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Analysis of ground reaction force and electromyographic activity of the gastrocnemius muscle during double support

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Abstract

Purpose: Mechanisms associated with energy expenditure during gait have been extensively researched and studied. According to the double-inverted pendulum model energy expenditure is higher during double support, as lower limbs need to work to redirect the centre of mass velocity. This study looks into how the ground reaction force (GRF) of one limb affects the muscle activity required by the medial gastrocnemius (MG) of the contralateral limb during step-to-step transition. *Methods:* Thirty-five subjects were monitored as to the MG electromyographic activity (EMGa) of one limb and the GRF of the contralateral limb during double support. *Results:* After determination of the Pearson correlation coefficient (r), a moderate correlation was observed between the MG EMGa of the dominant leg and the vertical (Fz) and anteroposterior (Fy) components of GRF of the non-dominant leg ($r=0.797$, $p<0.0001$; $r=-0.807$, $p<0.0001$) and a weak and moderate correlation was observed between the MG EMGa of the non-dominant leg and the Fz and Fy of the dominant leg, respectively ($r=0.442$, $p=0.018$; $r=-0.684$ $p<0.0001$). *Conclusions:* The results obtained suggest that during double support, GRF is associated with the EMGa of the contralateral MG and that there is an increased dependence between the GRF of the non-dominant leg and the EMGa of the dominant MG.

***Keywords*:** Double-inverted pendulum model; Double support phase; Ground reaction force; Ankle plantar flexor activity; Electromyography.

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1. Introduction

Several models have been suggested to describe human gait mechanisms [1-7]. Although neural and mechanical systems, including musculoskeletal dynamics, supra-spinal and afferent modulation, work together to minimise energy expenditure [8, 9], it is widely accepted that the recovery of mechanical energy during gait is incomplete, even though muscles work to compensate energy loss [6, 7, 10, 11]. Understanding how individual muscles contribute to specific tasks can provide important insights into neuromuscular control and walking mechanics. This understanding can also aid in developing improved prosthetic, orthotic and other assistive devices to mitigate neuromuscular impairments and designing more effective rehabilitation strategies.

According to the double-inverted pendulum model [7, 12] a major energy loss in walking is due to step-to-step transitions [12], which occur mainly during double support as the two leg forces need to redirect the centre of mass (COM) velocity from a downward and forward direction to an upward and forward direction. The leading leg strikes the ground, performing negative work on the COM and the energy lost may be restored through positive work by the trailing leg. The double-inverted pendulum model predicts that restoring mechanical work is done by the trailing leg at the instant of heel strike through a powerful plantar flexion [6, 13, 14]. Based on these assumptions, a relation between ankle plantar flexors muscle activity and heel strike force would be expected, as these muscles play a major role in propulsion in late stance during unimpaired walking [15, 16] and are important to provide body support [17-19], matching the second peak of the vertical ground reaction force [20]. Considering predictions of the double-inverted pendulum model, impact forces can demonstrate the role of the gastrocnemius muscle: it contributes to push-off and limits heel strike. According to Doets et al., 2009 [21], the larger the heel strike cost in the leading leg during heel strike, the higher the metabolic cost of walking, i.e., the energy dissipated during step-to-step transition explains 29% of the variance in the metabolic energy cost of walking.

Neurophysiologically, the regulation of human walking requires a close coordination of muscle activation between the two legs which seems to be achieved by a flexible neuronal coupling at spinal level (for reviews, see, for example, [22, 23]). During gait, a perturbation of one leg evokes a purposeful bilateral response pattern, with a similar onset latency on both sides, which is thought to be mediated at spinal level under supraspinal control [22, 24-26]. Also, there is evidence of bilateral interlimb coordination in homonymous muscle groups in the human, as each limb affects the strength of muscle activation and the time-space behavior of the other [27].

Many different groups of afferents from flexor and extensor muscles can influence the locomotor pattern. Most attention has focused on the action of group I afferents from ankle extensors, which contributes to 30-60% of ankle extensor activity [28]. Feedback-mediated reinforcement of ankle extensor muscle activity contributes, together with supraspinal drive and possibly spinal drive (i.e., central pattern generator), to propel the body forward. During the stance phase of the human step cycle, the ankle undergoes a natural dorsiflexion that stretches the soleus muscle [28]. In [29], it has been reported that the Hoffman reflex is relatively low at the time of heel contact, increases progressively during the stance phase, and reaches its maximum amplitude in the late stance phase. However, ankle extensor muscle velocity during normal walking is slow [30] suggesting that group Ia afferent pathways may not be effectively recruited during normal walking. In fact, there is a growing body of evidence suggesting that load information provided by group Ib afferents, arising from force-sensitive Golgi tendon organs, contributes substantially to ongoing muscle activity. In addition, it has been suggested that tendon organ feedback via an excitatory group Ib pathway contributes to the late stance enhancement of the soleus muscle activity [31]. In [32], it has been found that electrical stimulation of an extensor nerve at group I strength during walking in the cat spinal substantially increased ongoing activity in other extensor muscles (medial and lateral gastrocnemius).

Taking into account the above and that ground reaction force (GRF) is regarded as a representative measurement of gait, because it is the external force involved in walking and affects the acceleration of the body's centre of mass [33], the main purpose of this study is to investigate how much a relatively simple measure such as the heel strike force can explain plantar flexor muscle activity during propulsion of the contralateral leg. Such knowledge can, for instance, be used to evaluate patients' gait and the efficacy of prosthetic and orthotic devices. The results obtained will also contribute to understand the mechanisms involved in the double support phase of walking, specifically the function of plantar flexors.

2. Methods

2.1 Subjects

Thirty-five healthy female subjects were tested (age = 19.7±1.3 years, height = 1.65±0.045 m, body weight = 56±5.4 kg, non-dominant Q angle = 14.57±0.85 degrees; dominant Q angle = 14.7±0.96 degrees; mean±S.D.). Individuals not matching at least one of the following criteria were excluded: history of recent osteoarticular or musculotendon injury of the lower limb or signs of neurological dysfunction which could affect lower limb motor performance; history of lower limb surgery; lower limb anatomical deformities, Q angle below 14º or above 17º [18]. Biomechanical changes resulting from abnormal alignment may influence joint loads, the mechanical efficiency of muscles, and proprioceptive orientation and feedback from the hip and knee, resulting in altered neuromuscular function and control of lower limbs [19]. All subjects were right-leg dominant.

The study conformed to the ethical norms of the Institutions involved and to the Declaration of Helsinki, dated 1964. Informed consent was obtained from all participants.

2.2 Instrumentation

Vertical (Fz), anteroposterior (Fy) and mediolateral (Fx) GRF values were obtained from a force plate, model FP4060-10, from Bertec Corporation (USA), connected to a Bertec AM 6300 amplifier, with default gains and a 1000 Hz sampling rate. The amplifier was connected to a Biopac 16-bit analogical-digital converter, from BIOPAC Systems, Inc. (USA). The floor and the underlying structure were rigid and flat to minimise any vibrations. Also, the top of the force plate was at the floor level, which was obtained by having a raised walkway. To avoid measurement errors, a gap of 1-2 mm was left between the force plate and the surrounding floor. Reliability of measurements of GRF magnitude has an intraclass correlation coefficient (ICC) of 0.88 [34]. Medial gastrocnemius

(MG) electromyographic activity (EMGa) was monitored using Biopac Systems, Inc – MP 150 Workstation; TD150B steel electrodes were used with bipolar configuration, 20 mm between detection surfaces (centre to centre) and a reference electrode. At preferred walking speeds the total energy of the mean EMG value averaged across the gait cycle has been shown to be highly consistent [35]. Gait timing was measured by the Brower Timing system (IRD-T175, Utah, USA), which presents a sensitivity of 0.01 seconds. For each subject, the time interval measured was used to calculate the mean speed of walking in each trial. Two pressure transducers (TSD111, BIOPAC Systems, Inc.) were used to access gait cycles of the trailing leg. Q angle measurement was performed with a Baseline universal goniometer. Intra-rater reliability for measuring Q-angle in the supine position presents ICC values of 0.94 [36]. Skin impedance was measured with an Electrode Impedance Checker (Noraxon USA, Inc.). Signals obtained from the force plate, the electromyography and the pressure transducer were processed with Acqknowledge, version 3.8, from BIOPAC Systems, Inc.

2.3 Procedures

Subjects were asked to perform two series of three trials of walking at self-selected speed, as three strides of EMG data per subject provide reliable information [35]. The EMGa of the MG muscle of one leg (the trailing leg) and the GRF of the contralateral leg (the leading leg) were collected. In the first series, the EMGa of the dominant limb during propulsion and the GRF of the non-dominant limb at heel strike were monitored. In the second series, we collected the EMG signal of the non-dominant limb during propulsion and the GRF of the dominant limb at heel strike. Measurements were randomised to prevent possible influence from order or learning effects. The lateral gastrocnemius was not measured as it has been documented that EMG patterns of both heads of the gastrocnemius are similar in terms of the timing of activation during walking [20, 37]. In addition, it appears that the MG plays an important role in forward propulsion, whereas the soleus does not [38].

a) Skin and instrument preparation

Skin surface of the subjects' lower limbs was prepared to reduce electrical resistance to less than 5000 Ω [39]: shaving of the MG area; removal of dead skin cells with alcohol; removal of non-conductor elements with abrasive pad [40]. Measurement electrodes were placed at the MG centre, according to [41], and fixed with adhesive tape, to prevent displacement and to guarantee homogeneous and constant pressure. The reference electrode was placed on the patella. Between electrodes positioning and the beginning of measurements we set an interval not lower than 5 minutes [42]. Mean walking speed was verified using a photoelectric timing system, with sensors positioned 0.95 m apart, at floor level, on both sides of the force plate. Pressure transducers were placed on default anatomic locations (calcaneal centre and first metatarsophalangeal joint), which helped to measure the gait cycle time of the leg which had no contact with the plate [43]. All subjects used the same shoe type, in their size.

b) Measurement

All subjects walked along a 10 m walkway, as 10-12 m is the preferable interval to measure gait of young people, since it permits fast walkers to ‘get into their stride’ before any measurements are made [44]. Subjects were instructed to step on a force plate located in the middle of the walkway and to keep walking past the reference point without stopping. In each measurement, only the limb on which the GRF was measured had full contact with the plate and there was no extra load on it. Subjects walked for a minimum of 8 steps [45, 46]. According to the concept of gait optimisation, it is hypothesised that the neuromuscular locomotor system is best stabilised at the usual walking speed, that is to say, gait variability is also minimised at the usual walking speed [47-49]. Therefore, walking speed was freely chosen by each subject. Before the data acquisition session itself, subjects executed several trials to get used to the procedures.

EMGa data were collected by a one-channel unit at 1000 Hz. The signals were pre-amplified at the electrode site and then fed into a differential amplifier with an adjustable gain setting (12 - 500 Hz; CMRR: 95 dB at 60 Hz, input impedance of 100 MΩ and gain of 1000). Raw signals were digitised and stored on computer disks for subsequent analysis by the Acqknowledge software. After measurement, the EMG signal of MG during propulsion was processed and analised. Propulsion was defined as the time between the beginning of weight transfer from the calcaneous to the first metatarsophalangeal joint and the maximum peak of load of the first metatarsophalangeal joint. To guarantee valid results, we have previously taken measurements in each limb both with pressure (foot) switches and with the force plate. Comparing the signals obtained, we have concluded that by positioning pressure switches according to the indicated, we could use the defined time window to assess the EMG activity of the GM during propulsion. The EMG signal of MG during propulsion was filtered digitally with a zero-lag, second-order Butterworth filter with an effective band pass of 20–500 Hz and the root mean square (RMS) was calculated [40]. The signal was also normalised according to maximal voluntary contraction to reduce subject variability and to convert the EMG amplitude to an estimate of muscle activation [50]. Following a warm-up consisting of three submaximal isometric contractions, each subject was instructed to perform one series of three trials of maximal isometric plantar flexion force. Subjects were standing with the hip at 90º, the knee extended and the ankle in neutral position. They were asked to execute maximal isometric force for plantar flexion, under resistance, during 5 seconds, with one-minute rest between trials, being conformed that there was no EMGa [51]. The signals collected within the first and last seconds of each 5 seconds of isometric contraction were not used for analysis because of the possible occurrence of ankle movement at the initiation and completion of the test. Therefore, a 3-second window of EMG signal was used for analysis. This window of raw EMGa was processed using the RMS procedure to assess the electrical activity of the MG muscle.

GRF components (Fx, Fy and Fz) were filtered with a Butterworth filter and normalised according to weight [49, 52]. The maximum value of the heel strike impulse peak in the Fz, Fy and Fx trace were used for analysis. To account for possible effects due to anthropometrics, gait speed was normalized to leg length [53, 54]. To reduce the within-individual variability and increase statistical power, the calculated variables for the three trials for each subject were averaged [55].

2.4 Statistics

Data analysis was performed using the Statistical Package Social Science (SPSS), version 13.0, from SPSS Inc. (USA). Shapiro-Wilk test results and Histogram analysis have shown that data were normally distributed; as a result, we have used parametric statistics. The Paired-Samples T Test was applied to assess possible significant differences between dominant and non-dominant limbs in terms of EMGa during propulsion and GRF during heel strike. The Pearson Correlation Coefficient Test was used to assess the correlation between MG EMGa and GRF and between EMGa/GRF and speed.

3. Results

Figures 1-3 demonstrate the correlation between the EMGa of the non-dominant leg during propulsion and the GRF of the dominant leg during heel strike and between the EMGa of the dominant leg during propulsion and the GRF of the non-dominant leg during heel strike. According to the Pearson Correlation Coefficient Test, there was moderate correlation between the MG EMGa of the dominant leg during propulsion and Fz and Fy of the contralateral leg during heel strike ($r=0.797$, $p<0.0001$; $r=-0.807$, $p<0.0001$) and weak and moderate correlation between the MG EMGa of the non-dominant leg during propulsion and Fz and Fy, respectively, of the dominant leg during heel strike ($r=0.442$, $p=0.018$; $r=-0.684$, $p<0.0001$). These correlations indicate that the amount of variability in MG EMGa of the dominant leg explained by Fz and Fy of the contralateral leg is 63.5% and 65.12%, and that the amount of variability in MG EMGa of the non-dominant leg explained by Fz and Fy of the contralateral leg is 19.5% and 46.8%, respectively. Figure 4 shows how raw EMG signal and GRF profile vary for both limbs. No statistically significant correlations were observed between Fx of the dominant leg and EMGa of the contralateral leg (r=0.189, p=0.276) and between Fx of the non-dominant leg and EMGa of the contralateral leg (r=-0.184, p=0.291).

Although subjects walked at their comfortable speed and values of standard deviation between subjects were low, it was important to analyse the influence of speed variation on EMGa and GRF. The Pearson Correlation Coefficient Test showed a non-significant correlation between speed and EMGa of dominant (r=0.104, p=0.551) and non-dominant limbs (r=-0.187, p=0.282) and between speed and Fz, Fy and Fx of dominant (r=-0.102, p=0.558; r=-0.192, p=0.269; r=-0.284, p=0.099) and non-dominant limbs (r=0.055, p=0.792; r=-0.39, p=0.823; r=-0.017, p=0.923).

Comparing EMGa and GRF values obtained in dominant and non-dominant limbs (Table 1), there is not enough statistical evidence to conclude that there are significant differences in Fz and Fy at heel strike (p=0.18; p=0.358) and MG EMGa during propulsion (p=0.08). Differences were observed between Fx of dominant and non-dominant limbs at heel-strike (p<0.0001).

4. Discussion and Conclusions

The importance of active work during propulsion [56-58] leads to the need of understanding the mechanisms involved in step-to-step transition, and more specifically to assess the influence of the contralateral leg heel strike on the degree of ankle plantar flexors' muscle activity.

Experiments in this study have shown a statistically significant correlation between the MG EMGa of the dominant leg during propulsion and Fz and Fy of the contralateral limb during heel strike and between the MG EMGa of the non-dominant leg and Fz and Fy of the contralateral limb at heel strike. According to the double-inverted pendulum model, the activity of the leading leg in the double support phase can be designated by heel strike, as the force directed along the leg executes negative work. On propulsion of the trailing leg, an equal amount of positive work is performed, arousing the need to restore energy loss in the following heel strike. Transition between steps reaches an optimum level when propulsion and heel strike have the same magnitude and a short duration [7]. Looking at the step-to-step mechanism presented in [6, 7], the results of this study suggest that Fz and Fy are associated to the amount of activity required by the MG of the contralateral limb, which is consistent to the role of plantar flexors during propulsion, as they have been considered important contributors to vertical and horizontal acceleration [16, 17, 59, 60]. This finding corroborates the concept that the power activity of the trailing leg (propulsion) is related to that of the leading leg (stabilisation), and that the interaction between muscle powers during gait can reflect specific propulsion and control strategies that are related to each limb [61].

In this study, ankle plantar flexor activation timing has not been analysed; however, it seems that the activity of medial gastrocnemius activity preceded the contralateral heel strike, which is not surprising since, when a muscle is activated, it takes time before the muscle force is fully developed. The time taken to reach maximum force depends on factors such as muscle fiber type, activation level and contraction dynamics, but for isometric contractions, it ranges between 23-73 ms [62-64]. The preactivation period occurred before heel strike, and muscle activity within this period, is the result of feedforward control mechanisms. This is consistent with the evidence that the spinal co-ordination of bilateral leg muscle activation depends on a facilitation by supraspinal centres. Indeed, cerebellar contribution via reticulo-spinal neurons has been suggested in humans [65] and recent evidence was presented for a cortical (supplementary motor area) control of interlimb co-ordination [66]. Considering the information pointed above, it would be important in future studies to analyse plantar ankle flexors timing activity during double support.

It is becoming more and more accepted that, in addition to neural mechanisms, the mechanical properties of the body play a primary role in the dynamics and intrinsic frequencies with the complex nonlinear properties, to which the frequencies, phases and shapes of motoneuron signals must be adapted for efficient locomotion and motor control [67]. The results of this study demonstrate that there is a relation between the mechanics of the leading leg and the muscle activity of the trailing leg during step-to-step transition. The importance of ankle muscle activity in step-to-step transition is expressed in studies dedicated to this mechanism in transtibial amputed subjects [68] and subjects with total ankle arthroplasty [21]. These studies indicate that ankle impairment leads to a decrease of positive work by the trailing leg and a consequent increase of negative work by the leading leg, which partially explains the increased metabolic cost of walking. The results of our study demonstrate a higher correlation between MG EMGa and Fy, which corroborates its major importance in forward propulsion [38]. On the other hand, no significant correlation was observed between Fx at heel strike and MG EMGa during propulsion, which can result not only from the fact that MG major role is related to trunk support and forward displacement [16, 17] but also from the fact that Fx is the highest variable component [60]. This higher variability can explain the differences observed between dominant and non-dominant limbs at heel strike.

As to the results obtained, there are two important questions that need discussion. First, different values of Pearson correlation have been noted between the EMGa of dominant and non-dominant limbs and the GRF of contralateral limbs (the first have presented a moderate correlation and the second only a weak correlation). Differences between dominant and non-dominant limbs have been reported frequently, as lower limbs are not used equally during walking [69]. This asymmetry has been interpreted based on the support and mobility associated to each limb [70-72], as one leg contributes more to propulsion while the contralateral one is mainly responsible for support and body weight transfer during walking (dominant and non-dominant limbs, respectively) [73-75]. Therefore, it can be hypothesised that the higher correlation between the EMGa of the dominant limb and Fz and Fy of the contralateral limb results from the fact that the dominant limb contributes more to propulsion, and so it is more adapted to this function. On the other hand, evidence suggests that the dominant leg is stronger in plantar flexion [76] which allows accepting that during the double support phase, it is more related to Fz and Fy of the contralateral leg than the non-dominant limb. The second question is related to the growing evidence showing the compartmentalisation of the human gastrocnemius [77-80]. It has been demonstrated that portions of the same gastrocnemius muscle are activated differently, depending on the direction of the ankle force [79], and that surface EMGs recorded from the pinnated MG muscles are extremely selective [80]. Taking this information into account, it would be important, in future studies, to analyse the different activation patterns from distinct parts of the triceps surae muscle, as the possibility of having specific, localised MG regions involved in limb propulsion could be related to the finding that Fz and Fy only explain part of the MG EMGa.

Several studies agree that changes in walking speed are associated with increases in the intensity of muscle activation [70, 81-85]. The results of this study show that speed differences obtained between subjects were not related to MG EMGa and GRF. These findings can be explained by the fact that subjects walked at their own comfortable speed and mean values obtained were according to reference values [86]; in addition, standard deviation values were low, which is related to the high homogeneity of the sample. As to the influence of speed on GRF values, the results of this study are according to the ones obtained in [87], where the GRF increased linearly with gait speed only up to about 60% of the subjects' maximum speed. It is important to note that this study only addressed the correlation of subject walking speed on MG EMGa and GRF to exclude a possible effect of speed, as subjects were asked to walk at a comfortable speed. However, as changes in walking speed are associated with increases in the intensity of muscle activation and GRF magnitude, it would be important, in future studies, to analyse the influence of speed on the relation between MG EMGa of the trailing leg and GRF of the leading leg.

Another aspect that is important to note is related to the repeatibility of GRF peak measurements and limitations of the instruments. As stated in the instruments section, we have taken into account several considerations as to force platform mounting to avoid measurement errors. In addition, the coefficient of variation of GRF peak values obtained in each subject was almost always below 12.5%. Moreover, like in the present study, several other researchers used the GRF first peak value not only in healthy subjects [49, 88, 89] but also in subjects with pathology [90-93] and even as a measure to control the influence of an exercise program [94]. However, considering limitations in terms of repeteability of GRF peak measurements, it would be important in future studies to analyse the relation between muscle activity of one limb and the slope of the transient of GRF of the contralateral limb during double support.

Considering that the EMGa of the trailing leg was correlated with the magnitude of Fz and Fy of the leading leg, it would be important, in future studies, to assess how much of the negative work produced during heel strike might be compensated by this muscle.

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FIGURE CAPTIONS

Figure 1 – Correlation between MG EMGa of the non-dominant limb during propulsion and Fx of the dominant limb during heel strike (right), and between MG EMGa of the dominant limb during propulsion and Fx of the non-dominant limb during heel strike (left).

Figure 2 – Correlation between MG EMGa of the non-dominant limb during propulsion and Fy of the dominant limb during heel strike (right), and between MG EMGa of the dominant limb during propulsion and Fy of the non-dominant limb during heel strike (left).

Figure 3 – Correlation between MG EMGa of the non-dominant limb during propulsion and Fz of the dominant limb during heel strike (right), and between MG EMGa of the dominant limb during propulsion and Fz of the non-dominant limb during heel strike (left).

Figure 4 – Representation of one subject depicting how the raw EMGa of MG and the GRF profile vary for both limbs during double-support phase. At the left, one can see the absolute values of raw EMG signal of MG of the dominant leg (black) and the Fz (magenta), Fy (cyan) and Fx (green) of the non-dominant leg. At the right, the absolute values of raw EMG signal of MG of the non-dominant leg and the Fz, Fy and Fx of the dominant leg are shown.

TABLE CAPTION

Table 1 – Mean and standard deviation values of EMGa during propulsion and GRF at heel strike in dominant and non-dominant members and speed.

FIGURES

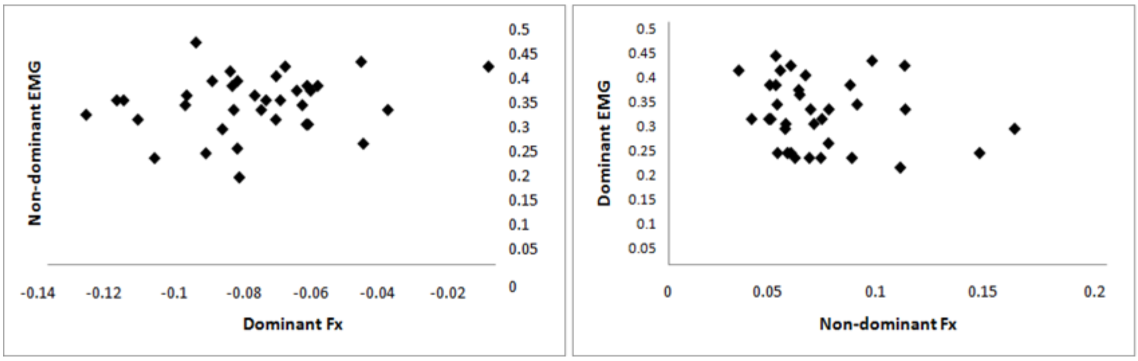

Figure 1

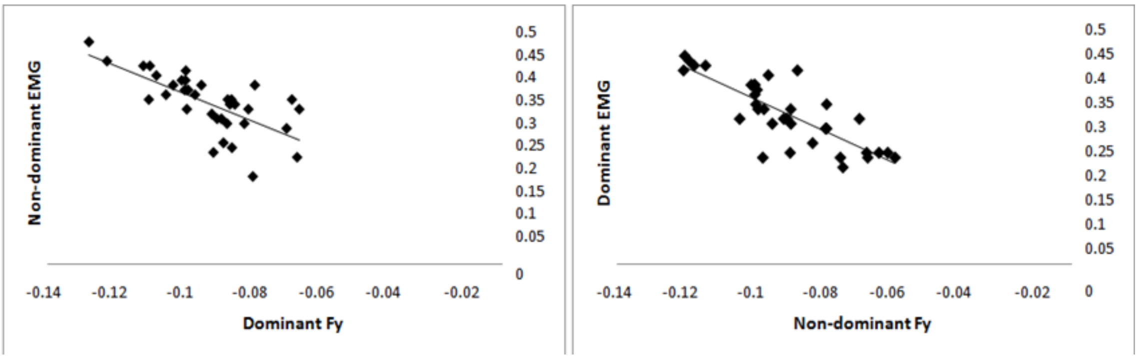

Figure 2

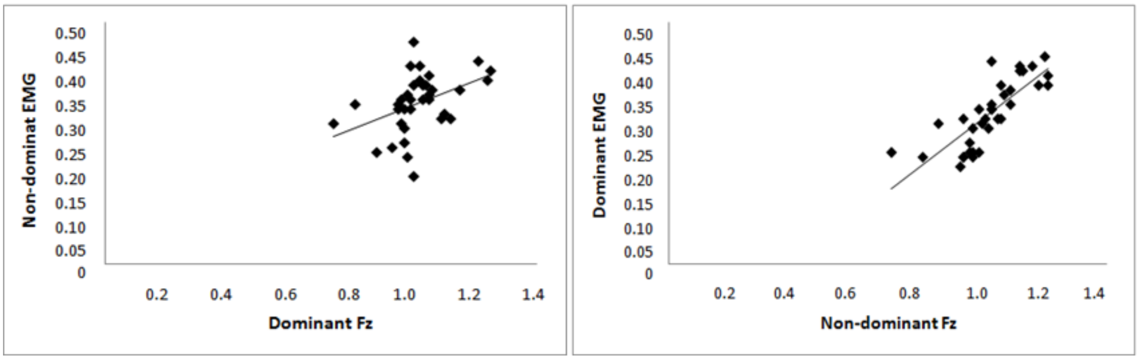

Figure 3

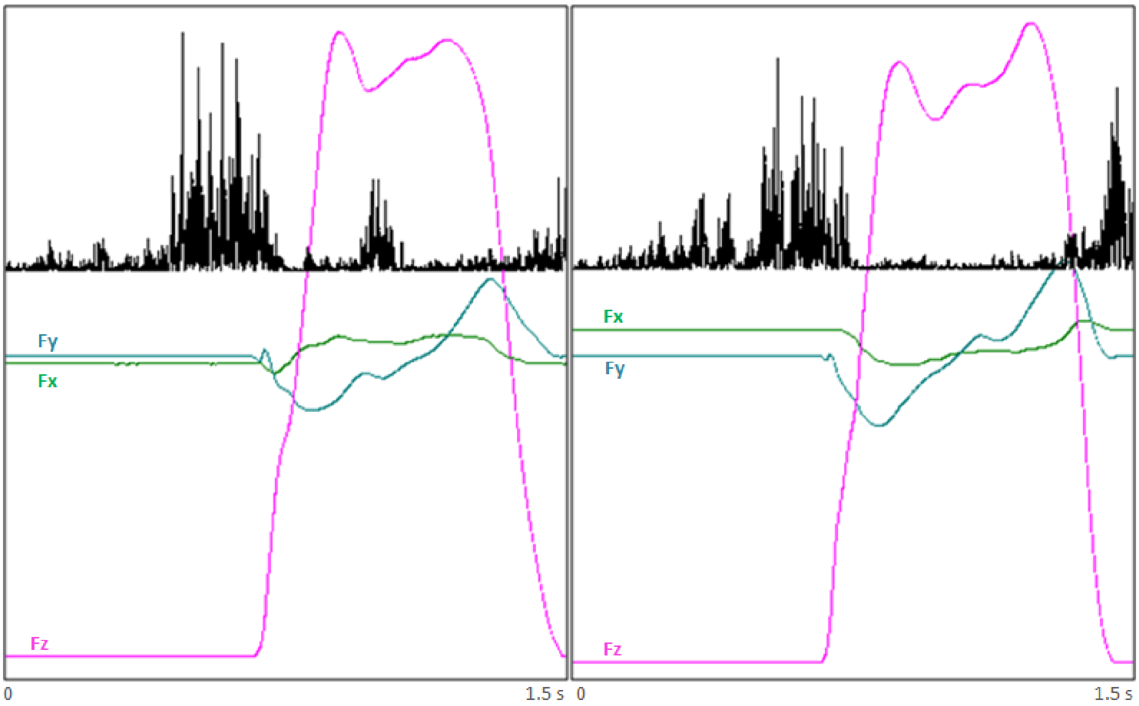

Figure 4

TABLES

Table 1

Trial	Component	N	Mean	Standard deviation
1	Dominant EMG	35	0.3110	0.0689
	Non-dominant:			
	Fz		1.0240	0.10960
	Fy		0.0917	0.01677
	Fx		0.0691	0.02771
	Speed		0.3010	0.04930
2	Non-dominant EMG		0.3330	0.06090
	Dominant:			
	Fz		1.0120	0.10570
	Fy		0.0941	0.01421
	Fx		0.0811	0.02329
	Speed		0.4490	0.06060