Crouch gait can be an effective form of forced-use/no constraint exercise for the paretic lower limb in stroke

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In hemiplegic gait the paretic lower limb provides less muscle power and shows a briefer stance compared with the unaffected limb. Yet, a longer stance and a higher power can be obtained from the paretic lower limb if gait speed is increased. This supports the existence of a 'learned non-use' phenomenon, similar to that underlying some asymmetric impairments of the motion of the eyes and of the upper limbs. Crouch gait (CG) (bent-hip bent-knee, about 30° minimum knee flexion) might be an effective form of 'forced-use' treatment of the paretic lower limb. It is not known whether it also stimulates a more symmetric muscle power output. Gait analysis on a force treadmill was carried out in 12 healthy adults and seven hemiplegic patients (1–127 months after stroke, median: 1.6). Speed was imposed at 0.3 m/s. Step length and single and double stance times, sagittal joint rotations, peak positive power, and work in extension of the hip, knee, and ankle (plantar flexion), and surface electromyography (sEMG) area from extensor muscles during the generation of power were measured on either side during both erect and crouch walking. Significance was set at *P* less than 0.05; corrections for multiplicity were applied. Patients, compared with healthy controls, adopted in both gait modalities and on both sides a shorter step length (61–84%) as well as a shorter stance (76–90%) and swing (63–83%) time. As a rule, they also provided a higher muscular work (median: 137%, range: 77–250%) paralleled by a greater sEMG area (median: 174%, range: 75–185%). In erect gait, the generation of peak extensor power across hip, knee, and ankle joints was in general lower (83–90%) from the paretic limb and higher (98–165%) from the unaffected limb compared with control values. In CG, peak power generation across the three lower limb joints was invariably higher in hemiparetic patients: 107–177% from the paretic limb and 114–231% from the unaffected limb. When gait shifted from erect to crouch, only for hemiplegic patients, at the hip, the paretic/unaffected ratio increased significantly. For peak power, work, sEMG area, and joint rotation, the paretic/unaffected ratio increased from 55 to 85%, 56 to 72%, 68 to 91%, and 67 to 93%, respectively. CG appears to be an effective form of forced-use exercise eliciting more power and work from the paretic lower limb muscles sustained by a greater neural drive. It also seems effective in forcing a more symmetric power and work from the hip extensor muscles, but neither from the knee nor the ankle. *International Journal of Rehabilitation Research* 00:000–000 Copyright © 2017 The Author(s). Published by Wolters Kluwer Health, Inc.

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Introduction

In hemiplegic gait the muscles of the paretic lower limb provide less mechanical power compared with the homologous muscles of the unaffected side (Olney *et al.*, 1994). Kinematic asymmetries follow, causing a visible limping, allowing expert professionals to infer also the dynamic deficit, mostly the reduced power of plantar flexion at push-off (McGinley *et al.*, 2003). This asymmetry might represent an example of learned nonuse (LNU). LNU is a behavioral interpretation of the observation that the paretic upper limb after stroke can recover some voluntary motion if the unaffected limb is restrained, thus leading to the forced-use (FU) of the disused limb. The origin of the model, rooted in the neurophysiology of the latest 19th century, is summarized in two recent reviews (Fritz *et al.*, 2012; Kwakkel *et al.*, 2015).

Taub must be credited for first exploiting the potential for rehabilitation of the LNU model when extended to upper limb rehabilitation in poststroke hemiparesis (Taub, 1980). He developed a therapeutic paradigm on the basis of (a) intensive graded practice of the paretic upper limb, (b) restraint applied to the unaffected limb (typically a padded mitt to be worn for most waking hours), and (c) behavioral methods ('transfer package') aiming at enhancing the patient's adherence and the generalizability of the outcome to daily living (e.g. keeping a daily diary, doing written assignments, receiving weekly telephone calls from the monitoring therapist). He named this approach CIMT, after 'constraint induced movement therapy' (Taub *et al.*, 2014). CIMT and many subsequent modifications are now applied worldwide, not only in hemiplegia after stroke (Meinzer *et al.*, 2007; Buesch *et al.*, 2010; Bolognini *et al.*, 2011; Reid *et al.*, 2015). All modified CIMT protocols emphasize the restraint as a requisite, and in some instances a sufficient ingredient, for FU. By contrast, Taub states that restraint 'is the least important component of CI therapy and can be dispensed with entirely if the training conditions are arranged appropriately' (Taub, 1980). Stated otherwise, unlike most of what the literature assumes, FU does not necessarily imply restraint: massive practice might be in itself an even more effective form of FU.

Very few studies have explored the LNU model as applied to the lower limb in man, although LNU was advocated by Taub as a determinant of the lower limb paresis in stroke (Taub *et al.*, 2014). He proposed a specific CIMT protocol also applied to patients with multiple sclerosis (Mark *et al.*, 2013). The 'constraint' ingredient was omitted because 'both lower limbs must be simultaneously engaged for weight-bearing on massive practice tasks' (Mark *et al.*, 2013). Tesio proposed three simple types of FU exercises (Tesio, 2001). These consisted of abdominal curl, sit-up, and walking modified so that the motion of the unaffected lower limb was hindered by a disadvantageous limb positioning in the case of abdominal curl and sit-up and by physical orthotic restraint of the hip, knee, and ankle joints in the case of walking. He named these exercises 'occlusive', thus emphasizing the analogy with the old established eye-patching treatment applied since centuries to the sound eye in strabismus (Tesio, 1991). In the present work, an even simpler form of FU exercise not requiring hindrances to the unaffected limb and specific for gait is proposed, that is crouch gait (CG). It is hypothesized that (i) in CG, the paretic lower limb of hemiplegic patients is actually forced to increase its muscle work and power, in agreement with the LNU model and (ii) in hemiplegic patients, both spatiotemporal and dynamic parameters of gait are more symmetrical between the subsequent steps compared with erect gait.

Participants and methods

Participants

Healthy adults were enrolled as controls for gait analysis. They were recruited mostly among the hospital staff, as well as students and the residents attending the laboratory. Only patients with mild hemiparesis were selected. They had been or were presently being treated in the rehabilitation department running the experiments. Also, they all had to be able to understand the text of the informed consent (see the Ethics section) and the researcher's instructions. They had to be free from comorbidities potentially affecting gait capacity, and in particular, balance deficits, sequelae of major orthopedic interventions, major rheumatic, and other neurologic diseases. They had to be able to walk in full autonomy for at least 50 m, with no balance deficits. Previous experience showed that all participants spontaneously adopt a lower speed while on the treadmill compared with ground walking, the more the worse their impairment. Only 'high-functioning' hemiplegic patients could adopt speeds greater than or equal to 0.3 m/s on the treadmill, with neither previous training nor high attention demands, and risks for stumbling and falling. However, lower speeds were considered to be little representative of walking in daily life.

Loading symmetry during standing

Loading symmetry on the two lower limbs was tested before gait analysis (see below) through static stabilometry (Balance Master platform; Neurocom Inc., Clackamas, Oregon, USA). The participants were requested (a) to sit up and then to stand still for 5 s on independent plates under each foot; (b) to stand still on the force plates with the knee extended or flexed at 30°, 60°, and 90°, in three subsequent 20 s trials. The platform records mean vertical forces. The mean result of three bouts is considered the outcome for each testing condition.

Gait analysis

Instrumental setting

The methods to record surface electromyography (sEMG) and joint kinematics and dynamics have been described in detail elsewhere (Tesio and Rota, 2008). Briefly, gait was analyzed on a split-belt, force treadmill (so-called GAFT, gait analysis on force treadmill), placed in a dedicated room 3.40 m long × 5.20 m wide × 3.70 m high. Only the sagittal plane of motion is considered here. In fact, external work and power in the frontal and horizontal plane represent 1–4% of the same variables measured in the sagittal plane (Cavagna, 1975; Tesio *et al.*, 1998a). The following gait parameters were recorded synchronously.

(1) *Ground reactions, joint kinematics, and dynamics*
Two parallel independent treadmills, 0.3 m wide × 1.26 m long (model ADAL-3D-F-COP-Mz; Medical Development, Tecmachine Hef, Andrezieux Boutheon, France), are mounted each on four three-dimensional piezoelectric force sensors KI9048B (Kistler Co., Wintherthur, Switzerland). Speed can be regulated in 0.1 m/s steps from 0 to 2.5 m/s. This arrangement realizes a virtually infinite force-sensorized straight path. In this study, the two half-treadmills ran at the same speed. Sample

frequency for force and speed signals was set at 250 Hz. Ground reactions are synchronized in space, not only in time, with joint excursions, thus allowing to compute joint moments (see below). Joint power is computed as the product of moment and joint rotation speed. Power in itself is a quantity which is positive by definition. Yet, as it is customary in physiology, we recalled the mechanism of production by defining power as generated or positive when joint moment and rotation shared the same direction (agonist muscles were contracting while shortening, thus providing 'positive' work) and as absorbed or negative otherwise (see below).

(2) *Definitions of lesion side and spatiotemporal gait parameters*

Side of lower limbs and limb segments: Of course, the lower limbs were defined as left or right in healthy controls; in hemiplegic patients, these were defined as paretic on the side of hemiplegia (contralateral to the brain hemispheric damage) or unaffected (on the same side of the hemispheric brain damage).

Step: The step was defined as the ensemble of kinematic, dynamic, and electrophysiologic events taking place from one foot strike to the next contralateral foot strike (the sequence of two consecutive steps being named a stride).

Step time: This was defined as the time between subsequent foot–ground contacts (e.g. the paretic step beginning with the ground contact of the paretic foot).

Step length: This was defined as the sagittal distance between the lateral malleolus of the posterior and of the anterior foot, respectively, at the ground strike of the forefoot.

Stance time: For each lower limb, this was identified from vertical forces exceeding 30 N (i.e. being above random noise) (for details, see Tesio and Rota, 2008). Stance time therefore spreads over two subsequent steps.

Single stance time: For each lower limb, the portion of stance when only that limb is on ground.

Double stance time: This was identified from vertical forces of both lower limbs exceeding 30 N.

Side of the step and of the step phases: Step and stance times were assigned a side depending on which lower limb struck the ground first: for example, the paretic step and stance time began with the ground strike of the foot of the paretic leg. The step length is computed when the contralateral foot strikes the ground, and hence it is named after the posterior foot: for example, the paretic step length is computed when the contralateral unaffected foot, in the anterior position, strikes the ground. In the clinical jargon, this length is also referred to as the 'posterior step' length. The double stance side was also named after the posterior foot. By naming both the step length and the double-stance time after the posterior limb, the propulsive role of the rear lower limb in providing the muscle power needed to keep the body system in motion is consistently emphasized (Tesio *et al.*, 1998a, 1998b). In this work, the terms 'push-off' and 'double stance' phases will be used quite interchangeably. More precisely, however, 'push-off' identifies the stride phase (about from 40–60% of the stride time) where the center of mass of the body is accelerated forward, mostly thanks to the power generated by the extensor muscles of the posterior limb. 'Double stance' simply indicates that both feet are in contact with the ground. The two 'phases' may not be temporally coincident because the 'push' phase is slightly anticipated (both in onset and in offset), with respect to the double-stance phase, but the difference is minimal at low walking speed and of no relevance in this study.

(3) *Body kinematics*

Ankle, knee, and joint excursions are estimated from the three-dimensional displacements of reflective markers placed on body landmarks (Davis anthropometric model and protocol (Davis *et al.*, 1991)) and 'captured' at 250 Hz by 10 infrared stroboscopic cameras placed on the walls around the treadmill (SMART-D optoelectronic system; BTS Bioengineering Spa, Milan, Italy).

(4) *sEMG and muscle power*

The sEMG was recorded bilaterally and wirelessly from the bellies of Tibialis Anterior, Lateral Gastrocnemius, Soleus, Semitendinosus, Gluteus Maximus, Vastus Medialis, and Rectus Femoris muscles through self-adhesive transmitting electrodes (FreeEMG; BTS Bioengineering SPA). The electrodes were positioned as per the SENIAM guidelines (Hermens *et al.*, 1999). Sample frequency was set at 1 kHz. The muscle groups providing power at a given joint can be easily recognized from (a) the joint being rotated and (b) the presence of sEMG from muscles acting on that joint. From the direction of joint rotation, the muscle power was defined as generated (contracting muscles acting on the joint are shortening and generating power) or absorbed (contracting muscles are lengthening and absorbing power). The joint power ascribed to any given muscle is approximate, given that several muscles act on the same joint. The role of the same muscle, however, can be compared across different walking conditions. Here, special attention was paid to the power generated by the plantar flexors, represented by the Gastrocnemius Lateralis, during erect and CG. In fact, these are known to be the main providers of the push-off power in erect gait (Meinders *et al.*, 1998; Lipfert *et al.*, 2014). After the experimental session, the recorded sEMG signals were off-line rectified (time integral 0.08 s) and filtered (band pass filter 10–450 Hz, time constant 0.08 s).

Crouch gait

CG is common and widely studied in patients with cerebral palsy (CP). In this condition, CG is characterized by excessive flexion of the hip and the knee, and dorsal flexion of the ankle during the stance phase of gait (Steele *et al.*, 2010); a 'stiff' knee (usually kept in slightly flexed position) is present during both the stance and the swing phase (van der Krogt *et al.*, 2010). The time course of power output from lower limb muscles is highly different from normal. In particular, the average power generation from the ankle plantar flexors is decreased compared with normal values (Hoffer and Perry, 1983) and, in hemiplegic CP children, on the paretic versus the unaffected side (Olney *et al.*, 1990). The time course of the sEMG at the lower limbs, an index of their neural recruitment, is also altered. In particular, the sEMG of hip and knee extensors and of the ankle plantar flexor moments span over the entire stance phase (Frigo and Crenna, 2009), whereas in erect gait, the sEMG amplitude peaks around heel contact for the former two muscle groups, and at push-off, for the latter muscle group, respectively (Dempsey *et al.*, 2012). The continuous activity of hip and knee extensors is required to prevent the collapse of the lower limb, given that passive ligamentous stability is only available in full extension. The continuous activity of both the dorsal and the plantar flexors fosters ankle stability. This sEMG pattern recalls that observed during walking in the rear limbs of quadrupeds (Alexander and Jayes, 2009), in human infants ('toddlers') up to the age of about 5–7 years (Ganley and Powers, 2005; Ivanenko *et al.*, 2007), and in the paretic lower limb of hemiplegic patients (Frigo and Tesio, 1986; Colborne *et al.*, 1993). Healthy participants asked to walk with knee flexed can reproduce the mechanics of CG observed in CP quite accurately (Nordez *et al.*, 2009).

Testing protocol

Participants (either healthy controls or patients) wore short pants, shirt, and light gym shoes. Reflective skin markers and sEMG probes were attached to body landmarks as per the Davis and the SENIAM protocols, respectively (see above). The overall weight of the on body equipment was 180 g. From preliminary testing, it was found that a safe and comfortable speed for all hemiplegic patients on the treadmill was 0.3 m/s. Once equipped, the participants had to stand quietly for about 15 s on the treadmill for weight and muscle length calibration. Then, they were asked to stand for about 6 s with the knee flexed at about 25° while looking straight ahead at a visual target (a black dot on the front wall of the room about 2.5 m far) placed at eyes' height. The task was repeated 2–3 times, with 5–6 s rest pauses, to enable the participants to become accustomed to the crouched posture. Then they had to walk. The treadmill reached the preset speed of 0.3 m/s gradually in about 5 s. Patients could adapt to walk on the treadmill for about 30 s. Then, a 2 min sitting pause was allowed. Afterwards, patients were again accustomed to the crouched posture (knee flexed at about 25°) yet, during gait on the treadmill for 30 s. Every 10–15 s, they were asked to look at the visual target; however, looking at the treadmill belt was tolerated. After a further 2 min pause, the experimental trial began. The participant was requested to walk spontaneously for 30 s, after which the treadmill gradually stopped in 5 s. After a 2 min pause, the trial was repeated with flexed knees for 30 s. Some of the patients could not maintain the requested knee flexion along the entire 30 s trial. However, flexion ranging from about 20° to 30° was tolerated. Only one series of six subsequent strides (12 steps in about 6–8 s) was analyzed. Given these requirements, all participants could complete the experimental trials. None of them lost balance or reported any discomfort during the test.

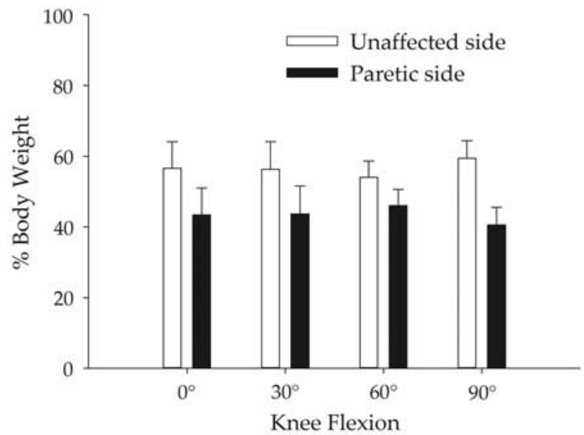
Data analysis

All signals were synchronized and off-line analyzed using a dedicated software (SMART-Suite; BTS Bioengineering SPA). Further computations and graphic representations were carried out using MATLAB (version 8; MathWorks Inc., Natick, Massachusetts, USA), STATA (version 14.0; STATA Corp., College Station, Texas, USA), and SigmaPlot (version 10.0; Systat Software Inc., San Jose, California, USA) software. Stride time was normalized to 100 time points. Results were averaged across six subsequent strides within each participant, and then grand-averaged across healthy or hemiplegic participants.

Statistics

The sample size was estimated with the aim of decreasing both within-participant and between-participant variability to an acceptable level. There are no rules of thumb in this field. Thus, the sample size was based on two previous studies. (a) In one study, the lower limb joints kinematics and dynamics were compared in six 'high-functioning' hemiplegic patients and 10 healthy controls during ground walking. A very similar variability was recorded between the two groups (Jonkers *et al.*, 2009, see their (table 2a and b). (b) Another study adopted the same experimental equipment and method of the present study, and compared results from eight patients walking on treadmill and 40 healthy controls walking on firm ground (Tesio and Rota, 2008, see their (figure 4). Again, a very similar variability was recorded, despite the difference in the sample size. This is not surprising as the treadmill imposes a known and constant speed across subsequent strides and trials, very difficult to achieve on firm ground. Beyond adopting a treadmill, the present study only recruited patients able to walk autonomously with no balance deficits, and thus *a priori* similar in the level of their walking impairment. Finally, the analysis was focused on within-participant correlations (e.g. between joints, mechanical and sEMG parameters, walking modalities), so that within-patient internal consistency, more than patients–controls differences, was the main support to conclusions. All considered, a sample of seven 'high-functioning' hemiplegic patients and 12 controls were considered to be representative of their respective populations in this specific

Fig. 1



Percent ratio of body weight loading on the lower limbs during standing (0°) and with knee flexed at 30°, 60°, and 90°. White and black columns refer to the unaffected and paretic side of seven hemiparetic patients, respectively (+SD). At all angles, the pairwise comparisons between the lower limbs were significant at $P<0.05$, with Bonferroni correction (see Methods section).

study. The peak power and the work performed at each joints were the variables of highest interest. Significance was set at P less than 0.05. A repeated analysis of variance (ANOVA) was performed on the loading on the lower limbs in stance at various knee flexion angles (and angle × side interaction). The Bonferroni correction was applied (see Fig. 1 below). For each joint (hip, knee, ankle), a repeated ANOVA model was tested, on each of the spatiotemporal, kinematic, and dynamic variables analyzed, across diagnostic classes (patients vs. controls), gait modalities (erect vs. crouch), lower limb side, and the class × modality × side interaction. For each joint, given that a change in any of the parameters tested would support an effect of the gait modality, the 'false discovery rate' correction (Benjamini and Hochberg, 1995) was adopted; this is more lenient than the more popular Bonferroni correction. The variance explanation attributable to the main or the interaction factors was computed as η^2-coefficients (Cohen, 1973) (see Tables 3–5 below).

Ethics

All participants signed an informed consent. The study was approved by the Ethic committee of the Istituto Auxologico Italiano, Milan, Italy.

Results

Participants

Demographic and clinical information on the participants enrolled is shown in Table 1. It can be observed that patients were older than controls (on average 58.0 vs. 26.7 years). However, both age groups fell into the 'adult' category considered in the walking literature. No relevant age-related differences in gait mechanics were expected: this holds by greater force if one considers the very low speed analyzed.

Safety and tolerance

No participant complained of pain or discomfort. No participant ever stumbled or fell.

Lower limb loading symmetry in standing

Figure 1 summarizes the vertical loading on either lower limb during the quiet stance and for a static progressively crouched posture: about 30°, 60°, and 90° (actually, most participants could only reach about 80°), for 20 s, as an average of seven hemiparetic patients. Only the limb side variable was significant (at $P<0.004$; $R^2=0.57$; repeated ANOVA of load across side and angle, and interaction).

It appears that (a) hemiparetic patients load predominantly on the unaffected limb; (b) the asymmetry persists at all degrees of crouch.

Gait analysis

Table 2 summarizes the spatiotemporal stride and step parameters. Speed was imposed at 0.3 m/s. During spontaneous gait, healthy participants had a step length of 0.25 (0.05) m; hemiparetic patients had a mean (SD) step length of 0.21 (0.05) m and 0.18 (0.05) m on the paretic and the unaffected side, respectively (the step of one side begins with the ground strike of the foot of that side, see Methods section). The ratio [mean (SD)] between the time duration of stance and of the whole stride was 70.28 (2.92)% in controls, and 69.91 (5.61)%

Table 1 Demographic characteristics of the study healthy adults ($n=12$) and patients ($n=7$)

	Healthy adults ($n=12$)	Hemiplegic patients ($n=7$)
Age [mean (SD)] (years)	26.7 (3.5)	58.0 (5.9)
Male (n)	2	5
Weight [mean (SD)] (kg)	64.0 (7.1)	66.7 (6.5)
Height [mean (SD)] (cm)	168.3 (8.5)	168.7 (4.0)
Dominant/paretic side (right) (n)	12	4
Onset [median (range)] (months)	1.6 (0.6–127)	
Type of lesion (n)		
Stroke, ischemic	5	
Stroke, aemorragic	1	
Surgery; low-grade glioma	1	
Side of lesion (n)		
Left	3	
Right	4	
Lesion site[a] (n)		
Total anterior circulation infarct	1	
Partial anterior circulation infarct	3	
Lacunar infarction	2	
Posterior circulation infarct	0	
Left-temporal-mesial	1	

[a]Bamford J (1992).

Table 2 **Grand-mean of spatiotemporal parameters in 12 healthy and seven hemiparetic participants (paretic and unaffected side) walking on a force treadmill at a speed of 0.3 m/s**

	Erect gait					
Spatiotemporal gait parameters	Stride time (s)	Stride length (m)	Stance time (s)	Swing time (s)	Double stance time (s)	Step length (m)
Healthy controls ($n=12$)	1.83 (0.28)	0.53 (0.10)	1.29 (0.23)	0.54 (0.06)	0.37 (0.10)	0.25 (0.05)
Hemiplegic ($n=7$)						
Unaffected side	1.52 (0.33)	0.43 (0.09)	1.17 (0.28)	0.34 (0.07)	0.38 (0.15)	0.18 (0.05)
Paretic side	–	–	1.06 (0.26)	0.45 (0.13)	0.34 (0.06)	0.21 (0.05)
Ratio paretic/unaffected (%)	–	–	90.47 (8.71)	132.96 (31.54)	96.34 (21.19)	122.75 (22.93)
	Crouch gait					
Spatiotemporal gait parameters	Stride time (s)	Stride length (m)	Stance time (s)	Swing time (s)	Double stance time (s)	Step length (m)
Healthy controls ($n=12$)	1.65 (0.41)	0.48 (0.12)	1.20 (0.34)	0.46 (0.09)	0.37 (0.14)	0.23 (0.07)
Hemiplegic ($n=7$)						
Unaffected side	1.28 (0.21)	0.38 (0.10)	0.96 (0.16)	0.31 (0.07)	0.30 (0.09)	0.14 (0.06)
Paretic side	–	–	0.92 (0.18)	0.37 (0.10)	0.30 (0.06)	0.19 (0.06)
Ratio paretic/unaffected (%)	–	–	94.91 (7.18)	120.27 (26.89)	100.62 (17.48)	148.12 (66.93)

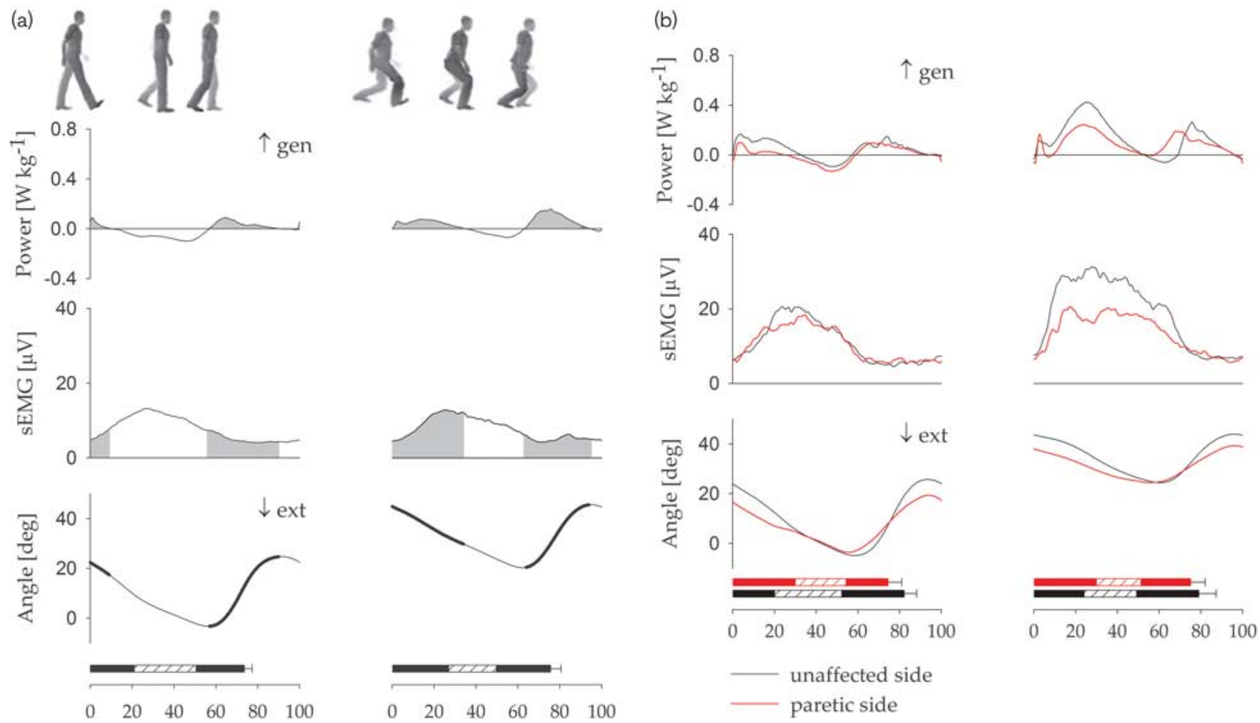
	Stride time		Stride length		Stance time		Swing time		Double stance time		Step length	
	R^2	P	R^2	P	R^2	P	R^2	P	R^2	P	R^2	P
	0.94	0.00	0.72	0.00	0.94	0.00	0.93	0.00	0.91	0.00	0.90	0.00
Repeated ANOVA modeling	P	η^2	P	η^2	P	η^2	P	η^2	P	η^2	P	η^2
Diagnostic category	0.04[a]	0.23	0.02[a]	0.27	0.02[a]	0.15	0.00[a]	0.43	0.27	0.035	0.00[a]	0.22
Side	NA	NA	NA	NA	0.66	0.01	0.12	0.07	0.64	0.01	0.26	0.04
Gait modality	0.00[a]	0.56	0.01[a]	0.1186	0.00[a]	0.45	0.00[a]	0.56	0.01[a]	0.16	0.00[a]	0.28
Interaction	0.22	0.03	0.83	0.00	0.33	0.09	0.20	0.13	0.12	0.15	0.92	0.01

Interaction = category × modality × side.
ANOVA, analysis of variance; NA, not applicable.
$P=0.05$.
[a]Significant after false discovery rate correction.

and 77.07 (2.09)% on the paretic and the unaffected side in patients, respectively. During CG, the step length was 0.23 (0.07) m in controls and 0.19 (0.06) m and 0.14 (0.06) m on the paretic and the unaffected side in patients, respectively. The ratio between the time duration of stance and of the whole stride was 71.99 (3.71)% in controls, and 71.41 (5.52)% and 76.01 (2.84)% on the paretic and the unaffected side in patients, respectively. Patients, compared with healthy controls, adopted, in both gait modalities and on both sides, a shorter step length (61–84%) and a shorter stance (76–90%), and swing (63–83%) time. In the hemiplegic group, crouch walking seemed to entail a lower asymmetry between the two lower limbs compared with erect gait. The % ratio between the double stance times (paretic vs. unaffected posterior foot) was 96.34 (21.19)% during erect gait and 100.62 (17.48)% during CG. Also, the % ratio between the swing time of the paretic and the unaffected lower limb was 132.96 (31.54)% during erect gait and 120.27 (26.89)% during CG. Table 2 provides the results of ANOVA modeling. These confirm the significance of the differences found across diagnostic categories (patients vs. controls) and across gait modalities (erect vs. crouch), but not the effect of crouch in decreasing the paretic/affected asymmetry of step length and stance time. Table 2 (bottom panel) shows the results of the ANOVA modeling (see Methods section) of each parameter (excluding the minimum–maximum joint angles), across the diagnostic category (patient vs. control), the lower limb or step side, the gait modality (erect vs. crouch), and their interaction.

Figure 2a refers to hip motion during erect (left column) and crouch (right column) gait in healthy controls. The human silhouette on top sketches the double and single stance phases of the analyzed lower limb (in black). Hip power (generated or positive upward), the Gluteus Maximus sEMG signal, and the sagittal excursions (extension downward) are given, from top to bottom, as a function of the standardized stride time, given on the abscissa. The horizontal bar marks the single (dashed) and the double (filled) stance times. The shaded areas under the power and sEMG curves highlight the times when power is generated, taken into account for computations in Tables 3–5. It can be seen that CG mostly entails a higher power output from the hip extensor muscles in the single stance phase (heretofore: ss phase) and during the late double stance (push-off, heretofore:

Fig. 2



(a) The curves show the grand-mean of six subsequent strides performed by 12 healthy adults during erect (left panels) and crouch gait (right panels) on a force treadmill, right step first, at a speed of 0.3 m/s. On the abscissa, the standardized stride time is given. The bottom horizontal bar shows the stance time (average + SD between the two lower limbs). Dashed and filled segments represent the average time of the single and the double stance phases, respectively. From top to bottom, the curves show the power (in W/kg), the surface electromyography (sEMG) from the extensor muscles (gluteus maximus), and the sagittal joint excursion (hip extension downward, see 'ext' arrow). The shaded areas and the thickened lines mark the stride phases when power is generated or positive (see 'gen' arrow and Table 3). (b) Replication of the information for an average of seven hemiplegic patients. The red and black colors refer to the paretic and the unaffected lower limb, respectively.

po phase). The sEMG signals are not increased while power is being generated during the po phase, so that the power output can be ascribed to a more advantageous muscle length because of the hip flexed posture (see legend for further details).

Figure 2b replicates the information shown in Fig. 2a for the sample of seven hemiparetic patients. The red and black colors are assigned to the paretic and the unaffected lower limb, respectively. It can be seen that, compared with healthy controls, shifting from erect to CG entails a higher increase in power output from the hip, prevailing for the unaffected lower limb and consistent with the increased level of the sEMG signal (see black curves in the right column, power and sEMG tracings).

Figures 3 and 4 replicate the information shown in Fig. 2 with respect to the knee and the ankle joint, respectively. The shaded areas highlighting the times of power generation are no longer provided and yet, these times are the ones considered for the computations presented in Table 4 (knee) and Table 5 (ankle), respectively).

Table 3 provides the numerical counterpart of Fig. 2a and b. For the hip motion, control and hemiparetic values, both during erect and CG, can be compared along the respective rows, across the columns showing eight dynamic, sEMG, and kinematic parameters. It is noteworthy that for joint excursion and rotation speed, distinct values are provided for the ss and the po phases. Tables 4 and 5, replicate the information shown in Table 3 for the knee and the ankle, respectively. The pattern of knee flexion-extension just after foot strike (so-called knee 'yielding', Fig. 4) is known to be highly variable across individuals, depending on their more or less 'compliant' walking style (Frigo and Tesio, 1986). Here, a very low positive work is provided by the knee in controls, or by the unaffected knee in patients, in erect gait, only. For these reasons, this 'early stance' phase will be neglected in inferential analyses on knee power and work. The overall picture can be summarized as follows: as a rule, patients provided a higher muscular work (median: 137%, range: 77–250%) paralleled by a greater sEMG area (median: 174%, range: 75–185%) compared with controls. In erect gait, the generation of peak extensor power across hip, knee, and ankle joints was in general lower (83–90%) from the paretic limb, and higher (98–165%) from the unaffected limb, compared with control values. In CG, peak power generation across the three lower limb joints was invariably higher in

Table 3 Hip motion

Hip	Erect gait								
	Peak power (W/kg)	Work (J/kg)	sEMG area (μV s)	Joint excursion (ss) (deg.)	Joint excursion (po) (deg.)	Joint rotation range (minimum–maximum) (ss) (deg.)	Joint rotation range (minimum–maximum) (po) (deg.)	Rotation speed (ss) (deg./s)	Rotation speed (po) (deg./s)
Healthy controls ($n=12$)	0.17 (0.06)	2.28 (1.04)	278.63 (213.93)	7.87 (5.81)	24.93 (9.71)	14.54–22.411	−2.51 to 22.61	29.06 (9.59)	48.88 (9.08)
Hemiplegic ($n=7$)									
Unaffected side	0.28 (0.05)	6.46 (1.75)	698.75 (336.32)	20.37 (12.81)	32.67 (5.19)	4.20–24.57	−4.22 to 27.73	35.29 (10.71)	59.29 (11.57)
Paretic side	0.15 (0.08)	3.56 (1.47)	484.23 (232.49)	11.15 (10.03)	22.00 (10.41)	5.63–16.78	−0.51 to 18.66	23.77 (8.39)	47.85 (21.22)
Ratio paretic/ unaffected (%)	55.30 (32.27)	55.87 (19.18)	68.22 (22.12)	61.59 (19.20)	67.15 (38.10)	–	–	85.03 (31.42)	78.24 (71.67)

Hip	Crouch gait								
	Peak power (W/kg)	Work (J/kg)	sEMG area (μV s)	Joint excursion (ss) (deg.)	Joint excursion (po) (deg.)	Joint rotation range (minimum–maximum) (ss) (deg.)	Joint rotation range (minimum–maximum) (po) (deg.)	Rotation speed (ss) (deg./s)	Rotation speed (po) (deg./s)
Healthy controls ($n=12$)	0.22 (0.09)	5.94 (3.78)	524.50 (424.49)	22.05 (9.60)	21.62 (10.01)	27.82–43.88	17.74–45.26	39.33 (16.49)	39.33 (16.49)
Hemiplegic ($n=7$)									
Unaffected side	0.51 (0.26)	14.88 (4.22)	1494.87 (789.84)	18.50 (6.48)	17.53 (6.28)	25.46–43.96	27.09–44.61	31.12 (11.74)	58.10 (12.02)
Paretic side	0.39 (0.22)	10.88 (4.96)	968.82 (625.90)	13.00 (7.05)	15.59 (5.53)	28.13–41.13	24.27–39.86	27.82 (12.06)	38.22 (7.31)
Ratio paretic/ unaffected (%)	85.08 (70.06)	71.76 (23.60)	91.37 (36.26)	61.88 (30.56)	93.49 (35.48)	–	–	71.67 (19.13)	71.67 (19.13)

	Peak power		Work		sEMG area		Joint excursion (ss)		Joint excursion (po)		Rotation speed (ss)		Rotation speed (po)	
	R^2	P	R^2	P	R^2	P	R^2	P	R^2	P	R^2	P	R^2	P
	0.83	0.00	0.89	0.00	0.89	0.00	0.90	0.00	0.74	0.04	0.72	0.05	0.86	0.00
Repeated ANOVA modeling	P	η^2	P	η^2	P	η^2	P	η^2	P	η^2	P	η^2	P	η^2
Diagnostic category	0.00[a]	0.37	0.00[a]	0.50	0.00[a]	0.33	0.61	0.01	0.59	0.01	0.08	0.10	0.11	0.07
Side	0.05	0.11	0.05	0.11	0.08	0.10	0.62	0.01	0.51	0.01	0.83	0.00	0.78	0.00
Gait modality	0.00[a]	0.44	0.00[a]	0.74	0.00[a]	0.52	0.00[a]	0.49	0.00[a]	0.32	0.20	0.06	0.03[a]	0.16
Interaction	0.04	0.24	0.04	0.25	0.69	0.06	0.00[a]	0.64	0.17	0.16	0.00[a]	0.41	0.01[a]	0.36

Dynamic, sEMG from the Gluteus maximus, and kinematic parameters (second row from top) during erect gait (upper section of the table) and crouch gait (lower section).
Only grand-means across strides and participants are reported. From top to bottom, the data rows refer to healthy controls, the unaffected and the paretic lower limb of the hemiplegic patients, and the paretic/unaffected % ratio, respectively. From left to right, each data column refers to the maximum power of the hip joint (peak power, W/kg; generated during extension), work (J/kg; or, the integral of power over time), the area under the sEMG curve (sEMG area, μV s), the hip joint excursion over time (joint excursion ss and po phases, deg.), and the average hip rotation speed (rotation speed ss and o phases, deg./s). Power, work, and sEMG are computed during the intervals in which extensor power is generated, as shown in Fig. 2a and b.
Interaction = category × modality × side.
ANOVA, analysis of variance; sEMG, surface electromyography.
$P=0.05$.
[a]Significant after false discovery rate correction.

Table 4 **Knee motion**

Knee	Erect gait: Peak power (W/kg)	Work (J/kg)	sEMG area (μV s)	Joint excursion (ss) (deg.)	Joint excursion (po) (deg.)	Joint rotation range (minimum–maximum) (ss) (deg.)	Joint rotation range (minimum–maximum) (po) (deg.)	Rotation speed (ss) (deg./s)	Rotation speed (po) (deg./s)
Healthy controls ($n=12$)	0.20 (0.15)	2.84 (2.28)	225.94 (150.16)	4.70 (4.43)	19.36 (16.44)	5.41–10.12	−0.79–18.57	11.76 (7.83)	51.73 (30.99)
Hemiplegic ($n=7$)									
Unaffected side	0.20 (0.14)	5.15 (2.11)	649.34 (296.90)	9.17 (6.32)	16.87 (13.74)	13.38–22.55	13.30–30.17	25.71 (19.38)	52.85 (18.43)
Paretic side	0.18 (0.08)	3.24 (2.46)	537.30 (490.96)	14.09 (11.52)	–	7.16–20.74	–	29.26 (17.20)	–
Ratio paretic/ unaffected (%)	76.47 (41.05)	61.51 (28.90)	71.24 (66.29)	116.00 (38.14)	–	–	–	101.90 (42.71)	–

Knee	Crouch gait: Peak power (W/kg)	Work (J/kg)	sEMG area (μV s)	Joint excursion (ss) (deg.)	Joint excursion (po) (deg.)	Joint rotation range (minimum–maximum) (ss) (deg.)	Joint rotation range (minimum–maximum) (po) (deg.)	Rotation speed (ss) (deg./s)	Rotation speed (po) (deg./s)
Healthy controls ($n=12$)	0.28 (0.16)	5.93 (3.78)	930.42 (606.02)	7.95 (5.93)	–	36.35–44.30	–	13.29 (7.37)	–
Hemiplegic ($n=7$)									
Unaffected side	0.32 (0.25)	5.53 (5.01)	1042.61 (679.57)	5.33 (6.04)	–	43.13–48.14	–	14.23 (18.68)	–
Paretic side	0.30 (0.17)	4.58 (3.20)	701.87 (300.21)	5.29 (2.47)	–	40.11–45.26	–	17.18 (8.75)	–
Ratio paretic/ unaffected (%)	82.64 (33.45)	60.39 (20.46)	74.32 (37.38)	66.40 (47.01)	–	–	–	113.56 (92.92)	–

	Peak power		Work		sEMG area		Joint excursion (ss)		Rotation speed (ss)	
	R^2	P	R^2	P	R^2	P	R^2	P	R^2	P
	0.82	0.00	0.78	0.00	0.77	0.01	0.68	0.12	0.73	0.06
Repeated ANOVA modeling	P	η^2	P	η^2	P	η^2	P	η^2	p	η^2
Diagnostic category	0.78	0.00	0.75	0.00	0.04	0.12	0.14	0.06	0.00[a]	0.22
Side	0.08	0.09	0.07	0.09	0.40	0.02	0.51	0.01	0.83	0.00
Gait modality	0.01[a]	0.20	0.01[a]	0.14	0.00[a]	0.45	0.13	0.08	0.01[a]	0.20
Interaction	0.11	0.18	0.02[a]	0.28	0.25	0.13	0.01[a]	0.31	0.03[a]	0.28

Dynamic, sEMG from the Rectus femoris and kinematic parameters during erect gait and crouch gait.
Power is given as generated during knee extension. Other indications as in Table 3.
Interaction = category × modality × side.
ANOVA, analysis of variance; sEMG, surface electromyography.
$P=0.05$.
[a]Significant after false discovery rate correction.

Table 5 **Ankle motion**

	Erect gait					
Ankle	Peak power (W/kg)	Work (J/kg)	sEMG area (μV s)	Joint excursion (deg.)	Joint rotation range (minimum–maximum) (deg.)	Rotation speed (deg./s)
Healthy controls ($n=12$)	0.35 (0.18)	4.25 (2.96)	367.44 (189.36)	14.23 (8.42)	1.93–13.86	29.58 (10.77)
Hemiplegic ($n=7$)						
Unaffected side	0.48 (0.24)	5.27 (1.67)	893.02 (652.86)	17.05 (11.25)	0.39–13.84	35.39 (16.46)
Paretic side	0.29 (0.12)	3.76 (1.89)	531.70 (191.11)	12.34 (4.26)	1.76–12.37	26.53 (10.20)
Ratio paretic/unaffected (%)	72.59 (41.09)	74.89 (44.05)	95.71 (77.62)	68.89 (36.23)	–	81.59 (35.75)
	Crouch gait					
Ankle	Peak power (W/kg)	Work (J/kg)	sEMG area (μV s)	Joint excursion (deg.)	Joint rotation range (minimum–maximum) (deg.)	Rotation speed (deg./s)
Healthy controls ($n=12$)	0.65 (0.27)	7.20 (3.24)	550.33 (334.50)	21.13 (7.28)	7.15–26.22	44.77 (13.99)
Hemiplegic ($n=7$)						
Unaffected side	0.94 (0.32)	9.89 (3.44)	1074.68 (572.75)	23.51 (9.17)	6.87–26.11	51.82 (14.55)
Paretic side	0.64 (0.28)	6.93 (2.97)	671.86 (251.96)	14.29 (4.43)	9.21–23.16	49.04 (19.17)
Ratio paretic/unaffected (%)	74.67 (40.55)	77.82 (42.07)	67.23 (64.30)	68.81 (32.27)	–	94.72 (54.30)

	Peak power		Work		sEMG area		Joint excursion		Rotation speed	
	R^2	P	R^2	P	R^2	P	R^2	P	R^2	P
	0.81	0.00	0.78	0.00	0.80	0.00	0.93	0.00	0.85	0.00
Repeated ANOVA modeling	P	η^2	P	η^2	P	η^2	P	η^2	P	η^2
Diagnostic category	0.20	0.05	0.46	0.02	0.00[a]	0.25	0.79	0.00	0.39	0.02
Side	0.05	0.11	0.30	0.03	0.10	0.08	0.30	0.03	0.16	0.06
Gait modality	0.00[a]	0.62	0.00[a]	0.54	0.02[a]	0.19	0.00[a]	0.56	0.00[a]	0.65
Interaction	0.48	0.07	0.72	0.04	0.99	0.00	0.16	0.15	0.39	0.09

Dynamic, sEMG from the lateral Gastrocnemius and kinematic parameters during erect and crouch gait.
Power is given as generated during plantar flexion. Other indications as in Table 3.
Interaction = category × modality × side.
ANOVA, analysis of variance; sEMG, surface electromyography.
$P=0.05$.
[a]Significant after false discovery rate correction.

hemiparetic patients compared with controls: 107–177% from the paretic limb and 114–231% from the unaffected limb. When gait shifted from erect to crouch, in hemiplegic patients at the hip, only, the paretic/unaffected ratio increased significantly. For peak power, work, sEMG area, and joint rotation, the paretic/unaffected ratio increased from 55 to 85%, 56 to 72%, 68 to 91%, and 67 to 93%, respectively. Tables 3–5 also confirm that the plantar flexors provide a higher positive power compared with the hip and knee. In both patients and controls, and erect and CG, the plantar flexors generated peak power and work about two times higher compared with the hip or the knee extensors.

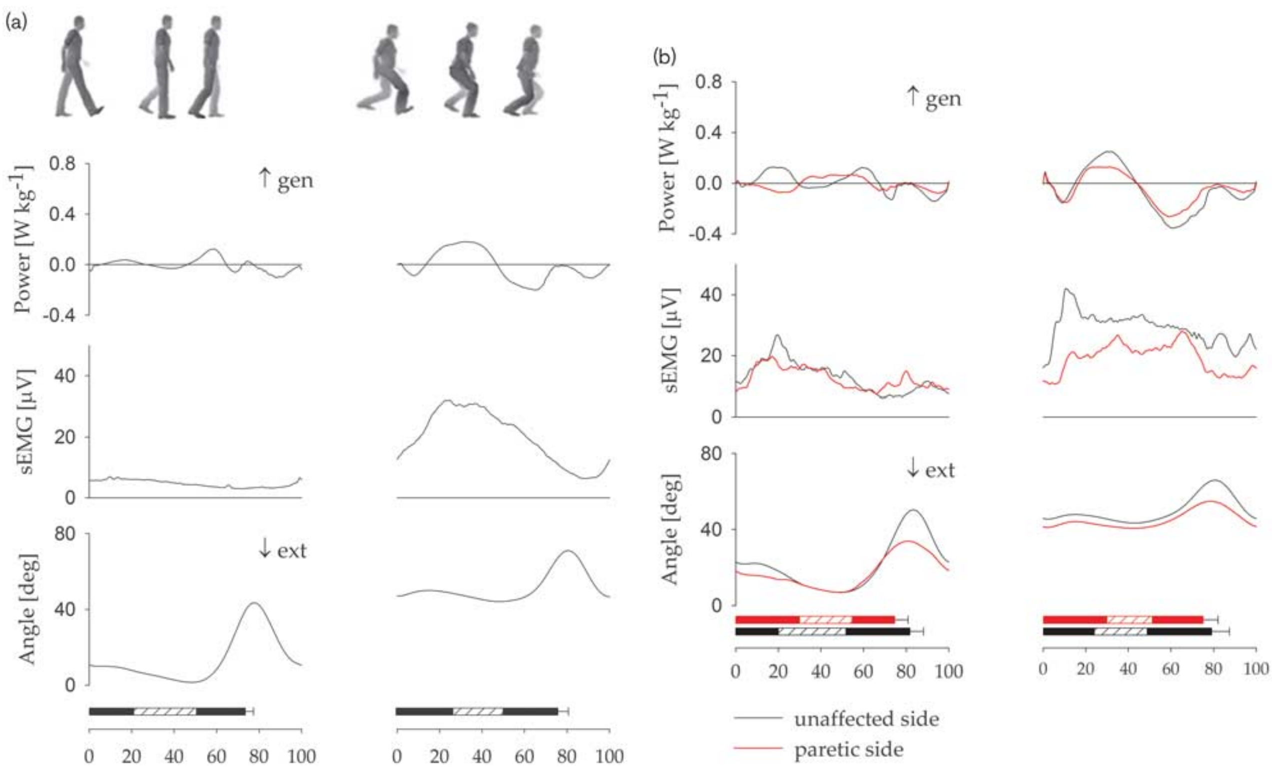
For hemiplegic patients, the paretic/unaffected ratios of dynamic and kinematic variables are graphically highlighted in Fig. 5. White and black bars refer to erect and CG, respectively. From top to bottom, panels refer to the hip, knee, and ankle joint, respectively.

It can be observed that CG consistently entails an increased paretic/unaffected ratio for peak power, work, and sEMG only for the hip joint.

Discussion

There are three main limitations of this study. First, the small size and the low severity of the sample of hemiplegic patients must be noted. The reasons for considering this sample size sufficient are given in the Methods section. Changes related to the crouched compared with the erect posture, asymmetries between the paretic and the unaffected step, and crouch-related changes in asymmetry could be detected anyway. Even greater asymmetries can be expected in patients more impaired than those recruited in this study. Second, treadmill walking may not be fully representative of ground walking. Differences in kinematic, dynamic, and sEMG parameters, however, have been shown to be minimal at low and intermediate walking speeds (Tesio and Rota, 2008). Finally, the speed imposed was very low (0.3 m/s). This ensured that all patients could comfortably adapt to the treadmill condition, mostly in the tiring crouch position. In addition, this speed is the one adopted on firm ground by many hemiplegic patients. It is true that the results presented here cannot be safely generalized to higher speeds. However, an even greater recruitment of lower limb muscles, with an even greater

Fig. 3



The panels replicate the information shown in (a) healthy controls and (b) hemiplegic patients. The variables refer to the knee joint and the surface electromyography (sEMG) signal from the rectus femoris.

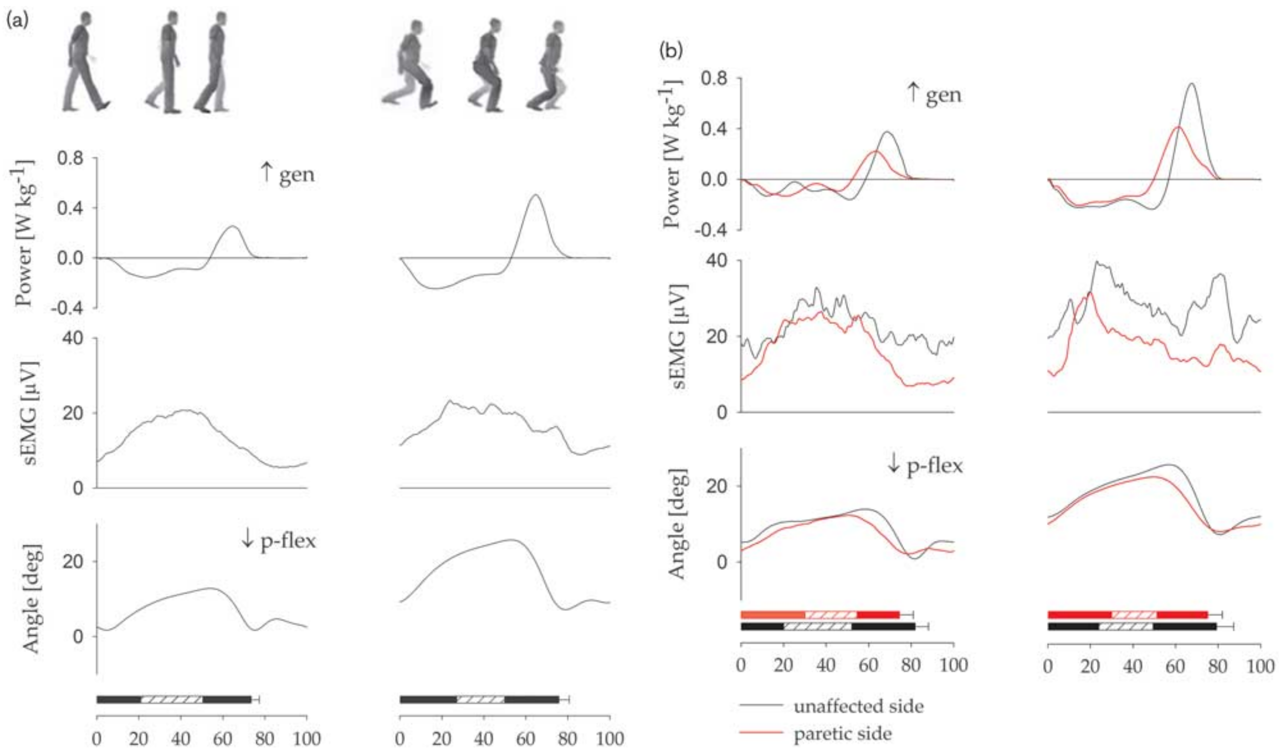
difference between erect and CG, can be expected at higher speeds (see above).

The results need to be discussed with respect to each of the two hypotheses generating the present study.

Hypothesis 1: crouch gait is an effective form of 'forced-use' exercise for the paretic lower limb

The results support the notion that both in healthy and hemiplegic participants the plantar flexors of the posterior limb are the main providers of the power required to keep the body system in motion during walking and that most of this power is exerted against the ground by the rear plantar flexor muscles during the push-off phase. This is consistent with the evidence that the energy-saving pendulum-like mechanism of gait is discontinued during double stance, when a strong injection of muscular power is requested to maintain the body in motion (Cavagna and Kaneko, 1977; Tesio *et al.*, 1998a, 2011). The increased power associated with CG cannot be entirely ascribed to the mechanical advantage provided by the greater length of muscle extensors entailed by the more flexed joint positions. In fact, the parallel increase of the sEMG areas shows an increase in the neural drive. In addition, the latter was probably underestimated. The sEMG areas were computed during the provision of generated ('positive') muscle power only. Yet, the time of electric activity anticipates the time of detectable force because of the time needed for impulse conduction, excitation–contraction coupling, alignment of pennate fibers with the tendon, and stiffening of the series elastic elements (Nordez *et al.*, 2009), so that the overall electromechanical delay for force onset was presumably around 50–80 ms in this study. The delay is known to be higher for contraction than for relaxation (Esposito *et al.*, 2016). Also, as a rule, the sEMG signal decreased monotonically from just before to just after the time of detectable positive power (Figs 2–4), so that the missed sEMG was of non-negligible amplitude. The crouch-related increase in joint power, work, and sEMG area was also observed in hemiplegic patients. In both gait modalities, the joints of the unaffected limb provided power and work higher than normal, whereas power and work were equal to or lower than normal in the paretic side. This is consistent with the finding that in asymmetric gait impairments such as those following stroke (Cavagna *et al.*, 1983), unilateral hip replacement (Tesio *et al.*, 1985), above-knee or below-knee amputation (Tesio *et al.*, 1998b), and knee rotationplasty (Rota *et al.*, 2016), the muscular power and work sustaining the motion of the body center of gravity with respect to the ground ('external' work and power) at push-off are about two to

Fig. 4

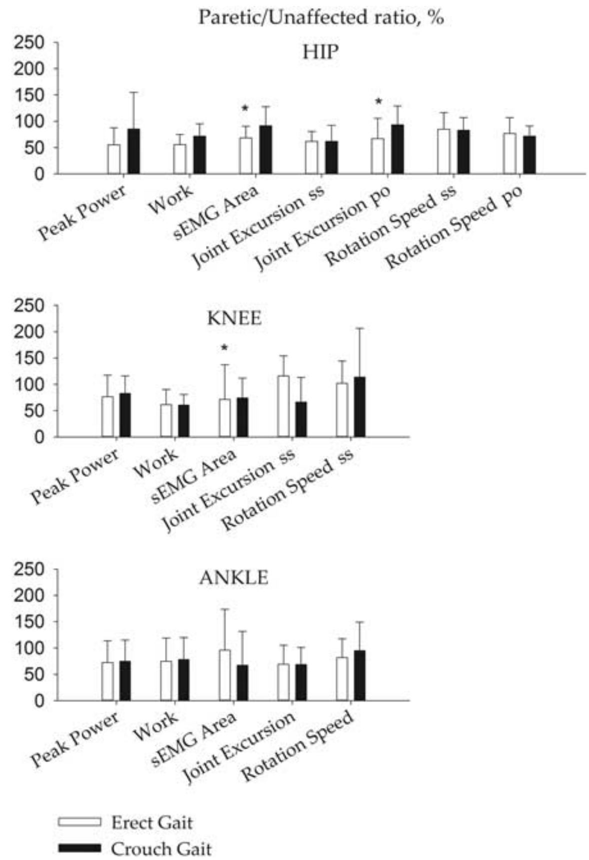


The panels replicate the information shown in (a) healthy controls and (b) hemiplegic patients. The variables refer to the ankle joint and the surface electromyography (sEMG) signal from the lateral Gastrocnemius.

five times lower when the posterior limb is the paretic one compared with the next unaffected push-off. The dynamic asymmetry during hemiplegic gait thus replicates the one observed between erect and crouched stance (Fig. 1) and it is reflected by the asymmetry in single and double stance time (both lower on the paretic, compared with the unaffected, side).

To sum up, the results are in agreement with the first hypothesis put forward in this study by showing that on the paretic side, the muscles of the lower limb can be actually 'awakened' by simply imposing a crouched posture that would make gait more difficult without their enhanced intervention. In fact, during CG, both the unaffected and the paretic lower limb are forced to be much more active to prevent their own collapse. In theory, isometric contractions would be sufficient to maintain the flexed posture of its lower limb joints during CG, thus allowing the body to 'pole-vault' passively over a flexed, yet stiffer, paretic limb. This was perhaps the case for hip and knee joints, showing a decreased excursion (Figs 2 and 3), but not for the ankle joint (Fig. 4), the excursion of which increased. In all cases, power and work were higher in crouched, compared with erect, gait. This seems to support the hypothesis that in hemiparetic patients, the forced recruitment of otherwise paretic muscles reflects disinhibition of a survived mechanism of neural control of gait, consistent with the conceptual framework of LNU. Whichever this mechanism, the results show that CG might be classified among the 'forced-use' exercises for the lower limb, as defined in a previous work (Tesio, 2001), although no restraint is imposed to the unaffected lower limb. Other gait modalities can presumably obtain an increased power output from the paretic lower limb. Dynamic results are available from studies where patients were asked for both a self-selected (0.73 m/s) and a maximal speed (1.26 m/s) (Milot *et al.*, 2007), and EMG results are available from a study on walking uphill (Werner *et al.*, 2007). The former study evidenced that the 'muscular utilization ratio' (the peak joint moment provided during push-off, compared with a maximal isokinetic moment) is increased at maximal, compared with spontaneous speed, more in hip than in calf muscles. In this study, the plantar flexor muscular utilization ratio on the paretic side increased on average by 13.1% at maximal, compared with self-selected, speed, that is much less than the two-fold increase in power caused by CG, compared with erect gait, despite the much lower speed adopted in the present study (0.30 vs. 0.73–1.26 m/s). However, the 'maximal' speed required to the hemiplegic patients in the above-cited study approached the one optimizing the pendulum-like recovery of mechanical energy (Cavagna and Kaneko,

Fig. 5



From top to bottom, the panels refer to (a) hip, (b) knee, and (c) ankle joints, respectively. The ordinate shows the % ratio of the mean values recorded from the paretic and the unaffected side, shown in Tables 3–5. White and black bars refer to erect and crouch gait, respectively. In all panels, each pair of bars refers to one of the parameters shown in the abscissa [peak power, work, surface electromyography (sEMG) area, joint excursion, and rotation speed, all computed while positive power is generated]. Asterisks mark the significant pairwise comparisons ($P<0.05$, Bonferroni correction for multiplicity; see Methods section).

1977); by contrast, shifting from erect to CG decreases markedly the efficiency of the pendulum (Wang *et al.*, 2003), thus imposing a much higher muscular work. With respect to walking uphill, it is noteworthy that no changes were observed in the sEMG output compared with level walking at the same, self-selected, speed so that the higher muscle work in that study can be ascribed to the mechanical advantage provided by a higher muscle length, not necessarily to a higher neural drive. Therefore, CG, even at the low speeds adopted in this work, seems to be the most effective way to elicit a higher neural drive toward the paretic lower limb muscles, thus contrasting their learned nonuse. Another attractive, yet entirely speculative, explanation for the strong effect of CG on recruitment of the muscles of the lower limb focuses on the considerable neurologic differences between erect and CG. The former is highly specific for the human adults. Their gait has many unique features across both quadruped and biped walking vertebrates. In particular, it implies full extension of the hip and knee (allowing passive stability through ligaments, with sparing of muscle contractions) and a heel-toes sequence during foot–ground contact, allowing for a high plantar flexor work. The kinematic sequence provides the calf muscles with the possibility for a wide stretch-shortening cycle fostering (the more, the faster the cycle) their positive power output (Lai *et al.*, 1985). By contrast, most parameters of CG replicate those of walking vertebrates (Alexander and Jayes, 2009), human infants up to the age of 5–7 years (Ganley and Powers, 2005; Ivanenko *et al.*, 2007), hemiplegic patients on their paretic side (see above; Brandstater *et al.*, 1983; Colborne *et al.*, 1993), and ancestors of the *Homo sapiens* (Wang *et al.*, 2003). Thanks to coordination at subcortical neural centers, this gait pattern might 'emerge' unaffected from the 'dissolution of the nervous system' caused by hemispheric stroke in adult men, an old established, jacksonian concept (Franz and Gillett, 2011). Of course, it remains doubtful whether this result can be transferred to spontaneous gait. A peculiarity of CG in healthy and hemiparetic individuals (on both sides), however, must be highlighted, that is the increase in the neural drive to the plantar flexors, contrary to what is observed in CP, at push-off. This suggests that CG in human adults calls into action a more mature pattern of recruitment, which speaks in favor of its capacity to 'awake' calf recruitment in erect gait, too.

Hypothesis 2: crouch gait entails a higher symmetry between the motion of the paretic and the unaffected lower limb, compared with erect gait

This hypothesis is not confirmed with respect to spatio-temporal parameters (Table 2, bottom row). By contrast, the hypothesis is confirmed for the peak power, positive work, joint excursion, and rotation speed during positive work at the hip joint (Table 3, bottom row). At the knee and, most importantly, at the ankle, the paretic/unaffected ratios of work and power do not change significantly between erect and CG (Tables 4 and 5, bottom rows). In this respect, therefore, CG is partly disappointing as it does not 'awaken' selectively, on the paretic side, the main engines of normal gait, that is the calf muscles.

Its effectiveness as a force-use procedure, in any case, justifies its application as a form of rehabilitation exercise to be tested in dedicated clinical studies. The crouched posture is uncomfortable and it may overload the hip and knee joints. A point to be clarified, therefore, concerns the patients' tolerance to longer and/or more numerous walks compared with the ones adopted in this study.

Acknowledgements

Conflicts of interest

There are no conflicts of interests.

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