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## Original article

# Effect of the Rectus Femoris motor branch block on post-stroke stiff-legged gait

Frédéric CHANTRAINE[1], Christine DETREMBLEUR[2] and Thierry M. LEJEUNE[2,3]

[1]Centre de Rééducation Fonctionnelle et de Réadaptation, Luxembourg ; [2]Rehabilitation and Medicine Unit, Université catholique de Louvain, Brussels, Belgium ; [3]Departement of Physical Medicine and Rehabilitation, Cliniques universitaire Saint-Luc, Université catholique de Louvain, Brussels, Belgium

*Abstract*

*Six hemiparetic stroke patients presenting with a stiff-legged gait underwent a motor branch block of* rectus femoris *(*RF*).*

*A gait analysis, including synchronous kinematic, dynamic, energetic recordings was performed before and after motor branch block of* RF. *The electrical activity of* RF, Vastus Medialis, Vastus Lateralis, Vastus Intermedius *was also recorded.*

*Before nerve block, the sagittal kinematic data showed a decrease in both the mean maximum knee angular displacement during swing phase and the mean angular knee velocity at toe-off. In the sagittal kinetic data, the mean knee extension moment was increased. The mean mechanical work, internal and external work, was increased and the mean energy cost was also decreased. After the motor branch block of* Rectus Femoris*, the sagittal mean knee angular displacement and the mean internal work of the gait were significantly improved.*

*Key words* : Stroke ; stiff-legged gait ; nerve block ; rectus femoris ; gait analysis.

### Introduction

Stiff-legged gait, also called stiff-knee gait, is commonly seen among post-stroke hemiparetic spastic patients. This gait pattern is characterized by a lack of normal knee flexion during the swing period of gait. The limited knee flexion may cause toe dragging, increase the risk of falls, and compromise the stability of gait. The compensatory movements performed to clear the toe at swing phase, such as ipsilateral hip circumduction or controlateral vaulting, could result in increased vertical displacement of the center of body mass and greater energy expenditure.

The physiopathology of the stiff-legged gait is not well understood. In the literature, several hypotheses are proposed. Most often Quadriceps muscle hyperactivity is mentioned but other causes such as hip flexor weakness and ankle plantar flexor muscle hyperactivity at terminal stance phase seem also to play an important role (Perry, 1992). In adult stroke patients, Kerrigan *et al.* (1991) showed that limited knee flexion during the swing phase was related to Quadriceps Femoris hyperactivity. Between its four heads, they pointed out the Rectus Femoris (*RF*) as a major contributor to this dynamic abnormalities. Sung and Bang (2000) confirmed this implication of the *RF* by measuring an increase of knee flexion after *RF* motor branch block.

Using dynamic simulation, Riley and Kerrigan (1998) showed that knee motion in swing phase was more sensitive to knee muscular moment than to hip muscular moment. Indeed, bi-articular muscles affected the motion of the knee more than the hip, suggesting that *RF* hyperactivity decreases knee flexion.

Although the implication of the *RF* in stiff knee gait has been well recognized in cerebral palsy children, leading to *RF* surgical transfer or release, or to *RF* Botulinum toxin injection, few studies have looked at its deleterious role in adult stroke patients.

In the present study, the *RF* implication on the stiff-knee gait was analysed by performing gait analysis before and after *RF* motor branch block. Short term motor branch block is currently used in the management of spastic patients to identify the muscle responsible for gait disturbance. This test would enable the result of more definitive therapy such as surgery or botulinum toxin injection to be predicted.

### Materials and methods

#### SUBJECT

Six hemiparetic and spastic chronic stroke patients presenting with a stiff-knee gait were recruited from our outpatient rehabilitation department (Table 1). They were able to walk independently irrespective of the walking surface and were familiar with walking on a treadmill. There were no

Table 1

Anthropometry and history of the patients

| N° | GENDER (M/F) | AGE (years) | Weight (Kg) | Height (m) | Onset of stoke (months) | Hemiparetic side (right/left) | Mean walking speed (Km $h^{-1}$) |
|---|---|---|---|---|---|---|---|
| 1 | M | 65 | 74 | 1,60 | 68 | left | 2 |
| 2 | F | 41 | 52.5 | 1.58 | 26 | right | 2 |
| 3 | M | 27 | 74 | 1.87 | 240 | right | 3 |
| 4 | M | 57 | 85 | 1.87 | 11 | right | 2 |
| 5 | F | 78 | 73 | 1.66 | 55 | left | 2 |
| 6 | F | 63 | 70.5 | 1.63 | 76 | left | 2 |

other major ongoing medical disorders. In the acute phase, all patients were treated in a Stroke Unit and thereafter followed an intensive rehabilitation program. All gave their informed consent prior to freely participating in the study. The Ethics Committee of the Medical School of the Université catholique de Louvain approved the study.

## Gait analysis

All subjects underwent a routine clinical examination before the *RF* motor branch block. Specifically, a Duncan-Ely test was performed to evaluate the Quadriceps spasticity before and after *RF* motor branch block. A gait analysis, including synchronous kinematic, dynamic, energetic and electromyographic recordings while walking on a treadmill (Mercury LT med, HP Cosmos, Germany) was performed before and after *RF* motor branch block. The gait was analysed at the same comfortable walking speed before and after the motor branch block (Table1).

### *Kinematic and dynamic*

Segmental kinematics were measured with the Elite system (BTS, Italy). Eighteen reflective markers were positioned on specific anatomical landmarkers following the Newington Hospital Helen Hayes model (Davis *et al.*, 1991). Six infrared cameras measured marker coordinates in 3 spatial planes at 100 Hz. Then angular displacements of the trunk, pelvis, hip, knee and ankle, were computed based on the determination of Euler angles with an x-y-z axis rotation sequence (Davis *et al.*, 1991). The knee angular velocity was then computed from its angular displacement.

The ground reaction force was collected with a force measuring treadmill (Dierick *et al*, 2004). The decomposition of superimposed ground reaction forces into left and right foot force profiles was performed following the algorithm described by Davis and Cavanagh (1993). The net sagittal moments at the knee were computed from the ground reaction force and segmental kinematics by the inverse dynamic model. Analysis was focused on sagittal knee motion during the swing phase : maximum knee flexion (K5) (Benedetti *et al*, 1998), range of knee flexion (difference between the maximum knee extension in stance, K3, and the maximum flexion in swing, K5), knee flexion velocity at toe off and the maximal net extension moment at the knee. The normal values of kinematic and dynamic come from Winter (1991) (Table 2).

### *EMG*

The muscle electrical activity (EMG) of Rectus Femoris (*RF*), Vastus Medialis (*VM*), Vastus Lateralis (*VL*), Vastus Intermedius (*VI*) and Semi-Tendinosus (*ST*) was recorded by a telemetry EMG system (Telemg, BTS, Italy). *VM, VL, ST* EMG were recorded with surface electrodes (Medi-Trace, Graphic Controls Corporation, NY, USA). *RF* and *VI* EMG were recorded with fine wire electrodes inserted in the muscle, except in one patient for psychological reasons.

The signal was digitised at 1000 Hz, full-wave rectified, and filtered (bandwidth 25-300 Hz). The onset and cessation of muscle activity were both detected visually. EMG activity of each muscle was normalised to 100% in time with the gait cycle before averaging. The normal patterns of muscle sequences come from Perry (1992)

### *Mechanical work*

The total mechanical work ($W_{tot}$) done by the muscles during gait was computed following the method described by Cavagna. The total mechanical work ($W_{tot}$) was calculated as the sum of external work ($W_{ext}$) and internal work ($W_{int}$). $W_{ext}$ was defined as the positive work done by muscles to move the centre of body mass ($COM_b$) relative to the surroundings (Cavagna *et al.*, 1975). $W_{ext}$ was computed from the measurement of the three dimensional (3D) ground reaction forces (GRFs), at a sampling rate of 100 Hz, following the method described by Cavagna and adapted to pathological gait (Detrembleur *et al.*, 2000).

$W_{int}$ was defined as the positive work done by the muscle to accelerate limbs relative to the $COM_b$ (Cavagna and Kaneko, 1976). $W_{int}$ was computed

Table 2

Results of clinical examination and gait analysis before and after motor branch block of RF

| | Nerve Block (lidocaïne-bupivacaïne) | | | Normal values |
|---|---|---|---|---|
| | Before | After | *P* | |
| Clinical examination | | | | |
| Duncan-Ely test | 2 [2-3] | 0.5[0-1] | 0.031 | 0 |
| *Kinematic variables* | | | | |
| Step length (m) | 0.5 ± 0.08 | 0.5 ± 0.08 | 0.421 | 0.52* |
| Stance time (% stride) | 69,00 ± 3.7 | 69 ± 2.6 | 1 | 65.65* |
| Maximum knee flexion during swing (K5) (degree) | 38.5 ± 15.2 | 40.4 ± 14.8 | 0.502 | 62.55* |
| Range of knee motion (K5-K3) (degree) | 40.5 ± 18.3 | 45.3 ± 16.8 | 0.04 | 54.32* |
| Angular Knee velocity at toe-off (°$s^{-1}$) | 107.7 ± 117.9 | 146.3 ± 96.6 | 0.11 | 224.95* |
| *Dynamic variables* | | | | |
| Maximal knee extension moment at preswing (N m $kg^{-1}$) | 0,40 ± 0.17 | 0.42 ± 0.22 | 0.764 | 0.157* |
| External work (J $kg^{-1}$ $m^{-1}$) | 0.43 ± 0.1 | 0.4 ± 0.09 | 0.21 | 0.313** |
| Internal work (J $kg^{-1}$ $m^{-1}$) | 0.18 ± 0.04 | 0.15 ± 0.04 | 0.029 | 0.144** |
| Total work (J $kg^{-1}$ $m^{-1}$) | 0.61 ± 0.1 | 0.56 ± 0.1 | 0.057 | 0.456** |
| Energetic cost (J $kg^{-1}$ $m^{-1}$) | 5.5 ± 1.5 | 5.2 ± 1.7 | 0.598 | 2.7*** |

* Normal values come from Winter (1991)
** Normal values come from Willems (1995)
*** Normal values come from Mc Ardle (1996)
Significance was assumed at $p < 0.05$ (highlighted values).

from kinematic data following the method described by Willems *et al* (1995). The body was divided into 7 rigid segments : head-arm-trunk (HAT), thighs, shanks and feet. The internal mechanical energy of the body segments corresponded to the sum of the rotational and translational energies of these segments due to their movements relative to the $COM_{wb}$. For each lower limb, the internal mechanical energy-time curves of the thigh, shank and foot were summed. Then, the $W_{int}$ of each lower limb and the HAT segment were calculated separately as the sum of the increments of the respective internal mechanical energy curves during one stride. Finally, $W_{int}$ during gait corresponded to the sum of the $W_{int}$ used to move the lower limbs and HAT segments. The normal values of mechanical work come from Willems *et al* (1995) (Table 2).

*Energetic*

The metabolic cost was determined from the oxygen consumption (Quark b$^2$, Cosmed, Italy ; Polar). Oxygen consumption ($\dot{V}o_2$) and carbon dioxide production ($\dot{V}co_2$) were measured throughout the treadmill test. Values were automatically converted by software (Cosmed Quark b$^2$ win, Version 5.1a) to standard temperature, pressure and dry $\dot{V}o_2$. The Respiratory Quotient (*RQ*) was computed as the ratio between $\dot{V}co_2$ and $\dot{V}o_2$. Each energetic measurement consisted of a rest period (the subject standing on the treadmill), followed by a walking period at a speed spontaneously chosen by the patient. Each period was maintained until a steady state was reached and maintained for a period of at least 90 sec. *RQ* remained always less than 1. The joules of energy expended per litre of oxygen consumed were computed based on *RQ*, following the Lusk equation (Mc Ardle *et al*., 1996). The energy expended at rest was subtracted from the energy expended when walking and then divided by the walking speed to obtain the net energetic cost of walking (*C*). The normal value come from Mc Ardle *et al*. (1996).

## Motor Branch block of rectus femoris

The motor branch block of the *RF* was performed following the method described by Sung and Bang (2000). The patient was lying supine with a slight external rotation of the limb. The femoral artery was palpated just below the inguinal ligament. A Teflon-coated nerve block needle was inserted around 1 cm lateral and 3 to 4 cm distal to the point of an arterial pulsation and alongside the medial border of the *RF*. The needle was manipulated to reach the motor branch of the *RF*. The needle placement was confirmed when a palpable contraction of the *RF* was obtained without contraction of the *VM* or *VL*, with an electrical stimulation of only 0.3 mA intensity, 1 ms duration. Then, a mixture of 1 ml of 2 % Lidocaïne and of 1ml of 1% Bupivacaïne solution was injected. The nerve block was performed at a single site. Half an hour after the injection, an Duncan-Ely test and a second gait analysis were performed.

## Data analysis

The Effect of *RF* selective motor branch block on Duncan-Ely test was tested with a Wilcoxon Signed rank test. The gait variables were compared before and after *RF* selective motor branch block with a paired t-test. All statistical analyses were

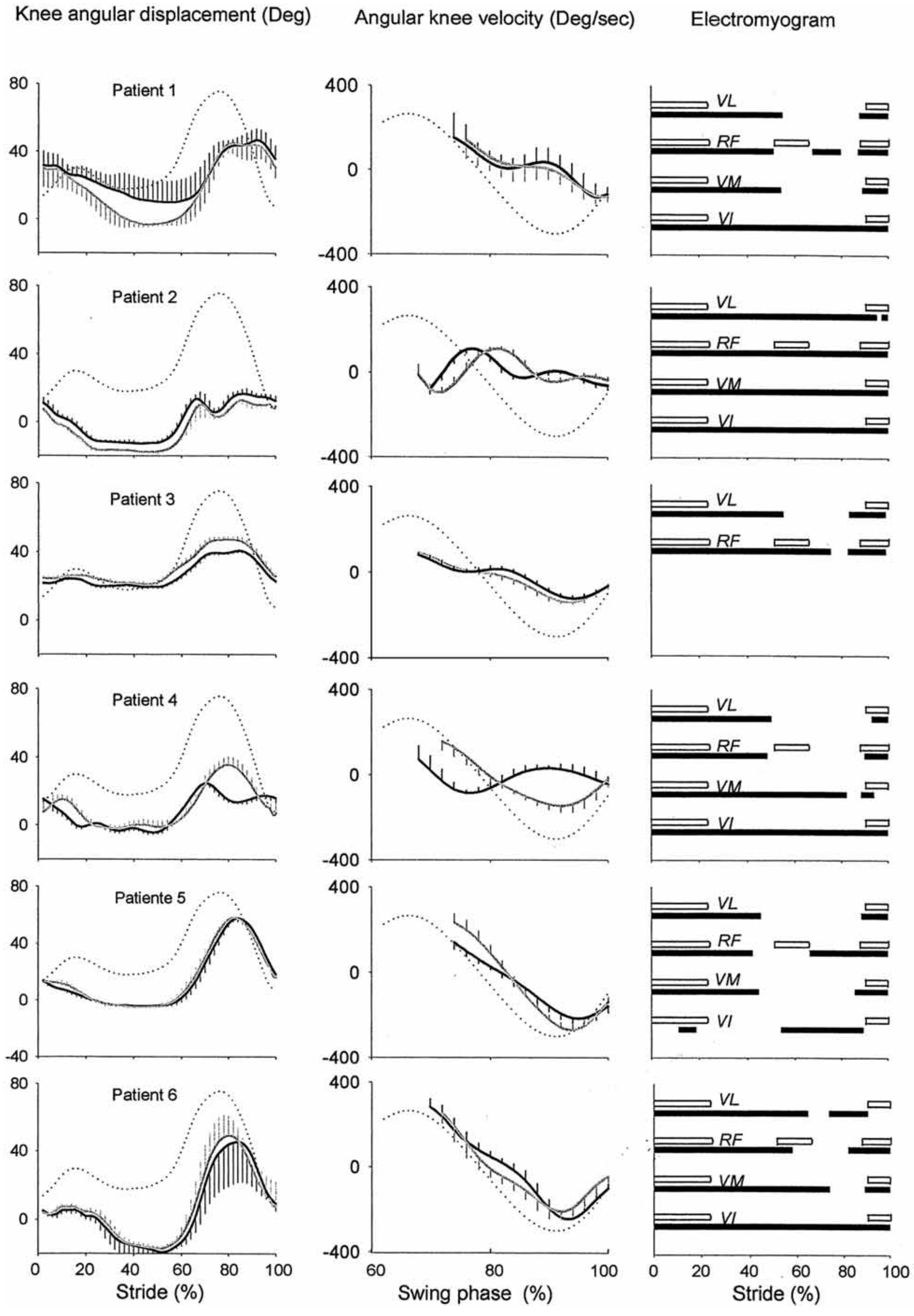


FIG. 1. — The first column shows the mean sagittal knee motion of the 6 patients during a stride normalised in 100%. The middle column shows the mean angular knee velocity during swing phase. The black line represents the mean data before motor branch block of *RF* and the grey line the mean data after nerve block. Dotted line is the normal curve (Winter, 1991). The last column shows the pattern EMG activity of *RF*, *VM*, *VL*, *and VI* for each patient before the motor branch block of the *RF*. In patient 3, only *RF* and *VL* were recorded for psychological reasons. The black rectangles represent the analysed muscle patterns and white rectangles represent normal patterns (Perry, 1993).

performed using sigmaSTAT 32 (version 2.0). A $p < 0.05$ was regarded as significant.

## Results

### Before RF motor branch block

The results obtained before the *RF* motor branch block are presented in the Table 2. Clinically, quadriceps spasticity was objectified by a pathological Duncan-Ely test (median score was equal to 2 [2-3]).

The left column of the figure shows the mean knee angular displacement before *RF* motor nerve block (black line) in each patient. During swing phase, knee flexion is decreased in all but one patient (patient 5). A dimpling or an abrupt flattening of the curve, characterizing a blocking of the rapid knee flexion or a reversal from knee flexion to knee extension is shown in patients 1-4. In patients 5 and 6, the shape of the curve is normal in swing.

The mean maximum knee flexion during swing (K5) was 38.5° ± 15.2°, 0.6 time smaller than the normal (62.5°) (Table 2). The mean range of knee motion (K5-K3) was 40.5° ± 18.3°, 0.75 time smaller than the normal value (54.3°) (Table 2).

The middle column of the figure shows mean angular knee velocity before *RF* motor branch block (black line) in each patient. The angular knee velocity is decreased at the toe-off except in patient 6. Moreover, the decrease in velocity was not regular at initial and mid swing except in patient 5. In patients 1-4, 6, we observed a trend toward acceleration at mid swing phase meaning a blocking of knee flexion.

The mean knee angular velocity at toe-off was 107.7 ± 117.9° $s^{-1}$, 0.5 time smaller than the normal value (225°$s^{-1}$) (Table 2).

The EMG pattern presented in the right column of the figure was very heterogeneous. In all but one patient (patient 5), *RF* presented a prolonged activity from preswing to midswing. All patients also presented an abnormal *VI* activity during these phases. The patient 5 that had no activity on the *RF* presented an abnormal activity on the *VI* and on the *VM*.

The mean maximum knee extension moment at the preswing phase was 0.4 ± 0.17 N m $kg^{-1}$, 2.5 times greater than the normal value référence (0.16 N m $kg^{-1}$) (Table 2).

The mean $W_{int}$ was 0.18 ± 0.04 J $kg^{-1}m^{-1}$, 1.3 times greater than the normal value (0.14 J $kg^{-1}m^{-1}$) (Table 2).

The mean $W_{ext}$was 0.46 ± 0.09 J $kg^{-1}m^{-1}$, 1.5 times greater than the normal value (0.31 J $kg^{-1}m^{-1}$) (Table 2).

The mean $W_{tot}$was 0.61 ± 0.1 J $kg^{-1}m^{-1}$, 1.3 times greater than the normal value (0.46 J $kg^{-1}m^{-1}$) (Table 2).

The energy cost was 5.5 ± 1.5 J $kg^{-1}m^{-1}$, 2 times greater than the normal value (2.7 J $kg^{-1}m^{-1}$) (Table 2).

### After RF motor branch block

The results obtained after the *RF* motor branch block are presented in Table 2.

The *RF* motor branch block was well tolerated by all subjects. After *RF* motor branch block, all patients reported a gait improvement, especially to the swing of their paretic limb. The transfer from decubitus to sitting position was also performed more easily. One patient reported themself able to walk faster after the block. No patient described quadriceps weakness or felt.

The Duncan-Ely test improved significantly in all patients, median score decreasing from 2 before to 0.5 after the block ($p < 0.05$) (Table 2).

Step length and stance phase duration were unchanged. The left column of the figure shows the mean knee angular displacement after *RF* motor branch block (grey line) for each patient. The dimpling or the flattening of the curve seen in the preblock curve disappears for patients 3 and 4 and decreases for patient 1. In patient 2, the dimpling remains the same after *RF* branch block. There are no changes in the curve of patient 5 and 6. The mean maximal knee flexion (K5) did not change significantly ($p = 0.502$) (Table 2). The mean range of knee motion (K3-K5), increased significantly from 40.5° to 45.3° ± 16.8° ($p < 0.05$).

In the middle column of the figure, the knee angular speed was compared before and after the block (grey line). After *RF* branch block, the dimpling of the curve was less important at mid swing and the decrease of angular speed was more regular. This trend was found in the patients where the dimpling and the flattening were decreased, but also in patient 6.

The mean extension knee moment was unchanged ($p = 0.764$).

The mean value of $W_{int}$ decreased significantly from 0.18 to 0.14 J $kg^{-1}m^{-1}$ ($p < 0.05$). The $W_{ext}$ was not affected by the motor nerve block ($p = 0.21$). The mean value of $W_{tot}$ tended to decrease from 0.62 to 0.53 J $kg^{-1}m^{-1}$ ($p = 0.057$). The energy cost was not improved.

## Discussion

In the present study, *RF* motor branch block improved knee flexion significantly during the swing phase. Despite the fact that patient 5 presented with a stiff-legged gait during free walking before the gait analysis, he had a normal sagittal knee angular displacement before *RF* motor branch block. This patient could have selected a treadmill walking speed slow enough (2 $kmh^{-1}$) not to launch a stretch reflex on the *RF* during the knee flexion.

Indeed, his *RF* did not show EMG activity during preswing to midswing.

The *RF* motor branch block normalised the knee motion of patient 4. In this patient, the good result is probably due to the fact that kinematic abnormalities were limited to the knee, whereas hip and ankle kinematic were almost normal (data not shown). On the contrary, patient 2 did not improve probably due to ankle equinus during stance phase which induced a knee recurvatum and slowed down knee flexion at toe-off.

In our study, the *RF* motor branch block significantly improved $W_{int}$ probably related to the improvement knee motion. Increased knee flexion would decrease the lower limb inertial moment and decrease the internal energy required to bring this limb forward during the swing phase. The improvement of knee motion would also reduce compensatory movements.

To our knowledge, only one study has looked at the effect on the motor branch block of *RF* on kinematic of stiff-legged gait (Sung and Bang, 2000). In their study, Sung and Bang have recorded the sagittal kinematics of the 31 patients and the EMG activity of *RF*, *VM* and *VL* without recording the activity of the *VI*. Their results showed a significant improvement of maximal knee flexion during swing period and a significant increase of the angular knee flexion velocity at toe-off.

In clinical practice, the motor branch block is often performed to temporarily analyse the gait pattern of the patient without the action of the spastic muscle (Filipetti and Decq, 2003). In our study, the *RF* motor branch block, allowed us to evaluate the repercussions of the activity of *RF* on stiff-legged gait. Indeed, the hyperactivity of the *RF* can be the cause of stiff-knee gait but, at the same time, could be necessary for gait stability of several hemiparetic patients. It already has happened that a patient could not walk any more after the activity of the *RF* was suppressed, due to lack of strength of the other heads of quadriceps. For this reason we prefer to perform a test-block before trying a more prolonged therapy such as a Botilinum toxin injection or a more definitive therapy such as tenotomy.

The motor branch block of the *RF* was performed irrespective of the electrical activity of the different heads of the quadriceps. The result was heterogeneous, and did not seem to depend on the number of heads with inappropriate activity at preswing to midswing. Nevertheless, Waters *et al.* (1980) observed better results after resection of the *RF* by the patient presenting electrical activity only on that muscle. That was confirmed by Sung and Bang (2000) who observed better results after motor branch block of the *RF* when it was the only quadriceps active during preswing and/or initial swing. In the present study, we could not confirm this assumption because we had no patient with only *RF* hyperactivity.

The decreased activity of hip flexor, or increased activity of triceps surae can also decrease the knee flexion at swing phase. Piazza and Delp (1996) used a muscle-driven simulation of normal swing phase to evaluate the maximum knee flexion. These simulations found that the normal knee flexion was decreased, in order of importance, by increased knee extension moment, decreased knee flexion velocity at toe-off and decreased hip flexion moment. The *RF* action was studied in particular and a *RF*-simulation was performed, showing that removing *RF* action resulted in excessive flexion and inversely, overactivity in *RF* decreased knee flexion.

Goldberg *et al.* (2003) found that many cerebral palsy children presenting with stiff-knee gait exhibit a low flexion knee velocity at toe-off, whereas their extension moment in swing phase is not increased. They proposed, since kinematic condition at toe-off was due to stance phase muscle activity, to study the link between muscle activity during stance and knee flexion velocity at toe-off. The same authors (Goldberg *et al.*, 2004), using a forward dynamic simulation of gait, showed that increased force in Soleus decreased knee flexion velocity. It is reasonable to believe that hyperactivity of Soleus decreases the knee flexion velocity and the maximum knee flexion at swing phase. In this same study, the simulation was also performed on the Iliopsoas, showing that an increased force in this muscle increased the maximum knee flexion velocity. The increased force in *RF* decreased the maximum knee flexion velocity, as described in our study where the knee flexion velocity at toe-off is tended to improve after motor branch block of the *RF*.

The stiff-legged gait can be caused by several factors : quadriceps hyperactivity, decrease of hip flexion moment and decrease of knee flexion velocity at toe-off. Gait analysis and motor nerve block could bring useful information to identify the muscle responsible for limited knee flexion for understanding the physiopathology and helping the clinician to treat the patient.

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Fr. CHANTRAINE, M.D.,
Centre National de Rééducation Fonctionnelle
et de Réadaptation,
Rue Englebert Neveu 2,
2224 Hamm (Luxembourg).
E-mail : f.chantraine@voila.fr.