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Muscle synergies with Walkaround® Postural Support vs. "cane/therapist" assistance

BACKGROUND: The main clinical measures of walking recovery in stroke patients were compared for training assisted by Walkaround® postural support (WPS) and conventional (CON) support by a cane/therapist.

OBJECTIVE: We attributed the differences between the trainings to modified muscular synergies that occurred during assistance by WPS.

METHODS: We studied the muscle activities of the primary knee and ankle joint movers in the paretic and non-paretic legs of sub-acute stroke patients during assisted walking with WPS and CON. Recorded signals were compared to normative data that were recorded during speed-matched gait trials in healthy subjects. The specific measures were the relative contribution of individual muscles, levels of cocontraction, and the timing of the maximum electromyography (EMG) activity during the walking sessions.

RESULTS: We found that, for most patients, the individual contribution of muscles were more similar to the healthy with the WPS assistance. In parallel, the cocontraction of the rectus femoris muscles in both legs was lower (by up to 39 %) during walking assisted by WPS than by cane/therapist gait support); the results from this case series (10 patients) showed that WPS might be the superior training scheme

CONCLUSIONS: These findings indicated that assistance by WPS changed the motor control output relative to CON assistance in most patients.

KEYWORDS: EMG, posture, rehabilitation, stroke, synergy

1. Introduction

Motor disability in stroke patients often manifests as reduced walking capacity. The current neurorehabilitation strategy for stroke patients is the intensive repetition of movements (task related exercise). This approach is based on the rationale that intensive task repetition mediates learning and neural plasticity through a process of sensory motor integration [8-9].

Intensive gait exercise could be assisted through the use of a harness to support the body and to prevent falls on a powered treadmill while the therapist assists the leg movement of the paretic leg [7, 3]. A more recent approach uses robot assistants to support the patient on a powered treadmill (Locomat, Hocoma AG; Volketswil, Switzerland). A randomized crossover study found that subjects with hemiparesis showed greater improvements in gait function and reduced lower-limb impairment measures following Locomat® training compared to equal periods with conventional physical therapy [3, 18]. Hesse et al. [11] demonstrated that the Advanced Gait Trainer® (Rehastim, Berlin, Germany) with a powered foot mover was more effective than therapist-assisted walking on a powered treadmill [29]. Another study [4] showed that the external forces provided by either an Advanced Gait Trainer® or a Locomat® could produce passive effects, possibly regarding the recovery process because the "patient is moved, and his muscles are not engaged".

Walking on a treadmill and overground walking are similar but involve different muscular strategies. Overground walking enables actual progression in space, which is more motivating to the patient than walking on a treadmill. An externally powered walker termed Walkaround® was introduced as the assistance for overground walking [27]. The Walkaround provides postural assistance and trunk orientation through a lumbar belt and prevents from falling. This walker (Fig. 1) enables partial body weight support, and facilitate long overground walking training sessions without hand support at speeds of up to 1.2 m/s. The lumbar belt allows the patient's center of mass to move appropriately during walking (up/down and left/right), which is a prerequisite for normal walking [28].

A recent clinical study on stroke subjects (Table 1) compared clinical scores (the Fugl-Meyer Score for the lower extremities, the Berg Balance Score, and the Barthel Index) and kinematical measures (the stride and the velocity) for two treatment groups. Patients were randomly assigned to a treatment group with Walkaround® postural support (WPS) and a control group in which conventional assistance by a cane/therapist (CON) was provided. The therapy comprised 30 minutes of daily exercise over four weeks (corresponding to a minimum of 16 sessions). The results of this study showed that the clinical scores of the stroke patients increased more for the WPS group than for the CON group (Table 1) with significant changes in Berg Balance Score. There were no reported side-effects in the study [24].

These findings motivated us to study the differences in motor control output (muscle synergies) when patients were assisted by the Walkaround® versus a cane/therapist. It has been shown [13-15] that surface polymyography recordings can be used to analyze motor control. These studies presented strong temporal and spatial organization of the muscle activities, which are known as synergies, for different types of walking. Hesse et al. [12] showed that the amplitude of gastrocnemius muscle contractions increased in the paretic leg and that only one patient showed healthy like patterns for the tibialis anterior muscle and the thigh muscles after therapy. Otter et al. [22] investigated whether healthy like gait recovery was accompanied by changes in the temporal patterns of muscle coactivation when patients walked on a treadmill with partial body weight support. The authors found that the improvement in clinical and kinematical measures was not associated with a more functional pattern of muscle activities. The authors concluded that physiological processes other than improved temporal muscular healthy like pattern coordination determined rehabilitation progress after a stroke. The working hypothesis for this study was that the muscles synergies produced by Walkaround® assistance (i.e., the WPS group, Fig. 1, left panel) are different from those produced by conventional assistance during gait exercise (i.e., the CON group, Fig. 1, right panel). We used the same methods in this study as in our previous research [24].

This study was designed with the objective of assessing the differences in the motor output between two different treatments (body postural support, as represented by WPS, and hand support, as represented by CON) under the same conditions and without the effects of training.

2. *Methods and materials*

2.1 Subjects

Ten sub-acute stroke patients (Table 2) and five healthy volunteers (49.3±5.6 mean age ± SD) participated in this study after signing an informed consent form, which had been approved by the local ethics committee. The inclusion criteria were the following: (1) this was the patient's first stroke (i.e., an ischemic or intracerebral hemorrhagic lesion), (2) the patient had a Functional Ambulation Categories score of at least 2 (i.e., the patient needed the continuous or intermittent support of one person to help him/her with balance or coordination) or at most 4 (i.e., the patient could walk independently on level ground, but required help on stairs, slopes or uneven surfaces), and (3) the patient became involved in the study at least one month after the stroke. The exclusion criteria for stroke patients were the following: (1) peripheral nerve lesions, (2) orthopedic problems (substantial restricted passive range of motion in the major lower limb joints,), (3) depression or poor motivational capacity, and (4) severe forms of aphasia or other cognitive problems that could hinder communication or cooperation. Table 3 shows the clinical outcome measures used in this study: the Barthel Index (BI), the Fugl-Meyer Scale (FM) for the lower extremities, the Berg Balance Scale (BB), the duration of the stance and swing phases for the non-paretic and paretic leg, and the velocity during overground walking.

2.2 Instrumentation

A diagram of the acquisition system and the data processing steps of the study are shown in Fig. 2.

We recorded the surface EMG from the prime movers of the leg joints: the tibialis anterior (TA), the lateral gastrocnemius (LG), the rectus femoris (RF), and the biceps femoris (BF) muscles. The recording electrodes were placed following the SENIAM protocol [10, 19]. We used disposable pre-gelled EMG Ag/AgCl electrodes with 10-mm flat pellets (GS26, Bio-Medical Inc., Warren, USA). The EMG signals were amplified by Biovision preamplifiers (Biovision Inc, Wehrheim, Germany). The gain of the preamplifiers was set to 1000. A reference electrode was placed over the patient's right knee. Signals were acquired using an AceLAB setup [21], which included a NI USB 6212 AD card (National Instruments, Inc., Austin, USA) with an AD resolution of 16 bits. We used a custom-designed acquisition software application that was created in LabVIEW (National Instruments, Inc., Austin, USA). The sampling rate was set to 1000 samples per second.

Along with the EMG, we recorded signals from two force sensing resistors (FSR) that were placed on the metatarsal and heel zones of the patient's sole. The FSR data were decimated by a factor of 10. The FSR signals were recorded to enable the gait phases to be separated during data analysis.

2.3 Data processing

The EMG data were filtered with a notch filter (50 Hz) and a first-order modified differential infinite impulse response (IIR) filter to remove the baseline offset. Next, we rectified the filtered data and generated the EMG envelopes [31]. We estimated the relative contribution (C_i) of the muscles using a method developed by Katz et al. in [16] which we adapted in our earlier study [20]. The relative contribution is the ratio of the contribution of a specific muscle to the total effort of all the muscles, as shown in Eq. 1:

$$C_i = 100\left(\frac{A_i}{A}\right),\ \mathrm{A} = \sum_{\mathrm{i-1}}^{4} \mathrm{A_i}\ ,\ i = 1,2,3,4 \qquad \text{Eq. 1}$$

where 1, 2, 3 and 4 denote the TA, LG, RF and BF muscles, respectively. A_i denotes the area under the EMG envelope of a muscle during the discrete interval, and A denotes the sum of the activities of all four muscles during the same interval. That is, the relative contribution C_i shows the level of activity of a specific muscle relative to the activation of all four muscles, which is expressed as a percentage. We calculated the C_i parameters for a paretic leg C_i^P and a non-paretic leg C_i^{NP}, in addition to the C_i^H. Values near 0 indicate that a muscle was not activated during the discrete time interval while other muscles were, and values near 100% indicate that the muscle was the main contributor to the function [11].

We segmented the recorded signals to strides and analyzed a series of representative steps, after excluding the outliers during the initiation and termination of the walking. We also cropped the stance phase into seven equal segments and the swing phase into three equal segments [2]. We calculated the ratios $D_i^{P,WPS}$, $D_i^{P,CON}$, $D_i^{NP,WPS}$ and $D_i^{NP,CON}$ using the values of the relative contributions from the data obtained during the WPS and CON gaits and a healthy gait at a speed (v ≈ 0.4 m/s) that matched the patients' gait (Eq. 2):

$$D_i^P = \frac{C_i^H - C_i^P}{C_i^H}$$

$$D_i^{NP} = \frac{C_i^H - C_i^{NP}}{C_i^H},\ i = 1,2,3,4 \qquad \text{Eq. 2}$$

where 1, 2, 3 and 4 denote the TA, LG, RF and BF muscles, respectively. The parameters D_i^P (for a paretic leg) and D_i^{NP} (for a non-paretic leg) correspond to the ratios between muscle contributions in a stroke patient and those in a patient with a healthy gait. Negative values of D_i^P and D_i^{NP} indicate higher muscle contribution in a stroke gait than in a healthy gait, while positive values of D_i^P and D_i^{NP} indicate a lower contribution from a stroke gait than from a healthy gait.

The heel contact and toe-off events were used to determine the phases of the gait cycle automatically by setting the threshold at 5% of the maximum force (where the sum of the signals was measured at the heel and metatarsal zones). The analysis was performed using a custom-designed program in a Matlab (The Mathworks, Natick, MA, USA) environment.

2.4 Procedure

A recording session was conducted in the morning to minimize any differences incurred by daily activities. The patients were asked to walk a distance of 10 meters on level ground, twice with WPS and twice with CON. The patients relaxed for approximately 10 minutes in between measurement sessions to reduce the effects of fatigue.

The surface EMG for healthy subjects (in a year matched group) was assessed during the subjects' gaits at v≈0.4 m/s.

3. Results

Table 3 shows the clinical measures for the patients at the entry point of the study.

Fig. 3 shows the EMG signals and the ground reaction forces for several strides in the middle of a 10-m-long path, which was measured during the CON (left panels) and WPS (right panels) gaits.

Fig. 4 shows the rectified and filtered EMG signals from the paretic and non-paretic legs of patient ID=P7, for a representative single stride, when assisted by CON and WPS. For the example shown, the T_{swing} of the non-paretic leg lasted for 23% of the stride period during the CON gait and 34% during the WPS gait. The T_{swing} for the paretic leg increased from 26% when the patient walked with CON to 38% when the patient was assisted by WPS. The mean gait speeds were v = 0.43 m/s (CON) and v = 0.48 m/s (WPS). Fig. 4 also shows the EMG envelopes for a healthy gait.

The relative contributions were plotted as a cocontraction map, in which the most active muscle had the largest surface, Fig. 5. Fig. 5 shows the data for stroke patient ID=P7 for the muscles in the paretic and non-paretic legs during the WPS- and CON-assisted gaits. The relative contributions were calculated for 100-ms-long intervals using Eq. 1.

Fig. 6 shows the D_i^P and D_i^{NP} ratios for the non-paretic and paretic legs of patient ID=P7. The 0 value signifies that the muscle contribution in the stroke gait is equal to the muscle contribution in a healthy gait. The step cycle was divided into 10 intervals, the stance phase was divided into 7 equal intervals, and the swing phase was divided into 3 equal intervals. The grey bars correspond to the stance phase, and the red bars correspond to the swing phase in Fig. 6. Fig. 7 shows the averaged and normalized ratios of the muscles' contributions for the paretic and non-paretic legs of all of the patients during the CON and WPS gaits.

4. Discussion

The WPS gait speed was approximately 10% higher than the CON gait speed (Table 1). This difference followed the comfort of the patient when walking. This difference in gait speed can also cause differences in muscle activation [1, 5, 24].

Fig. 3 shows that the swing phase of the gait was prolonged for the WPS gait compared to the CON gait. For the entire group of 10 patients, on average, the swing phase has changed from 26±7 % to 32±8 % in the non-paretic leg and from 27±8 % to 35±7 % in the paretic leg (WPS *vs.* CON). The prolonged swing phase resembled that of the healthy gait in the age-matched group, although the anticipated asymmetry between the paretic and non-paretic legs remained.

The individual muscle contributions and the cocontraction map (Figs. 4 - 7) enabled us to assess more detailed dynamic properties of the muscle contributions during the gait and to analyze these results from the perspective of results in the literature [25]. Fig. 4 shows that the EMG profiles for both the paretic and the non-paretic leg were different from those for the healthy gait. However, the WPS-assisted gait patterns resembled healthy gait patterns more than did the CON-assisted gait patterns. WPS assistance affected the EMG patterns in the non-paretic leg much more noticeably than in the paretic leg. To better understand the modified EMG patterns, we analyzed the relative muscle contributions (Fig. 5). The simultaneous activities of different muscles can be analyzed using this map. Each muscle contribution was plotted as a colored area. The sum of all activities was 100%. The horizontal axes were normalized to the duration of the stride (100%).

The positive values in Fig. 6 show that the patient's muscle contributions were lower than for healthy muscle contribution, while the negative values indicate that the patient's muscle contributions were higher than for healthy muscle contribution. Fig. 6 shows the relative muscle contributions with their standard deviations for ten gait strides. By far, the largest difference (by an order of magnitude) in the ratios was found for the RF muscle (see the 3rd row of plots in Fig. 6). The differences in the ratios were a fraction smaller during the WPS gait than the CON gait, except for the ratio for the RF muscle in Fig. 7. The standard deviations were large, which is to be expected given the number of patients and the differences in their functional status.

The results of the case series of 10 patients recorded are summarized below.

1. The TA activity during the WPS gait was more similar to that of a healthy gait than that of a CON gait. The contribution of the TA muscle during the stance-swing transition period, which was observed in the WPS gait but not in the CON gait, plays an important role in controlling ground clearance by causing rapid dorsiflexion of the foot [30-31].

2. The cocontractions of the TA and LG were reduced in the non-paretic leg when assisted by WPS than by CON (Fig. 7). The cocontraction of the TA and LG muscles increases ankle stiffness and is likely a compensatory strategy that develops to cope with the deficits in the paretic leg. Lark et al. reported that inadequate ankle-joint stiffness during the single support phase compromised the control of forward and downward body momentum, leading to difficulties with gait initiation and termination when the gait was initiated by the non-paretic leg [17]. By providing postural body support and forward momentum in the direction of walking at a pre-set speed, it might be suggested that WPS offered just enough support to prevent compensatory cocontractions (see Fig. 5, lower left panel).

3. The BF activity during the swing-stance transition period for the WPS gait resembled that of a healthy gait. This might indicate that eccentric contractions decelerated the leg at the end of the swing phase, allowing a more normal heel contact. This behavior can be observed from the peaks in Fig. 5 at the end of the swing phase in the non-paretic and paretic legs.

4. A healthy BF contracts concentrically at the beginning of the stance phase, producing hip extension as the leg approaches the ground. This pattern was only observed for the WPS gait in the paretic leg (Fig 5).

5. The activity of BF at the stance-swing transition may seem pathological because this activity is not observed in the healthy gait pattern in Fig 5. However, several studies on healthy EMG patterns [23, 26] have reported inconsistencies in the second phase in BF activity, consisting of an eccentric contraction that causes the hip joint to flex. The main role of the BF muscle at this point might be to assist knee flexion and to provide a normal swing.

6. The major difference in the level of activity of stroke patients compared with that of healthy subjects was attributed to RF activity. The differences were one order of magnitude larger than for the other three muscles (Fig. 6). This increased activity was probably partly a result of the reduced activity of the LG and partially a result of the leg not being fully extended during the stance (i.e., corresponding to a crouched gait). The reduced LG activity during the mid-stance phase suggested that there might be decreased ability for push-off. This reduced activity was observed less during the CON gait than during the WPS gait. This difference could most likely be attributed to the use of hand support, but it may also have been a consequence of the momentum provided by WPS.

7. Figs. 6 and 7 show low individual relative contributions, as estimated by Eq. 2, for WPS support compared to CON support. The coefficient D_i is in the interval 0 and 1. The low values (close to 0) showed that the activity of a specific muscle in a stroke patient was similar to that in healthy subjects. This result demonstrates the synergy changes caused by WPS

support in gait exercise while also suggesting that further or additional assistance is required to reduce the activity of the RF muscle.

8. The shift of the maximal value in RF can be seen in WPS group compared to CON group. This delay was noticed in all patients in paretic leg of up to 14 %. No phase shift was noticed in non paretic RF muscle, Fig. 7. This indicates the increase of temporal symmetry between paretic and non-paretic leg observed by peak changes of D_i coefficients during stance phase in Fig. 7 in WPS group compared to CON group.

5. Conclusion

An EMG analysis was performed on the primary movers in the leg during the gait in stroke patients, with body postural assistance (WPS) and assistance by conventional means (CON): larger effects were observed for assistance by WPS compared to CON. The results from this small clinical trial suggest that the WPS produced near-normal ankle flexor and extensor activity, as well as hip extensor/knee flexor (BF) activity. The difference between the CON and WPS gaits was smaller for knee extensor/hip flexor (RF) activity. WPS support affected both the paretic and non-paretic legs in all patients. WPS did not eliminate the asymmetry between the paretic and non-paretic leg, but it did result in a prolonged swing phase that resembled a healthy pattern more than the CON pattern.

The conclusions from this study which included 10 stroke patients call for a larger controlled randomized clinical trial.

Declaration of interest

There is no conflict of interest.

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Tables

Table 1: Averaged differences in outcome measures between the end and beginning of the study for the WPS and CON therapies [24]; the acronyms are the same as in Table 1, and SD denotes the standard deviation.

ΔBI±SD	ΔFM±SD	ΔBB±SD	Δv±SD	ΔT_P±SD	ΔT_{NP}±SD
10.78±4.63	2.44±1.51	7.22± 3.73*	0.1±0.02	-0.32± 0.1	-0.21±0.08

The asterisks * denote significant differences ($p<0.05$).

Table 2: Data for the randomly selected patients in this study

Subject ID	Sex	Stroke onset (days)	Stroke type	Paretic side	Age (years)
P1	Female	93	Ischemic	Right	57
P2	Female	108	Ischemic	Right	57
P3	Male	44	Ischemic	Right	69
P4	Male	57	Hemorrhagic	Left	53
P5	Male	87	Hemorrhagic	Left	53
P6	Male	196	Ischemic	Left	76
P7	Male	44	Ischemic	Right	45
P8	Male	76	Ischemic	Right	45
P9	Female	60	Ischemic	Left	62
P10	Male	72	Ischemic	Right	79

Table 3: Patients' clinical scores and kinematic measures at the onset of therapy; the acronyms denote the Barthel Index (BI), the Fugl-Meyer Score (FM), the Berg Balance Score (BB), the mean velocity of gait (v), the stride time interval (T), the paretic leg (P), and the non-paretic leg (NP); SD denotes the standard deviation

Subject ID	BI max 100	FM max 34	BB max 56	V (SD)	T NP (SD)	T P (SD)
P1	80	22	48	0.41 (0.02)	1.94 (0.08)	1.94 (0.07)
P2	95	25	51	0.51 (0.27)	1.93 (0.07)	1.87 (0.06)
P3	99	20	--	0.43 (0.16)	1.53 (0.06)	1.54 (0.03)
P4	92	21	44	0.36 (0.03)	2.61 (0.29)	2.50 (0.18)
P5	97	22	48	0.38 (0.02)	2.14 (0.04)	2.23 (0.04)
P6	85	23	26	0.38 (0.01)	1.57 (0.08)	1.57 (0.09)
P7	87	30	42	0.30 (0.01)	2.09 (0.04)	1.98 (0.05)
P8	98	32	51	0.42 (0.01)	1.84 (0.04)	1.82 (0.08)
P9	58	14	17	0.27 (0.05)	2.70 (0.20)	2.82 (0.07)
P10	80	22	38	0.34 (0.02)	2.00 (0.08)	1.92 (0.06)

Figure captions and Figures

Figure 1: A stroke patient (ID=P4) is shown assisted by a Walkaround® (WPS) (left panel), as well as by a cane and a therapist (CON) (right panel) during an assessment session in the "Dr Miroslav Zotović" Rehabilitation Clinic, Belgrade. The assessment included measurements of kinematics and dynamics [6, 21], as well as the surface EMG (for the tibialis anterior; lateral gastrocnemius; rectus femoris; and biceps femoris).

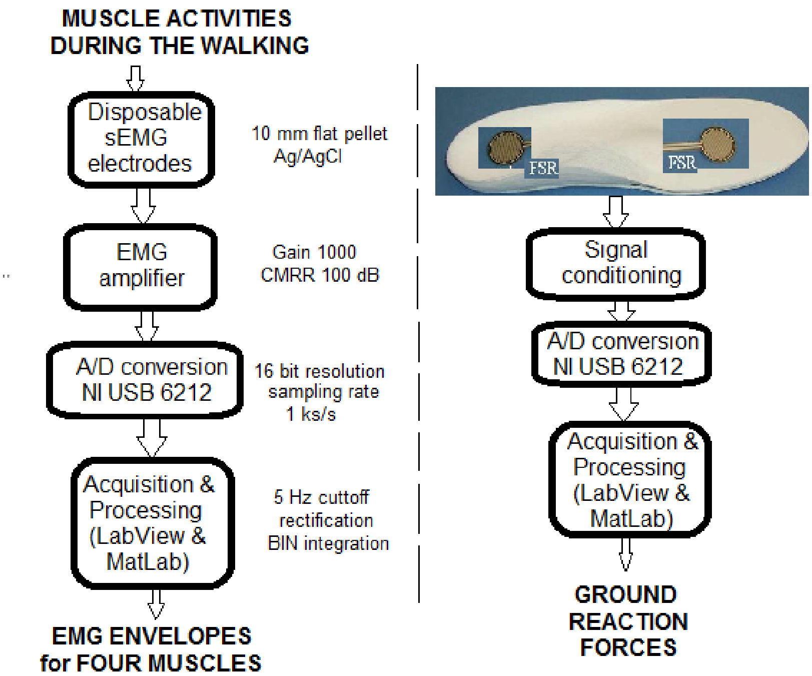

Figure 2: Flow diagram of the data processing steps

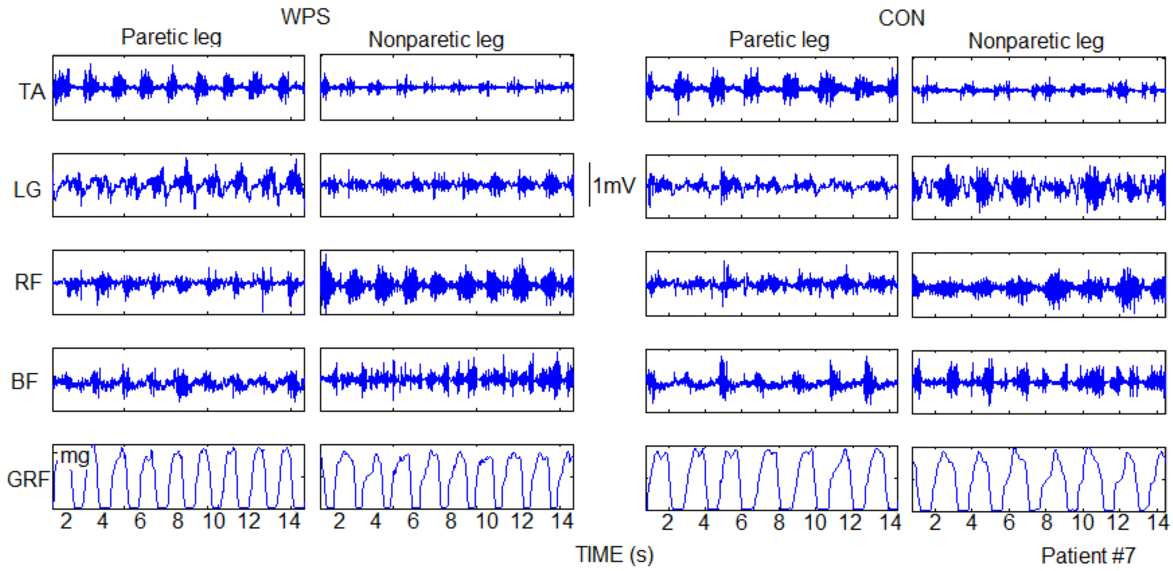

Figure 3: Surface EMG signals from patient ID=P7 during assistance by CON and WPS; the acronyms are defined as follows: TA – tibialis anterior, LG – lateral gastrocnemius, RF – rectus femoris, BF – biceps femoris. The bottom panels show the ground reaction forces as assessed by the FSR; note that the WPS graphs include one more step during the same interval

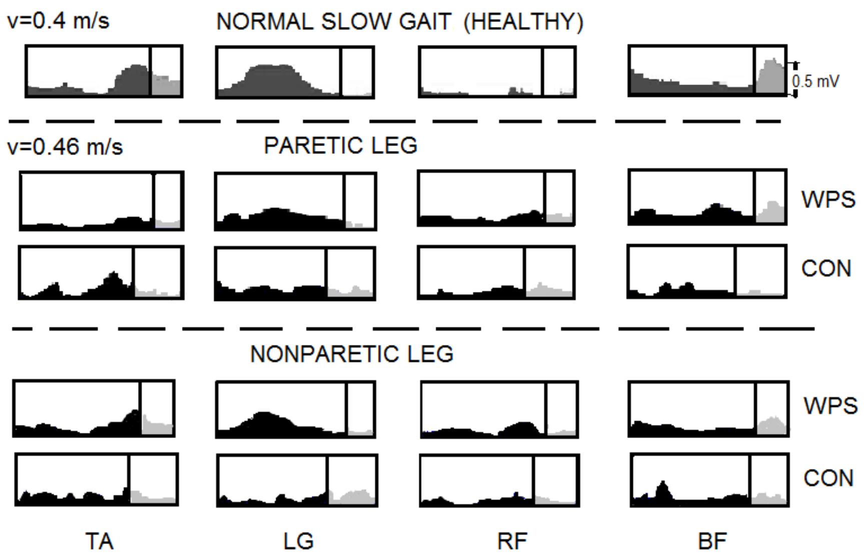

Figure 4: The top panels show the averaged EMG envelopes during a slow healthy gait (v = 0.4 m/s). The lower panels show the EMG envelopes for a single stride (for stroke patient ID=P7) during the CON gait and the WPS gait at speeds of 0.43 m/s and 0.48 m/s, respectively. The acronyms for the muscles are as follows: TA – tibialis anterior, LG – lateral gastrocnemius, RF – rectus femoris, BF – biceps femoris. The vertical lines indicate the end of the stance phase. Note the difference in the ratio of the durations of the stance/swing phases.

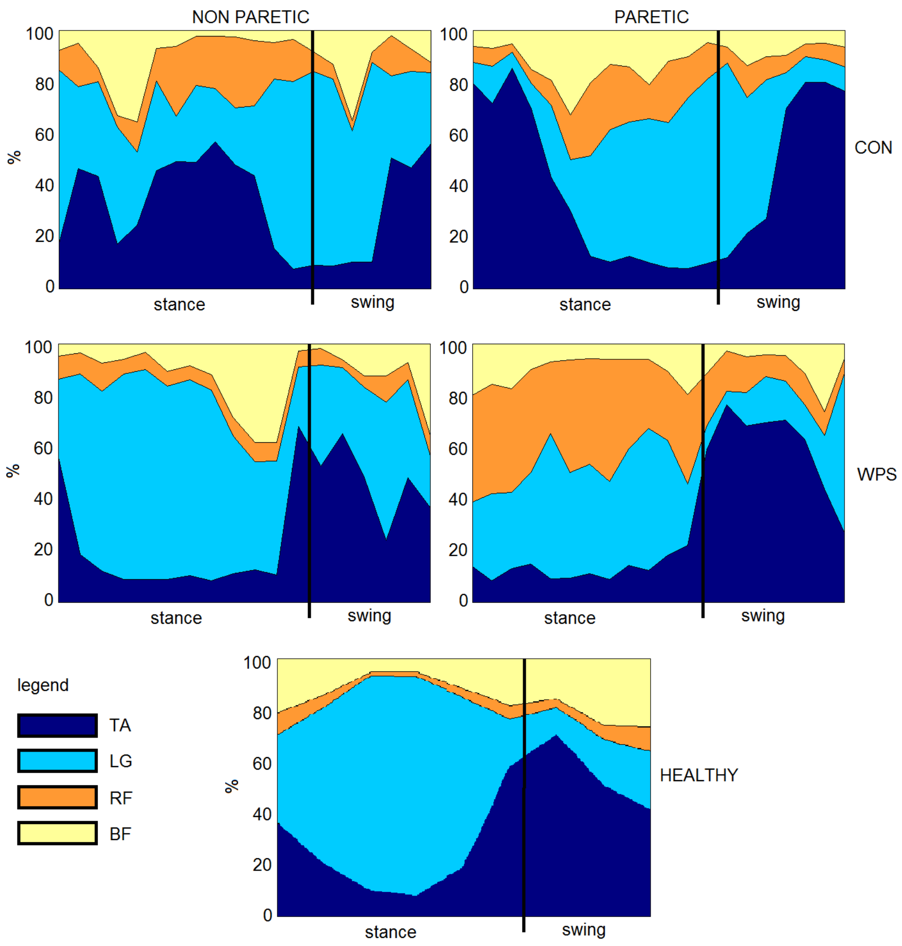

Figure 5: Calculated Ci parameters for time intervals of T=100 ms during a single stride period for paretic and non-paretic legs (top panels) and a healthy gait (bottom panel) (see text for details)

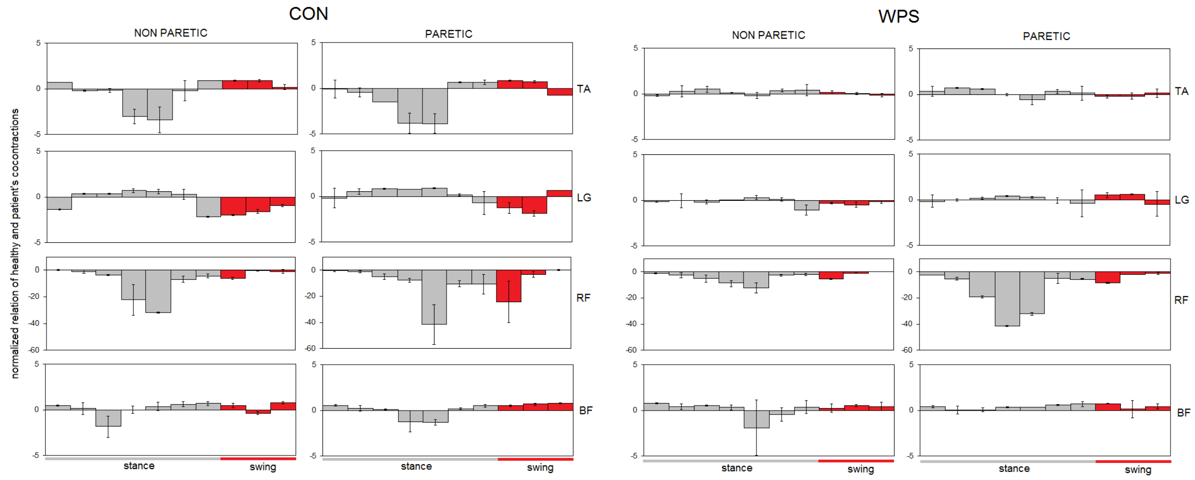

Figure 6: The mean ± standard deviation of the contribution of single muscle activities for non-paretic and paretic legs during the WPS and CON gaits (patient ID=P7) compared with data for a healthy gait (see text for details)

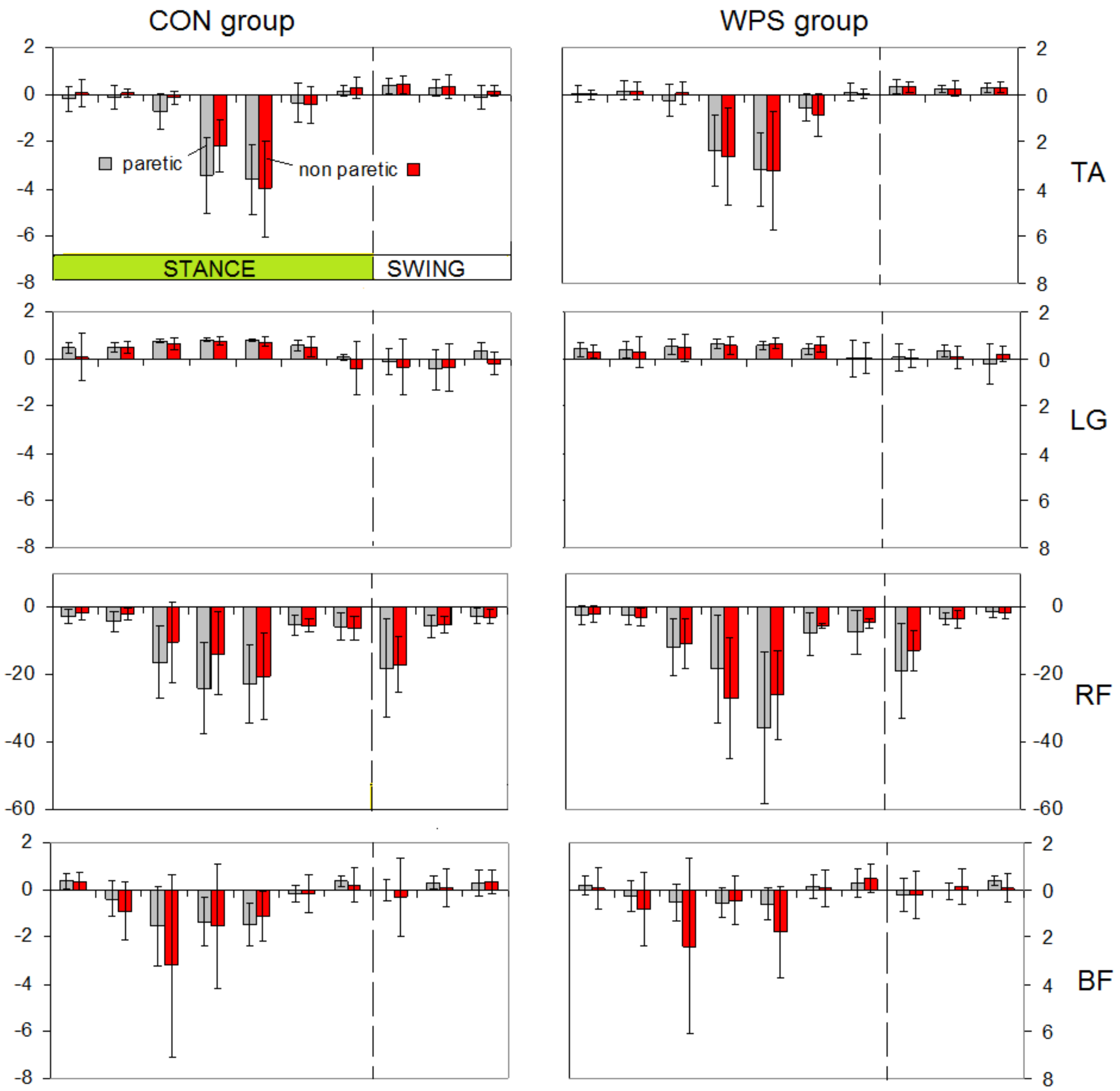

Figure 7: Mean ± standard deviation of the ratios of the relative contributions (for all 10 patients) of individual muscles for non-paretic and paretic legs for the WPS and CON gaits compared with the contributions of the same muscles for a healthy gait.