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Effects of a knee–ankle–foot orthosis on gait biomechanical characteristics of paretic and non-paretic limbs in hemiplegic patients with genu recurvatum

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ABSTRACT

Background: A knee–ankle–foot orthosis may be prescribed for the prevention of genu recurvatum during the stance phase of gait. It allows also to limit abnormal plantarflexion during swing phase. The aim is to improve gait in hemiplegic patients and to prevent articular degeneration of the knee. However, the effects of knee–ankle–foot orthosis on both the paretic and non-paretic limbs during gait have not been evaluated. The aim of this study was to quantify biomechanical adaptations induced by wearing a knee–ankle–foot orthosis, on the paretic and non-paretic limbs of hemiplegic patients during gait.
Methods: Eleven hemiplegic patients with genu recurvatum performed two gait analyses (without and with the knee–ankle–foot orthosis). Spatio-temporal, kinematic and kinetic gait parameters of both lower limbs were quantified using an instrumented gait analysis system during the stance and swing phases of the gait cycle.
Findings: The knee–ankle–foot orthosis improved spatio-temporal gait parameters. During stance phase on the paretic side, knee hyperextension was reduced and ankle plantarflexion and hip flexion were increased. During swing phase, ankle dorsiflexion increased in the paretic limb and knee extension increased in the non-paretic limb. The paretic limb knee flexion moment also decreased.
Interpretation: Wearing a knee–ankle–foot orthosis improved gait parameters in hemiplegic patients with genu recurvatum. It increased gait velocity, by improving cadence, stride length and non-paretic step length. These spatiotemporal adaptations seem mainly due to the decrease in knee hyperextension during stance phase and to the increase in paretic limb ankle dorsiflexion during both phases of the gait cycle.

1. Introduction

About one half of stroke survivors present with motor impairments such as muscle weakness, abnormal muscle tone and sensory impairments, often combined with spasticity or contractures of the paretic lower limb muscles. About 52 to 85% of hemiplegic patients regain the capacity to walk, however, their gait differs from that of healthy subjects (Bohannon, 1987; Eng and Chu, 2002). Hemiplegic gait is characterized by alterations in spatio-temporal and kinematic parameters. For example, during swing phase, peak knee flexion is frequently reduced (stiff knee gait) (Hutin et al., 2010; Robertson et al., 2009; Stoquart et al., 2008) with or without a reduction in ankle dorsiflexion (equinus or equino-varus) (Mancini et al., 2005; Pittock et al., 2003; Pradon et al., 2011). In stance phase a “dynamic” knee hyperextension, known as genu recurvatum often occurs (Goldie et al., 1996; Perry, 1992; Von Schroeder et al., 1995).

Genu recurvatum has been reported in approximately one half of hemiplegic patients (whatever the origin of the hemiplegia: stroke or traumatic brain injury) (Hogue and McCandless, 1983; Morris et al., 1992). From a biomechanical point of view, this kinematic impairment is characterized by a ground reaction force vector that passes well in front of the knee. This phenomenon generates a knee extensor moment to prevent collapse during stance phase. This symptom is frequently associated with knee pain and therefore limits the patient's autonomy in daily life activities. Genu recurvatum may be caused by several phenomena: i) spasticity or weakness of the quadriceps muscle, ii) spasticity and/or contracture of the triceps surae and iii) impaired proprioception (Inaba, 1967; Perry, 1992).

Different types of treatment exist for genu recurvatum in hemiplegic patients, such as orthotic devices (Farncombe, 1980; Morinaka et al., 1982, 1984), rehabilitation techniques (Basaglia et al., 1989; Morris et al., 1992; Stanic et al., 1978) or surgical interventions (Caillet et al., 1998). Ankle–foot orthoses (AFOs) have been shown to be effective for the treatment of genu recurvatum in stroke patients when the main cause is spasticity or contracture of the triceps surae muscle (Fatone et

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al., 2009). However, in patients with severe knee recurvatum, caused by spasticity or weakness of the knee extensor muscles, the AFO may not be effective (Bleyenheuft et al., 2010). In this case, a knee–ankle–foot orthosis (KAFO) may be appropriate (Bleyenheuft et al., 2010). This type of orthosis is composed of several segments which act on both the knee and the ankle joint. KAFOs were initially developed to counteract quadriceps weakness in patients with poliomyelitis (Genêt et al., 2010). Indeed, this type of orthosis mechanically reduces genu recurvatum by preventing knee hyper-extension during the stance phase of the gait cycle through a stop in the metal joint, without interfering with knee flexion during swing phase (Farncombe, 1980; Morinaka et al., 1982, 1984). They also limit abnormal degrees of plantarflexion during swing phase (Yakimovich et al., 2006). Although KAFOs have been widely prescribed for patients with poliomyelitis (Genêt et al., 2010), they are more rarely prescribed to reduce genu recurvatum in stroke patients. The main reason is that there is a school of thought that practicing gait with a KAFO could delay the recovery of normal movement and increase spasticity (Davidson and Waters, 2000; Sackley and Lincoln, 1996). Moreover, such orthoses are difficult to don and may be considered unaesthetic by the patients.

Three studies have assessed the effects of KAFOs on spatio-temporal gait parameters in adult hemiplegic patients (Farncombe, 1980; Morinaka et al., 1982, 1984). The results showed that by reducing genu recurvatum during stance phase, the KAFO led to a more effective gait with an improvement in gait velocity. However, to our knowledge, the mechanism(s) underlying these gait improvements has not been evaluated. It is important to improve understanding of kinematic and kinetic adaptations of both limbs which occur when wearing a KAFO, in order to justify (or not) prescription for stroke patients.

The aims of this study were therefore i) to evaluate the effectiveness of a KAFO on gait in hemiplegic patients with genu recurvatum due to spasticity or weakness of the quadriceps and ii) to quantify the biomechanical adaptations induced by the KAFO on both lower limbs. To that end, we assessed the modifications of spatio-temporal, kinematic and kinetic gait parameters induced by a KAFO in hemiplegic patients using a three dimensional (3D) gait analysis system. The angle of knee extension during stance phase was chosen as the primary outcome measure to quantify the effectiveness of the KAFO. We hypothesized that the KAFO would decrease knee hyperextension of the paretic limb during stance phase. In addition, we hypothesized that this alteration in paretic knee joint kinematics would induce modifications of the kinematic and kinetic parameters of the non-paretic limb.

2. Methods

2.1. Subjects

Eleven chronic hemiplegic subjects (7 males and 4 females, Table 1) volunteered to participate in this study (age: 51 (15) years; height: 168 (10) cm, mass: 70 (15) kg). Inclusion criteria include the following: over 18 years old, hemiplegia following a stroke occurring more than 6 months prior to study participation (chronic-phase), with knee hyperextension during stance phase of gait (−17 (11)°; range: −1.6 to −35.6°), ability to walk 10 m without walking aids, and prescription of a carbon KAFO in the last 6 months which has been worn daily for at least one month before inclusion. All the patients exhibited either spasticity or weakness of quadriceps. Because all patients exhibited several potential etiologies for the genu recurvatum, a clinical examination was used to determine the main cause: spasticity of quadriceps (n = 6), weakness of quadriceps (n = 2) and spasticity of triceps surae (n = 3). This study was approved by the local ethics committee and all subjects provided written informed consent prior to participation in any study-specific procedures.

Table 1
Subject demographics.

Patient	Gender	Age (years)	Mass (kg)	Height (cm)	Paretic limb	Time since hemiplegia (months)
1	M	62	85	181	Left	372
2	F	55	86	166	Left	96
3	M	48	72	167	Right	84
4	F	75	48	151	Right	156
5	F	67	51	169	Right	672
6	M	45	78	168	Right	72
7	F	55	55	165	Left	384
8	M	21	72	167	Right	72
9	M	58	86	172	Right	207
10	M	39	58	154	Left	264
11	M	40	84	186	Left	156

2.2. Procedure

Each patient performed two sessions of gait analysis at their preferred walking velocity without (control condition) and with their own KAFO (KAFO condition). For each gait analysis, subjects wore their own shoes. Each condition was carried out in a 10 m gait corridor, which allowed at least eight successive gait cycles to be recorded. Six trials were carried out (thus >48 gait cycles were recorded for each patient). Each patient performed the two gait analyses successively with a 10 minute rest period in between.

2.2.1. Knee–ankle–foot orthosis (KAFO)

The KAFOs were all the same maker, made from carbon and were all custom made for each patient. Each KAFO was made up of 3 parts: crural, leg and foot. The crural and leg parts were made from pre-impregnated carbon and the foot part was made of polypropylene. The crural part included an anterior thigh cuff, approximately 10 cm wide, the lower portion of which sat 2–3 cm above the patella. The leg part was constituted of a posterior cuff fixed over the popliteal fossa (as high as possible without interfering with sitting). The upper edge was cut so as to free the two hamstring tendons. The width of the posterior leg cuff was 10 to 12 cm on average. It extended down into two lateral carbon stiles which were connected to the heel and the foot parts. The foot part was composed of an ankle–foot orthosis (AFO) which was inserted into the shoes. The total weight of each KAFO was approximately 800 g. The adjustment of the knee joint angle was decided during a multidisciplinary medical consultation. This angle should be as close as possible to 0° depending on the etiology of the genu recurvatum and to ensure the stability of the knee during the stance phase of gait.

2.2.2. Gait analysis

Gait was analyzed using a motion capture system with 8 optoelectronic cameras (Motion Analysis Corporation, Santa Rosa, CA, USA, sampling frequency 100 Hz). The trajectories of 30 reflective markers placed on anatomical landmarks using the Helen Hayes marker set (Kadaba et al., 1990), were collected and filtered using a fourth-order zero-lag Butterworth low-pass-filter, with a 6 Hz cutoff frequency (Winter et al., 1974). In the KAFO condition, reflective markers were disposed directly on the KAFO joint (stop joint and AFO) in the axis of the centers of rotation of the knee and the ankle of the paretic limb. Ground reaction forces were measured synchronously with the kinematic data using two force plates (AMTI, Watertown, MA, USA, sampling frequency 1000 Hz) staggered along the walkway.

2.3. Data analysis

Calculation of spatio-temporal parameters and joint kinematics was carried out using OrthoTrack 6.5 software (Motion Analysis Corporation, Santa Rosa, CA, USA) with Dempster's anthropometric table (Dempster, 1955) and inverse kinetics calculations were carried out on the kinetic data (Grood and Suntay method). During the gait sessions, 3 types of parameters were calculated: spatio-temporal, kinematic and kinetic gait parameters.

- Spatio-temporal parameters of the gait cycle were quantified for both limbs (paretic and non-paretic): velocity, cadence, step length, step width, stride length, stance phase and swing phase duration. Temporal parameters of symmetry between the two limbs were quantified using the stance phase duration symmetry ratio (paretic limb stance phase duration/non paretic limb stance phase duration) and the swing phase duration symmetry ratio (paretic limb swing phase duration/non-paretic limb swing phase duration) for the two conditions (Roth et al., 1997). The closer the symmetry ratio is to the value of 1, the more the gait is symmetrical. If the symmetry ratio is higher than 1 then the paretic limb parameter is higher than the non-paretic limb parameter. These two measures have been determined to be valid, reliable and useful for the measurement of hemiplegic gait quality (Brandstater et al., 1983; Dettmann et al., 1987; Wall and Ashburn, 1979).
- Kinematic parameters were quantified in the stance and swing phases of gait: peak knee extension; peak knee flexion; peak hip extension; peak hip flexion; peak ankle plantarflexion; and peak ankle dorsiflexion.
- With regard to the kinetic parameters, the internal flexion/extension moments at the hip, the knee and the ankle during stance phase were quantified for both limbs with and without the KAFO. Internal joint moments were normalized for body weight and reported in Newton meters per kilogram ($N \cdot m \cdot kg^{-1}$). The weight of the orthosis was not considered in the calculation because: i) it could be considered as negligible with regard to the weight of the thigh, the shank and the leg and ii) it is very difficult to evaluate its real impact on joint kinetics using an inverse kinetics calculation because the mass distribution of the different parts varies from one patient to another and over the gait cycle.

2.4. Statistical analysis

Kolmogorov–Smirnov tests were conducted before the statistical analysis and did not confirm that data were normally distributed. Mean values and standard deviations are given in the text to allow comparison with values in other studies. The median values which were used for the statistical analysis are given in Tables 2 and 3. In order to measure the effects of the KAFO on the primary outcome measure (angle of knee extension during stance) a Wilcoxon test was used (control vs KAFO condition). The secondary parameters were also evaluated using a Wilcoxon test. The threshold of significance was fixed at $P<0.05$. Statistical analysis was performed using Statistica 7 (StatSoft, Inc., Tulsa, OK, USA).

3. Results

3.1. Spatio-temporal parameter (Table 2)

Gait velocity was significantly greater in the KAFO condition than in the control condition (+21%, $P=0.025$). Stride length and cadence were also significantly greater in the KAFO condition (respectively, +15%, $P=0.030$ and +11%, $P=0.049$). There was no significant difference between the two conditions for step width ($P=0.384$). Step length of the non-paretic limb was greater in the KAFO condition (14%, $P=0.005$) and swing phase duration of the paretic limb was significantly shorter in the KAFO condition (−29%, $P=0.003$).

3.2. Gait symmetry

Swing phase duration asymmetry ratio was significantly lower in the KAFO condition compared to the control condition (from 1.93 (0.77) to 1.27 (0.10), $P=0.014$), meaning that the symmetry between the paretic and non-paretic limbs increased. However, there

Table 2
Spatio-temporal and kinematic parameters.

Spatio-temporal parameters	Without KAFO		With KAFO	
Velocity ($m \cdot s^{-1}$)	0.48 [0.57 (0.36)]		0.80 [0.73 (0.34)][a]	
Stride length (m)	0.82 [0.78 (0.33)]		1.06 [0.92 (0.35)][a]	
Cadence ($step \cdot min^{-1}$)	72.0 [79.2 (25.4)]		65.1 [88.9 (23.4)][a]	
Width length (cm)	22.3 [21.8 (6.3)]		20.4 [20.9 (5.5)]	
Step length non-paretic limb (m)	0.36 [0.35 (0.18)]		0.44 [0.40 (0.20)][b]	
Step length paretic limb (m)	0.46 [0.42 (0.16)]		0.52 [0.48 (0.15)]	
	Non-paretic side	Paretic side	Non-paretic side	Paretic side
Stance phase duration (%)	78.6 [75.8 (10.2)]	62.3 [62.6 (7.3)][c]	68.9 [72.3 (8.4)]	59.5 [62.2 (6.2)][c]
Swing phase duration (%)	21.4 [24.2 (10.2)]	37.7 [37.4 (7.3)][c]	31.1 [27.7 (8.4)]	40.5 [37.8 (6.2)][c]
Stance phase duration (s)	0.94 [1.26 (0.70)]	0.77 [1.01 (0.53)][c]	0.88 [1.08 (0.71)]	0.75 [0.91 (0.58)][c]
Swing phase duration (s)	0.36 [0.33 (0.10)]	0.60 [0.59 (0.20)][c]	0.37 [0.36 (0.10)]	0.48 [0.46 (0.20)][c,a]
Stance phase				
Peak hip extension (°)	−2.7 [−4.6 (12.8)]	−2.6 [−3.0 (10.4)]	−7.2 [−7.5 (10.5)]	−0.9 [−2.9 (8.4)]
Peak hip flexion (°)	40.1 [41.5 (9.2)]	28.7 [26.9 (10.0)][c]	41.2 [38.9 (11.2)]	31.3 [30.6 (9.3)][c,a]
Peak knee extension (°)	7.0 [5.1 (7.4)]	−11.8 [−16.2 (11.9)][c]	0.6 [2.2 (6.1)]	−7.8 [−7.6 (7.4)][c,a]
Peak knee flexion (°)	35.3 [40.0 (9.1)]	9.1 [13.2 (9.5)][c]	37.4 [38.3 (6.4)]	14.7 [14.5 (9.3)][c]
Peak ankle plantarflexion (°)	0.1 [−0.8 (7.6)]	−12.1 [−13.4 (9.7)][c]	1.0 [0.2 (7.2)]	−3.1 [−5.3 (5.1)][c,a]
Peak ankle dorsiflexion (°)	19.4 [18.3 (7.1)]	0.9 [1.4 (10.4)][c]	21.0 [19.4 (5.4)]	8.1 [8.2 (5.3)][c,a]
Swing phase				
Peak hip extension (°)	14.5 [7.6 (17.0)]	8.9 [7.2 (12.1)]	5.5 [4.3 (12.3)]	4.7 [4.2 (12.3)]
Peak hip flexion (°)	41.9 [42.5 (10.1)]	32.8 [31.9 (9.5)]	42.7 [41.9 (11.5)]	30.1 [32.8 (9.3)]
Peak knee extension (°)	26.0 [22.7 (15.2)]	0.6 [1.7 (6.7)][c]	17.4 [15.3 (11.9)][b]	6.8 [5.5 (7.5)][c,a]
Peak knee flexion (°)	69.0 [64.2 (9.8)]	28.0 [26.9 (16.2)][c]	68.2 [64.9 (9.7)]	29.5 [28.9 (16.8)][c]
Peak ankle plantarflexion (°)	8.7 [11.0 (6.3)]	−3.8 [−3.6 (8.1)][c]	11.9 [11.4 (7.1)]	2.8 [2.6 (5.3)][c,a]
Peak ankle dorsiflexion (°)	−7.3 [−3.5 (10.0)]	−11.4 [−13.3 (8.5)][c]	−6.9 [−6.1 (9.2)]	−2.6 [−2.4 (4.5)][a]

Median values for the gait parameters of non-paretic and paretic limbs, without and with KAFO (mean and standard deviation in brackets).
[a] Significant difference between the two conditions for the paretic limb ($P<0.05$).
[b] Significant difference between the two conditions for the non-paretic limb ($P<0.05$).
[c] Significant difference between the two limbs ($P<0.05$).

Table 3
Internal joint moments ($N \cdot m \cdot kg^{-1}$).

	Without KAFO		With KAFO	
	Non-paretic side	Paretic side	Non-paretic side	Paretic side
Initial double contact				
Hip moment	0.58 [0.64 (0.21)]	0.36 [0.40 (0.27)][a]	0.63 [0.63 (0.22)]	0.39 [0.34 (0.25)][a]
Knee moment	−0.10 [−0.12 (0.20)]	−0.23 [−0.30 (0.22)]	−0.11 [−0.09 (0.17)]	−0.10 [−0.12 (0.15)][b]
Ankle moment	0.07 [0.16 (0.21)]	0.25 [0.28 (0.18)]	0.11 [0.07 (0.17)]	0.02 [0.09 (0.18)][b]
Simple support				
Hip moment	0.15 [0.13 (0.20)]	0.18 [0.20 (0.21)]	0.04 [0.03 (0.14)]	0.05 [0.06 (0.24)][b]
Knee moment	−0.20 [−0.24 (0.22)]	−0.44 [−0.54 (0.52)]	−0.11 [−0.09 (0.19)]	−0.25 [−0.27 (0.34)]
Ankle moment	0.86 [0.83 (0.20)]	0.38 [0.62 (0.36)]	0.69 [0.71 (0.27)]	0.46 [0.48 (0.26)][a]
Final double contact				
Hip moment	−0.32 [−0.34 (0.15)]	−0.21 [−0.24 (0.22)][a]	−0.45 [−0.44 (0.19)]	−0.46 [−0.35 (0.24)][a]
Knee moment	0.09 [−0.05 (0.18)]	−0.15 [−0.21 (0.19)][a]	0.06 [0.08 (0.15)]	−0.10 [−0.09 (0.13)][a]
Ankle moment	0.63 [0.62 (0.19)]	0.46 [0.41 (0.23)]	0.65 [0.68 (0.14)]	0.51 [0.53 (0.20)][a]

Median values for internal joint moments of non-paretic and paretic limbs, without and with KAFO (mean and standard deviation in brackets).
[a] Significant difference between the two limbs ($P<0.05$).
[b] Significant difference between the two conditions for the paretic limb ($P<0.05$).

was no significant difference between the two conditions for the stance phase duration asymmetry ratio (from 0.82 (0.11) to 0.86 (0.08), $P=0.132$).

3.3. Kinematic parameters (Table 2; Fig. 1)

3.3.1. Stance phase

Peak knee extension (knee hyperextension) of the paretic limb was significantly lower in the KAFO condition compared to the control condition ($P=0.029$). Peak hip flexion ($P=0.047$) and peak ankle dorsiflexion ($P=0.010$) were significantly greater in the KAFO condition whereas peak ankle plantarflexion ($P=0.023$) was significantly lower. There were no significant differences between conditions for peak hip extension ($P=0.943$) or peak knee flexion ($P=0.542$) of the paretic limb or for any of the kinematic parameters of the non-paretic limb.

3.3.2. Swing phase

Peak knee extension ($P=0.031$) and peak ankle dorsiflexion ($P=0.008$) of the paretic limb were significantly greater in the KAFO condition whereas peak ankle plantarflexion ($P=0.001$) was significantly lower. There were no significant differences between conditions for peak hip extension and flexion (respectively, $P=0.099$ and $P=0.726$) or for peak knee flexion of the paretic limb ($P=0.492$). Peak knee extension ($P=0.047$) of the non-paretic limb was significantly lower in the KAFO condition. There were no significant differences between conditions for peak knee flexion or for the kinematic parameters of the hip and ankle joints of the non-paretic limb.

3.4. Kinetic parameters (Table 3)

The knee flexor moment ($P=0.047$) and ankle plantarflexion moment ($P=0.008$) were significantly decreased during initial double stance phase and the hip flexion moment was significantly decreased during single support phase ($P=0.019$) in the paretic limb. There were no significant differences between conditions for the kinetic parameters of the non-paretic limb.

4. Discussion

The principal aim of this study was to evaluate the effectiveness of a KAFO on gait in hemiplegic patients with genu recurvatum due to spasticity or weakness of the quadriceps. The significant decrease in paretic knee hyperextension (the primary outcome measure) during the stance phase of gait confirms that the KAFO was indeed effective in reducing the genu recurvatum. The secondary aim was to quantify the biomechanical adaptations induced by the KAFO in both the paretic and non-paretic limbs. The results showed that wearing a KAFO resulted in improvements of spatio-temporal parameters as well as the kinematic and kinetic parameters of the paretic hip and ankle joints during both stance and swing phase, with few changes in the non-paretic limb.

As expected, the KAFO significantly decreased the primary outcome measure: knee hyperextension of the paretic limb (from −16.2 (11.9)° to −7.6 (7.4)°) during stance phase. However, the genu recurvatum was not totally resolved. This could be due to the diversity of etiologies of the genu recurvatum in the sample (Inaba, 1967; Perry, 1992) and/or the individual settings of the stop angles. Similarly, the internal knee flexor moment was significantly reduced by the KAFO but not eliminated. Indeed, in the KAFO condition, the resultant ground reaction force vector still passed in front of the knee but the knee flexor moment lever arm was reduced because the setting of the stop limited knee hyperextension. Despite the fact that a degree of genu recurvatum remained, the mean decrease of around 8° of hyperextension during stance confirmed that the KAFO evaluated in this study is clinically useful for the reduction of genu recurvatum in hemiplegic patients (Klejman et al., 2010).

The improvements in the spatiotemporal parameters found in the KAFO condition are in accordance with previous studies. Three studies have previously reported the spatiotemporal benefits of such orthoses (Farncombe, 1980; Morinaka et al., 1982, 1984). Morinaka et al. (1982) showed, in 36 hemiplegic patients with genu recurvatum, that a KAFO, with a genucentric knee joint, improved ambulation by decreasing instability during stance phase and increasing gait velocity. The same authors (Morinaka et al., 1984) found similar results after several months of gait training with a KAFO and showed that muscle activation patterns, assessed by surface electromyography, tended to be normalized after the training. Farncombe (1980) showed that after a 6 month physical gait training program with an anti-recurvatum orthosis ("Swedish knee cage"), the patients were able to walk without the orthosis and controlled their knee extension during stance phase. However, all these studies were carried out using heavier types of KAFO and did not involve detailed kinematic and kinetic analyses. In the present study, the increase in gait velocity was associated with an increase in cadence and stride length. The step length was increased only for the non-paretic limb whereas the duration of the swing phase was decreased only for the paretic limb. The next part of the discussion will relate these changes with the biomechanical changes caused by the KAFO.

It is important to note that the biomechanical action of the KAFO is not limited to the reduction of knee hyperextension of the paretic limb during stance phase. The results of our study showed that the KAFO also significantly increased peak hip flexion and peak ankle dorsiflexion during stance phase. The increased dorsiflexion during

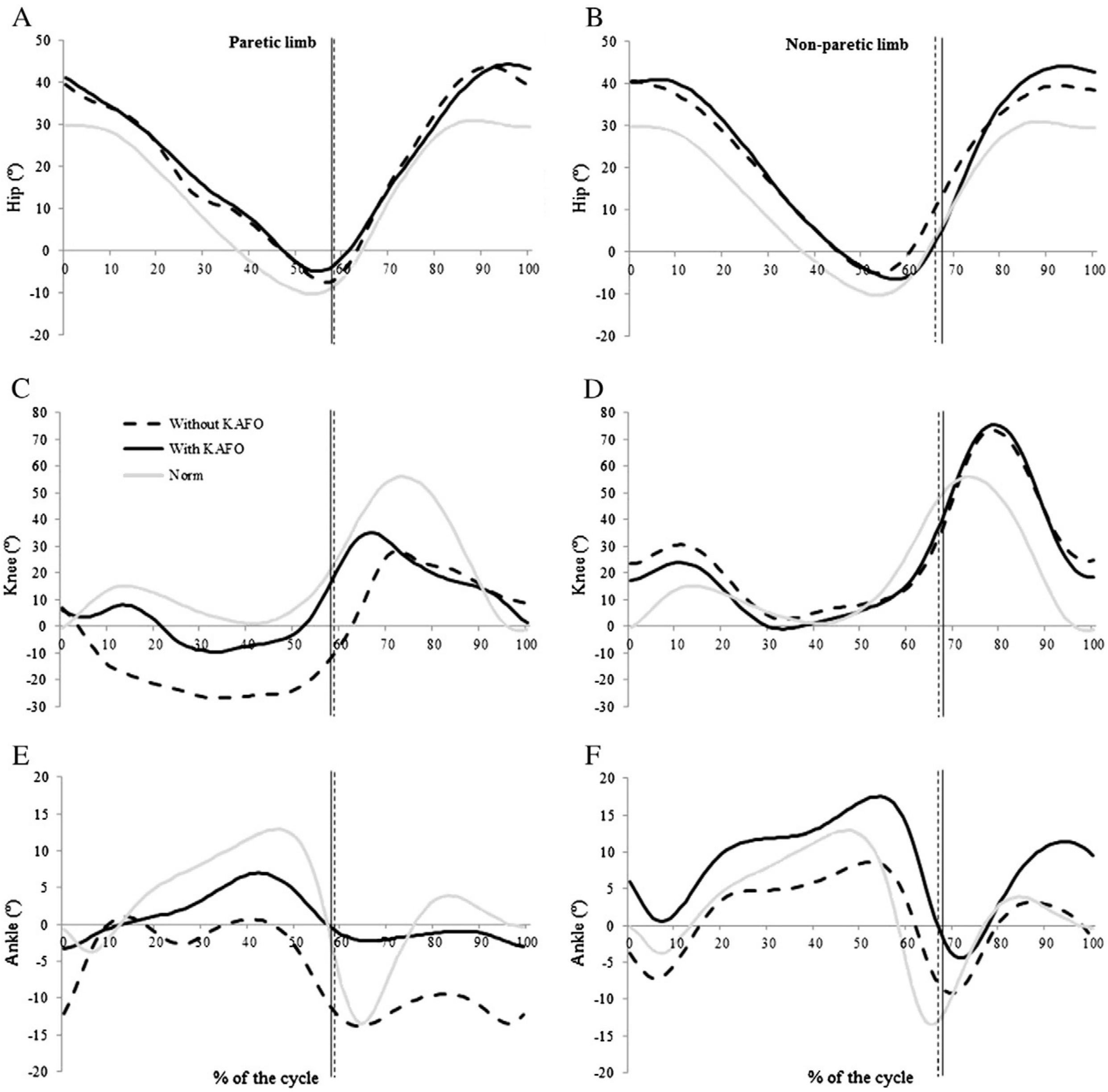

Fig. 1. Normalized gait cycle of the mean hip (A–B), knee (C–D) and ankle (E–F) flexion/extension of the paretic limb (left panels) and non-paretic limb (right panels) of a typical patient. The solid black line indicates the KAFO condition, the dotted black line the control condition (without KAFO) and the solid gray line normal values. The solid black horizontal line represents the beginning of swing phase for the KAFO condition and the dotted black horizontal line the control condition for the beginning of swing phase.

stance phase is likely to be mechanically related to the decreased knee hyperextension. Because of the multiple neurological consequences of the pyramidal syndrome which occurs after stroke, it is difficult to pinpoint one etiology which may be responsible for the biomechanical changes, particularly since they are often combined (Bleyenheuft et al., 2010). For the majority of the patients in our sample, the knee hyperextension could be explained either by quadriceps spasticity or weakness, triceps surae spasticity or by proprioceptive impairments around the knee. The KAFO was therefore well adapted to limit the genu recurvatum, without restricting dorsiflexion during stance phase. For some of the patients ($n=3$), the lack of ankle dorsiflexion during swing phase was mainly due to a contracture of the triceps surae muscles. These patients exhibited toe walking gait during stance phase. In these cases, the KAFO was adjusted in order to maintain a moderate equinus, and to preserve a few degrees of genu recurvatum. This probably explains why peak ankle dorsiflexion was significantly increased concomitantly with the decrease in peak knee extension (knee hyperextension) during stance phase. The increased peak hip flexion during stance phase is more difficult to explain. These kinematic adaptations during paretic limb stance phase (increased peak hip flexion and peak ankle dorsiflexion) may be directly linked to the increased stride and step lengths of the non-paretic limb (respectively, +15% and +14%) (Lelas et al., 2003). However, these changes were small and are likely not to be clinically relevant (Klejman et al., 2010).

The unilateral deficits that occur after stroke frequently result in an asymmetrical gait pattern which is generally characterized by an increase in paretic limb swing phase duration (Perry, 1992). The results of this study clearly show that the KAFO induced a significant decrease in swing phase asymmetry ratio mainly due to the decrease in paretic limb swing phase duration. The increased dorsiflexion which occurred during swing phase of the paretic limb is not surprising as the aim of the AFO part of the KAFO is to restrict plantarflexion during swing phase (in other words, to maintain the ankle in dorsiflexion). The decreased peak knee extension of the non-paretic limb during swing phase with the KAFO is more difficult to explain. Our main hypothesis is that the reduction in knee extension during early swing phase can be caused by the fact that the rigid AFO limits dorsiflexion in late stance, thus reducing the forward movement of the tibia. Whatever the cause of this result, it is very important as it explains the decrease in swing phase duration and the lack of modification of the paretic limb step length.

For the same reason as mentioned above, the increased knee extension of the non-paretic limb during swing phase (just before heel strike) could be due to the increase in gait velocity and/or to the improvement of kinematic parameters of the paretic limb during stance phase. This increase in knee extension during swing phase of the non-paretic limb associated with the increased dorsiflexion during stance phase of the paretic limb could explain the increase in the non-paretic limb step length. Finally, the lack of changes in the stance phase kinematic parameters of the non-paretic limb could explain the lack of modification of the step length in the paretic limb. However, the stance phase asymmetry ratio was not changed (0.82 (0.11) to 0.86 (0.08), $P=0.132$) since the stance phase duration was not significantly modified in either limb. Hence, the significant increase in gait velocity is only due to the decrease in swing phase duration which is related to the improved kinematics.

5. Conclusion

This study showed the beneficial effects of a carbon KAFO on biomechanical gait parameters in hemiplegic patients with genu recurvatum. In particular, the KAFO increased gait velocity by improving both the cadence and the stride length and by decreasing the swing phase asymmetry ratio. This study is the first to evaluate the mechanisms underlying these gait improvements. The changes in spatio-temporal parameters were mainly due to a decrease in the genu recurvatum during stance phase and to an increase in paretic limb ankle dorsiflexion during both phases. However, interpretation of these results must be cautious because of the small sample of hemiplegic patients included. In clinical practice, wearing a standard KAFO is not well accepted and often refused by patients. It therefore seems necessary to propose a lighter functional orthosis, which is robust and esthetically acceptable, such as the type used in this study. Further studies are needed in order to assess the effects of the KAFO setting on gait in hemiplegic patients as well as to carry out electromyographic analysis in order to better understand the changes which occur.

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