and Matlab files. This marked the completion of one trial. Ten trials were completed for each subject. If the subject were to lose balance and fall during one of the trials, then the trial was re-started.

Data Analysis

All data analysis was done in Matlab. Electromyography data was first filtered through an 8^{th} order Butterworth bandpass filter of 16-512 Hz to eliminate any noise or artifact. The DC component was removed from all EMG data, and full-wave rectification was performed. Initially, parameters of EMG data that were examined included cross-correlation, power spectral density, amplitude coherence, and phase coherence. All EMG analysis that was done was completed between each possible pairing of muscles: tibias anterior – soleus, tibias anterior – medial gastrocnemius, and medial gastrocnemius – soleus. These muscle pairings include two antagonistic muscle pairs, and one synergistic muscle pair. Although the hypothesis of the current study does not aim to examine the in-phase oscillations between the synergistic muscles of the leg (as these do not contribute to joint stiffness), they were still analyzed to confirm reasonable results. At this phase, visual observation was done to examine the properties of the filtered signals, as well as the properties of EMG parameters.

Phase coherence and amplitude coherence were then analyzed more in depth. Phase coherence analyses were run on each muscle pair, and the frequency range over which phase coherence angle returned 0 (in-phase oscillations) was noted, and it was ensured that these values were lower than 3 Hz. Therefore, the condition of low-frequency in-phase oscillations was met. Amplitude coherence values were then averaged over this frequency range of in-phase oscillations, and this value was recorded for each muscle pairing, for each trial.

Center-of-pressure (COP) values were obtained from the force and moment data collected from the force plate. First, all COP values were referenced to the mean, rather than being referenced exactly to the center of the force plate. This was done in order to correct for variability in standing position between subjects. Average displacement for each COP value was calculated, and the root-mean-square error of displacement was calculated. The displacement and trial time values were used to then calculate the average velocity of the COP over the 30-second trials. COP velocity in the time domain was initially used as a measure of postural sway; however, further analysis led us to use RMS power in the frequency domain. By using RMS power, we were able to look for correlation with EMG amplitude coherence, which was also in the frequency domain.

For each muscle pairing, the average amplitude coherence over the in-phase frequency range was plotted against the COP RMS power for that trial. Scatterplots were created for each muscle pair that each contained 10 points, one for each trial. Correlation coefficients were obtained, and the plots were also analyzed visually and qualitatively. Scatterplots were also created in order to plot power spectral density of each muscle under 3 Hz, against the COP RMS power for each trial.

Results

Initial analysis included plotting amplitude coherence over a low-frequency range for which the EMG signals were in-phase, against the COP RMS power for each trial, for each muscle pairing. These scatter plots can be seen in figures 2, 3, and 4 below. The results noted are for one subject. Out of this subject's ten trials, two data points were removed due to abnormally large COP power values. Data from the other subjects was analyzed, but results are not included, due to lack of significant correlation and existence of many outliers.

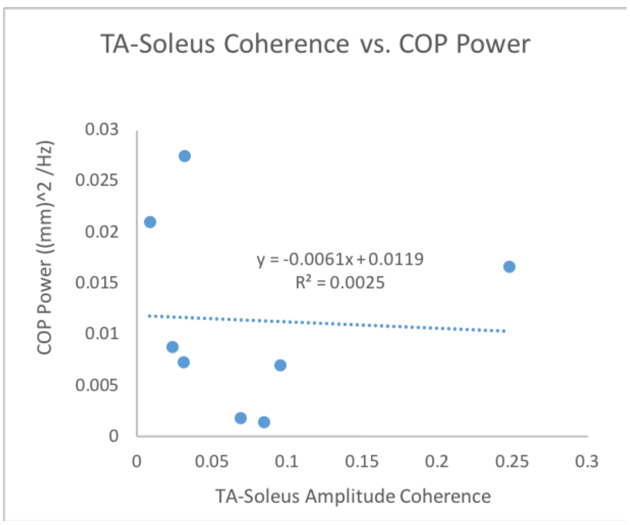

Figure 2. Tibialis anterior-soleus amplitude coherence average vs. center-of-pressure power peaks for Subject #1.

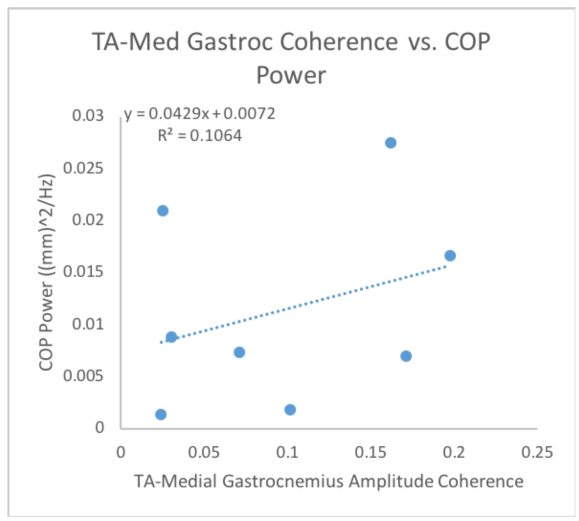

Figure 3. Tibialis anterior-medial gastrocnemius coherence average vs. center-of-pressure power peaks for Subject #1.

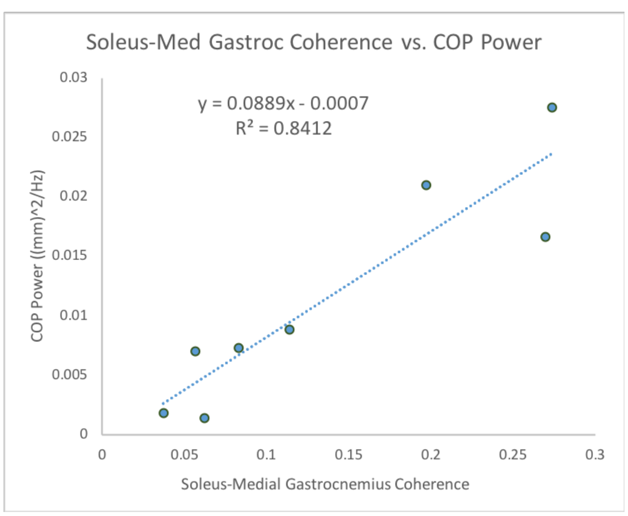

Figure 4. Soleus-medial gastrocnemius coherence average vs. center-of-pressure power peaks

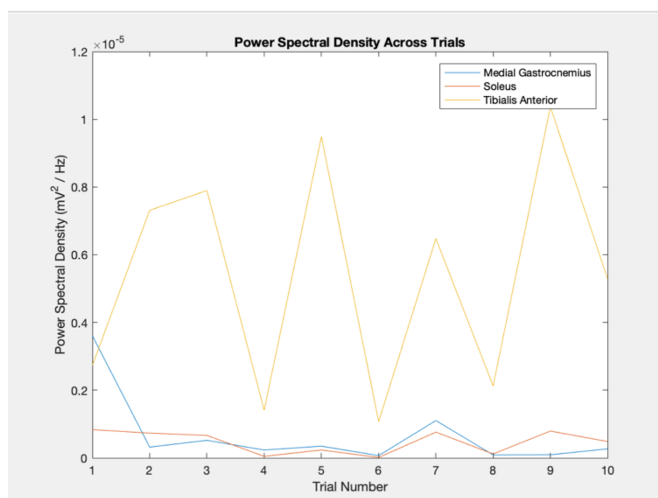

Figure 5. Power spectral density of tibialis anterior, soleus, and medial gastrocnemius across trials

After this analysis was done, we decided to also examine the EMG power spectral density of the individual muscles across the trials, and how this related to the COP power peaks as well. Power spectral density for all 3 muscles as a function of trial number can be seen in Figure 5 above. Multiple regression analysis was also done in Microsoft Excel to analyze the relative contribution EMG coherence and EMG power spectral density had to center-of-pressure power, our quantification of sway. A table of regression results, along with their significance values can be seen in Table 1 below. This data is also from the same subject as figures 2, 3, 4, and 5 above. Data from other subjects was analyzed, but results were not included due to lack of significant data and existence of many outliers.

Input #1 (Coherence)	Input #2 (Power Spectral Density)	Output	R-Square	Significance (p-value)
Tibialis anterior-soleus	Tibialis anterior	COP Power	0.3104	0.4755
Tibialis anterior-soleus	Soleus	COP Power	0.1378	0.7433
Tibialis anterior-medial gastrocnemius	Tibialis anterior	COP Power	0.0316	0.9377
Tibialis anterior-medial gastrocnemius	Medial gastrocnemius	COP Power	0.2113	0.6220
Soleus-medial gastrocnemius	Soleus	COP Power	0.1631	0.6408
Soleus-medial gastrocnemius	Medial gastrocnemius	COP Power	0.3457	0.3463

Table 1. Multiple linear regression analysis of EMG coherence, EMG power spectral density, and center-of-pressure power

Discussion

The correlation values seen in Figures 2 and 3, in which EMG coherence is plotted against COP power, are close to 0. This may indicate the absence of a correlation between coherence and COP power; however, this data is only taken from a small amount of subjects. The correlation value for EMG coherence of the soleus and medial gastrocnemius, plotted against COP power, is 0.8412, which may indicate a positive correlation between these two values. Because the medial gastrocnemius and the soleus are synergist muscles and both contract on the posterior side of the leg, high amplitude coherence between these two muscles at the low-frequencies for which the oscillations are in phase likely indicates a high amount of contraction on the posterior side of the leg. Therefore, this may be contributing to the correlated high degree of sway seen because of mechanical sway, rather than neural sway. Because the oscillations for all three combinations of muscles were in-phase at extremely low frequencies (<1 Hz), we may have been measuring sway caused by mechanical imbalances, as opposed to that caused by common drive. We may have been observing the slow push-pull mechanism that occurs due to mechanical imbalances, and perhaps at these extremely low frequencies, antagonistic muscles were out-of-phase and may have had high amplitude coherence. However, this correlation still shows the importance of in-phase, low-frequency oscillations in contributing to postural sway.

Figure 5 describes the power spectral density values of all three muscles for one subject across all trials. It is evident from this figure that the soleus and medial gastrocnemius are working in synergy, as their power spectral density values go through peaks and troughs at the same time, and the values themselves are similar. The tibialis anterior, on the other hand, has power spectral density values much larger than those of the soleus or medial gastrocnemius. It still has peaks and troughs at the same points as the other two muscles, but on a much larger scale. Because these power spectral density values have such a large range, these might have been the result of errors in data recording. It is not likely that there was such a large fluctuation in values between trials, because each trial was the same task, and should have required similar activity of the tibialis anterior. The electrodes may not have been fully attached to the skin, and may have been accidentally recording movement of the muscle, as opposed to electrical activity within the muscle. However, the general shape of the wave and its similarity to that of the other two muscles recorded shows that the muscles had similar power outputs for each trial. This possibly could have contributed to the postural sway experienced by the subject during each trial.

We then conducted multiple linear regression analysis, treating EMG coherence and EMG power spectral density as two inputs that could be combining and contributing to the postural sway effect. A similar analysis was done in a study examining Alzheimer's Disease using EEG; the researchers observed the effects of PSD and coherence individually, and then the effects of the two together on a physiological measurement (8). Therefore, we thought to apply the same to EMG, as we believed both the similar shapes of the waveforms, as well as the power over these shapes, may be contributing to the fluctuations in joint stiffness, and therefore, to postural sway. The results of the multiple linear regression for one subject can be seen in Table 1 above. All of the R-squared values obtained were less than 0.35, indicating the model not being a good predictor for the data. In addition, all of the p-values obtained were very high, indicating no significant results.

In order to obtain significant results, there could be many changes to both the data collection methods and data analysis methods. First, I believe that the problem being studied requires an interdisciplinary approach; a team consisting of students versed in neuroscience and electrical engineering will be vital to accurately collecting and analyzing the data. Including electrical engineering students would be extremely useful for completing signal processing and analyzing the different parameters of the waveforms. With regards to subject recruitment, an equal ratio of both males and females should be studied. The subjects should ideally be at similar levels of athleticism and should have similar levels of muscular strength, in order to standardize for personal differences between level of muscle activity and contraction. In addition, subjects should be within one age group, perhaps all subjects will be of college age. Prior to beginning data collection, standard measurements of electromyography should be taken while standing flat-footed, standing on toes, and standing on heels. This will be to ensure that the EMG data is within the normal and acceptable range for individual activation of both the anterior and posterior muscles of the leg, prior to inducing coactivation of the muscles for the experimental protocol.

With regards to experimental design, the current protocol will only need to be refined and standardized in order for significant results to be produced. One large problem that we experienced throughout different phases of data collection was standardizing the trial beginning and end time between force plate and EMG systems. A trigger system was built; however, the technology was not reliable, and would often fail to signal the start of a trial for one of the two systems. Therefore, there were multiple times in which the trial had to be restarted to ensure that both systems began recording at the exact same time. We recommend using a similar trigger system, but creating a sturdier solder connection to ensure proper functioning. Extensive testing of the trigger system should be done with the experimental protocol, as if a real experiment were taking place, to make sure that the additional electrical pathways do not interfere with the data being obtained from either the EMG system or the force plate system. In this study, only COP variations in the anterior-posterior plane were measured and analyzed; however, future work should investigate COP variation in then medial-lateral direction as well. Although the muscles being studied contract in the anterior-posterior plane, they may be contributing to sway in the medial-lateral direction (3). In the future, a data collection system that includes visual feedback of EMG data may also help to induce coactivation of the antagonist muscles. While still standing in the the tandem positon, the subject will be able to visualize the level of activation of a specific muscle, as a percentage of its maximal activation. The subject should then try to increase their activation to hit the "target" for each muscle. Maximal activation data will be collected prior to the start of data collection, and visual feedback would be customized for each subject. These measurements could likely be taken during the previously discussed period of EMG recording of individual contraction of each of the muscles, prior to the data collection of the 10 trials.

There are also aspects of the data analysis that can also be improved in future aspects of this study. In order for efficient digital signal processing to be possible, sampling frequency during data collection should be optimized for data analysis. This should allow noise and artifact to be sufficiently removed from the data. In addition, in the current study, we are using phase coherence as a measure of joint stiffness because we do not have a way of directly measuring joint stiffness. A study by Zatsiorsky et al. discussed decomposing the stabilogram into two different components: rambling and trembling. The rambling trajectory describes the migration of the reference point around which body equilibrium in maintained, and the trembling trajectory describes the body's oscillations with respect to the rambling trajectory. They note that the trembling trajectory, which is COP variation relative to rambling trajectory, is a function of restoring force acting on some parameter of mechanical compliance (12). If the out-of-phase EMG oscillations can be paralleled with the restoring force, then it may be possible to parallel the described mechanical compliance with joint stiffness. In this equation, out-of-phase EMG oscillations would be acting on existing joint stiffness in order to affect the trembling component of COP variation. Joint stiffness could therefore be measured indirectly through the changes of out-of-phase oscillations and of COP variation. In addition, if this method were successful, multiple regression analysis could be done to determine the relative contributions of out-of-phase oscillations and joint stiffness (otherwise measured by in-phase oscillations) to postural sway, measured by COP variation.

If these modifications were incorporated into the experimental design, significant results could likely be obtained. If the hypothesis were supported, and it was found that amplitude coherence over in-phase low frequency oscillations was positively correlated with postural sway, this information could be used to develop neuromuscular training programs and rehabilitation systems for victims of neuromuscular deficit. In addition, if further analysis yielded power spectral density as also having a correlation with postural sway, these parameters could also be included in the rehabilitation program. Multiple linear regression would show how much contribution each of these EMG parameters have to postural sway, and this could be used to design a feedback system in which patients practice standing on top of a force platform with electrodes attached to their leg muscles, similar to the experimental design. Patients would then theoretically be able to practice co-activation, but would also be able to practice control of fluctuations in joint stiffness. If this is successful, the potential threat of instability would be a significantly decreased problem for those with neuromuscular deficit.

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