Université de Montréal

CONTROL OF POSTURE AND STABILITY OF THE DOUBLE-JOINT (Shoulder / Elbow) ARM

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Ce mémoire intitulé :

CONTROL OF POSTURE AND STABILITY OF THE DOUBLE-JOINT (Shoulder / Elbow) ARM

presenté par

Pavel Mihaltchev

A été évalué par un jury composé des personnes

suivantes:

Président - rapporteur: Sylvie Nadeau
Directeur de recherche: Mindy Levin
Membre de jury: David Ostry

Sommaire. Ce projet visait à décrire le contrôle postural dans le système bi-articulaire du bras chez les patients ayant subi un accident vasculo-cérébral. L'objectif principal était de tester l'hypothèse que le contrôle de la position du bras est produit par des changements à la configuration référante du bras et par la précision des régions dans l'espace articulaire dans lesquelles l'activation ou co-activation musculaire se produit. Le projet est basé sur le modèle λ de l'hypothèse de point d'équilibre du contrôle moteur, qui propose que les mouvements sont produits en déterminant la configuration référante du corps dans l'espace. Dix sujets sains et 13 sujets hémiparetiques ont participé à l'étude. Les participants devaient résister l'application d'une force externe produite par les deux moteurs du torque qui contrôlent un bras robot. Après un délai aléatoire, la charge externe était enlevée sans préavis et la main du sujet se déplaçait donc involontairement à un nouveau point. Les résultats ont montré que les sujets hémiparetiques produisaient des torques initiaux plus bas par rapport aux sujets sains et leurs trajectoires étaient moins dispersées dans l'espace. Généralement, les patients ont conservé leur capacité de produire une configuration référante du bras. Cependant, ils avaient des valeurs d'instabilité augmentées. Notre conclusion principale est que bien que le mécanisme fondamental de production de mouvement peut être conservé chez les sujets avec hémiparesie, les modifications du système nerveux central et les changements dans les propriétés biomécaniques du membre supérieur influencent la commande centrale ce qui a pour conséquence d'augmenter instabilité dans le bras hémiparétique.

Mots clés: instabilité, modèle λ, contrôle moteur, mouvement bi-articulaire, hémiparésie, interaction torque - angle.

Summary. The present project aimed to describe the postural control of the double-joint arm in patients suffering from stroke. The principle objective of the project was to test the hypothesis that the control of arm position is produced by changing the referent configuration of the arm and specifying the areas in hand and joint space in which muscle activation or co-activation occurs. The project is based on the λ-model of the equilibrium point hypothesis of motor control, which proposes that movements are realized by determining the referent configuration of the body in space. Ten healthy and 13 hemiparetic subjects participated in the study. The participants were asked to resist on externally applied load by matching their hand force with the force produced by two torque motors. After a random delay period the external load was suddenly removed and the subject's hand performed an unintentional transition to a new point in space. Generally, the patients preserved their ability to produce a referent configuration of the arm. However, the slopes of the double-joint invariant characteristic were steeper than those of healthy subjects which was accompanied by increased levels of cocontraction. Patients with stroke also had increased values of instability after unloading. In these subjects there was a prolonged time of stabilisation with an increased number of oscillations of the hand around the final position. Our major conclusion is that even though the basic mechanism of movement production may be preserved in patients with stroke, the central nervous system modifications and changes in biomechanical properties of the arm following stroke-related brain damage influence the central command, which results in increased arm instability.

Key words: instability, λ-model, motor control, double-joint movement, stroke, torque - angle interaction, arm, hemiparesis.

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Accord des coauteurs

Liste des coauteurs de l'article

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Titre de l'article : Control of double-joint arm posture in patients with unilateral brain damage.

Soumis : Experimental Brain Research

Coauteurs : Archambault P., Levin M., Feldman A.

À titre de coauteur de l'article identifié ci dessus, je suis d'accord pour que Pavel Mihaltchev inclue cet article dans son mémoire de maîtrise qui a pour titre : «Control of double-joint arm posture in patients with unilateral brain damage.»

[redacted]
Philippe Archambault — 20/3/03 — Date

[redacted]
Anatol Feldman — 26/3/03 — Date

[redacted]
Mindy Levin — 20/3/03 — Date

List of Abbreviations

ACA	Anterior Cerebral Artery
AG	Agonist
AN	Anconeus
ANT	Antagonist
BB	Biceps Brachii
BBT	Box and Blocks Test
BR	Brachioradialis
C command	Coactivation Command
CNS	Central Nervous System
CPG	Central Pattern Generator
CSI	Composite Spasticity Index
CVA	Cerebral Vascular Accident
D	Decrement of Decay
DP	Deltoideus Posterior
EMG	Electromyography
EP	Equilibrium Point
FM	Fugl-Meyer Assessment Scale
IC	Invariant Characteristic
IREDs	Infrared Emitting Diodes
LE	Lower Extremity
MCA	Middle Cerebral Artery

MVE	Maximum Voluntary Effort
PCA	Posterior Cerebral Artery
PM	Pectoralis Major
Q position	Actual Position
R command	Reciprocal Command
RC	Referent Configuration
R position	Referent Position
ROM	Range Of Motion
SD	Standard Deviation
SR	Stretch Reflex
TB	Triceps Brachii
TE	Torque at the Elbow
TP	Transition Point
TS	Torque at the Shoulder
UE	Upper Extremity

Dedication

I would like to dedicate this Master's thesis and to thank profoundly some special people who played crucial roles in my life, who supported me during difficult periods and who shared my moments of joy and disillusionment. This work would not have been possible without:

Dr. Mindy Levin – my director - a scientist with formidable experience, mentor and friend in difficult times.

My wife – we could not make it without her courage and love.

My parents and my sister – for their education, patience, understanding and constant spiritual support.

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I am very grateful to the other co-authors of the paper: Dr. Anatol Feldman who's ideas form the basis of part of this work and Phillippe Archambault (PhD) for his theoretical work and programming skills. Thanks to all the students and personal of the IRM. Especially I would like to thank:

Stella Michaelsen, Ruth Dannenbaum, Sheila Schneiberg, Valeri Goussev, Martin Foisy and Carmen Cirstea as well as all the members of the motor control lab at the IRM.

Chapter I. Literature review

The understanding of the deficits underlying movement disorders in patients following a cerebro-vascular accident (CVA) or stroke is a principal area of neuroscience research. The brain tissue is highly vulnerable to oxygen deprivation, which places this kind of pathology among the diseases with the highest rate of mortality - 23.8% of the patients with stroke aged between 55 and 64 years old (Ostbye et al., 1997). Stroke is a sudden disturbance of brain blood circulation and supply resulting in an acute onset of neurological dysfunction, with resultant signs and symptoms that correspond to involvement of focal areas of the brain (O'Sullivan, 1994). Two major circulatory impairments result in stroke. Strokes can be *ischemic,* the result of a thrombus, embolism or conditions that produce low systemic pressures. The resulting lack of cerebral blood flow disrupts cellular metabolism and leads to injury and death of tissues. Strokes can also be *hemorrhagic,* with abnormal bleeding into extra-vascular areas of the brain secondary to aneurysm or trauma. Hemorrhage results in increased intracranial pressures with injury to brain tissues. *Intra-cerebral hemorrhage* is caused by rupture of a cerebral vessel with subsequent bleeding into the brain. Primary *cerebral hemorrhage* (non-traumatic spontaneous hemorrhage) typically occurs in small blood vessels weakened by atherosclerosis producing an aneurysm. *Subarachnoid hemorrhage* occurs from bleeding into the subarachnoid

space typically from a saccular aneurysm affecting primarily large blood vessels. Developmental defects that produce weakness in the blood vessel wall are major contributing factors in the formation of an aneurysm. The main risk factors leading to stroke are hypertension, heart disease and diabetes. Symptoms vary according to the site of damage and type of stroke.

1. Sites of CVA

The middle cerebral artery (MCA) is the most common site of circulatory occlusion in stroke (O'Sullivan, 1994). It supplies the entire lateral aspect of the cerebral hemisphere (frontal, temporal, and parietal lobes) and sub-cortical structures, including the internal capsule (posterior portion), corona radiata, globus pallidus (outer part), most of the caudate nucleus, and the putamen. The most common characteristics of MCA involvement are contra-lateral spastic hemiparesis and sensory loss of the face, upper extremity (UE), and lower extremity (LE), with the face and UE more involved than the LE. Lesions of the parieto-occipital cortex of the dominant hemisphere (usually the left one) typically produces aphasia. Lesions of the right parietal lobe of the non-dominant hemisphere (usually the right hemisphere) typically produce perceptual deficits (e.g. unilateral neglect, apraxia, and spatial disorganization). Homonymous hemianopsia (a visual field defect) is also a common finding.

- The anterior cerebral artery (ACA) supplies the medial aspect of the cerebral hemisphere (frontal and parietal lobes) and sub-cortical structures, including the basal ganglia (anterior internal capsule, inferior caudate nucleus), anterior fornix, and anterior four fifths of the corpus callosum. The most common characteristic of ACA syndrome is contra-lateral hemiparesis and sensory loss with greater involvement of the lower extremity since the somatotopic organization of the medial aspect of the cortex includes the functional area for the lower extremity.

- The two posterior cerebral arteries (PCAs) arise as terminal branches of the basilar artery and each supplies the corresponding occipital lobe and medial and inferior temporal lobe. They also supply the upper brainstem, midbrain, and posterior diencephalon, including most of the thalamus. Occlusion proximal to the posterior communicating artery typically results in minimal deficits owing to the collateral blood supply from the posterior communicating artery (similar to ACA occlusion). Occlusion of thalamic branches may produce hemi-anesthesia (contra-lateral sensory loss) or thalamic sensory syndrome (a persistent and unpleasant hemi-body sensation). Occipital infarction produces homonymous hemianopsia, visual agnosia, prosopagnosia (inability to recognize faces), or, if bilateral, cortical blindness. Temporal lobe ischemia results in an amnesic syndrome with memory loss. Involvement of subthalamic branches may affect the subthalamic nucleus or its pallidal connections, producing a wide variety of deficits. Contra-lateral hemiplegia occurs with involvement of the cerebral peduncle.

- The basilar artery supplies the pons, the internal ear and the cerebellum. Complete occlusion results in quadriplegia, locked-in syndrome (the consciousness and sensations are preserved, the only possible movement is vertical gaze) and often death. The mortality is high and the survivors remain with severe dysfunctions.

2. Mortality rate, epidemiology

CVA is the third leading cause of death in North America (Brown et al., 1996). 38% of the victims die within 30 days of stroke onset and more than 50% die in the subsequent 5 years. Stroke victims make up a rather large population in contemporary health-care: 16.2 per 100,000 men and 24.4 per 100,000 women or 335,000 Canadians had a stroke in the period 1982 – 1991 (Mayo, 1996). Other studies (Heart Disease and Stroke in Canada, 1997) reported a rate of 0.8 for every 1000 individuals. In Canada the incidence is higher 1.5 / 2 per 1000 which corresponds to 50,000 victims every year, 60% of whom are women. Presently, there are 200,000 patients with stroke living in Canada and 36,000 of them live in Quebec. In the United States there are more than 2 million stroke patients. CVA is not only a medical problem but also a complex social problem involving patients and their family members. In addition, financial demands required for health support of the concerned population are significant and load considerably health-care budgets.

3. Motor deficit related to CVA

The most common motor sign associated with CVA is hemiparesis, a term used to describe the generalised disruption in the control of the contralateral half of the body. Hemiparesis results from different pathophysiological mechanisms at the central and peripheral levels. A stereotypical distribution of increased tone - flexors of the arm and extensors of the leg, and muscle weakness in both agonist and antagonists has been observed (Wernicke, 1889; Mann, 1896). Gowers (1893) reported more accentuated weakness in the distal parts of the limbs which may explain the earlier recovery of more proximal segments. Recently (Bohannon, 1991), showed that the weakness in the arm extensor muscle groups is not necessarily related to the increased tone in their antagonists, and the range of motion (ROM) is mostly affected by the strength of the former. On the other hand, Levin and colleagues (2000) showed that the mechanism underlying range of motion and strength deficits may be problems in the regulation of stretch reflex (SR) thresholds in elbow flexors and extensors. Also, dysfunction in masticatory and respiratory musculature may also be present due to their bilateral representation in the cortex (Gowers, 1893). At the same time, CVA could affect, to a certain degree, the ipsilateral hemicorps. Colebatch and Gandevia (1989) reported reduced force in the ipsilateral upper limb with an opposite distribution on the contralateral side, i.e. the proximal muscles were weaker than the distal ones, results confirmed by Bohannon and

Andrews (1995). Gransberg and Knutsson (1983) recognised three principal elements of hemiparesis: increased stretch-reflex excitability, decreased voluntary muscle activation and modified agonist / antagonist muscle co-activation. Such alterations can be observed as variations of the electromyographic (EMG) activity during physical performance. For instance, at the elbow, prolonged transition time between flexion and extension movement (Hammond et al., 1982), co-contraction of agonist-antagonists during voluntary movements with impairment of agonist recruitment and antagonist inhibition (Hammond et al., 1988), absence of co-contraction during isometric contractions (Tang and Rymer, 1981) and abnormal spatial recruitment patterns (Bourbonnais et al., 1989; Dewald et al., 1995) have been reported. Another important component of hemiparesis is an increased muscle resistance to passive stretch due to intrinsic changes of the muscle structure (replacement of muscle fibers with connective tissue – Given et al., 1995) or by the impairment of central mechanisms governing reflex activity, a condition known as spasticity. Wiesendanger (1991) defined spasticity as a motor disorder developing gradually and caused by partial or complete loss of control of the supraspinal levels on the spinal cord. It is characterized by modified activation patterns of motor units depending on central and sensory influx, which results in co-activation, associated movements and abnormal postural control. Spasticity is difficult to characterise and quantify. Generally, one component of spasticity may be elicited by stretching of the passive muscle. The mechanisms underlying spasticity are not completely

understood, however several physiological changes may partially explain this phenomenon:

- ***Increased motoneuronal excitability*** – activation threshold decreases, so that motor responses can be provoked with sub-threshold stimuli (Burke, 1988);
- ***Collateral sprouting*** – afferents make new synapses on vacant efferent sites (Bishop, 1977), which increases sensory input to motoneurons;
- ***Change in pre-synaptic and reciprocal inhibition (Burke, 1988; Katz and Rymer, 1989).*** Because of the changes at supraspinal levels, the central control of peripheral influx is altered. This means that sensory information from Ia fibers is not organized by the CNS. For the same reason many of the Ia interneurons are no longer under the control of the CNS leading to impaired reciprocal inhibition.

The role of spasticity in the production of the motor disorder is not completely understood since movements are still disrupted even after medical treatment of spasticity (Watanabe et al., 1998).

4. Motor control theories

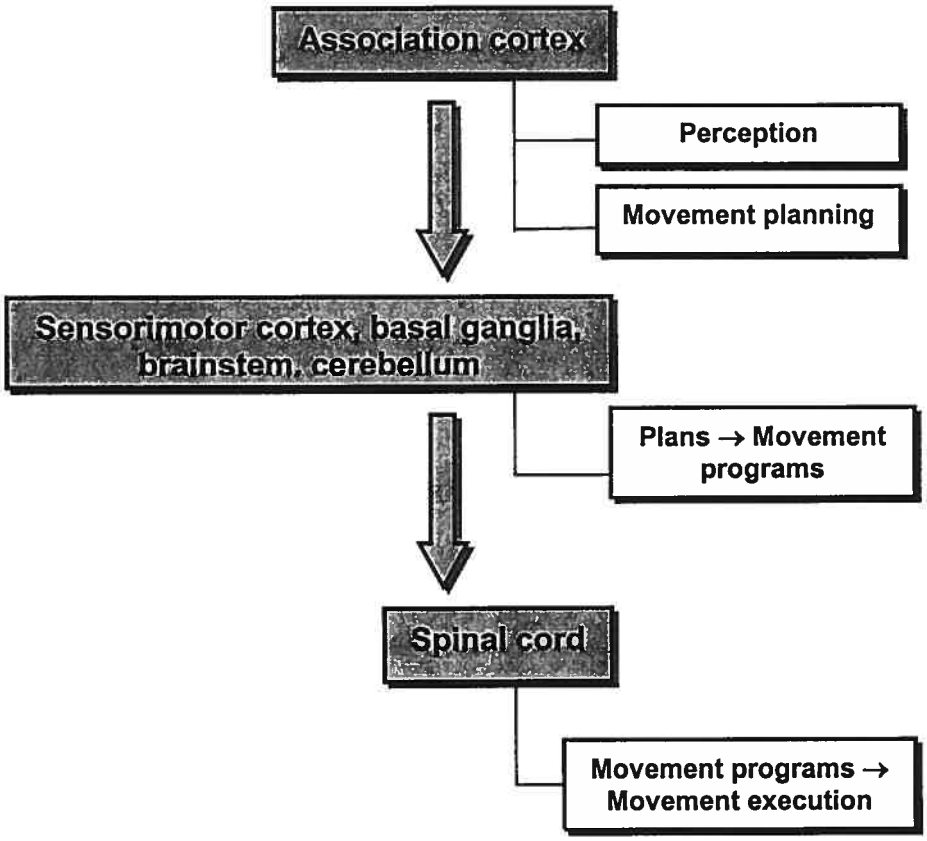
Motor control deficits as well as normal motor functioning can be described in the framework of different motor control hypotheses that integrate physiological, biophysical and biomechanical approaches. Notable scientists have influenced the development of this area of

science. Existing knowledge of postural and equilibrium control has evolved and passed through several stages, which can be summarised into three broad theories:

4.1 Reflex theory introduced by Sherrington at the beginning of the 20th century, considers reflexes as a basis for movements and neuro-muscular co-ordination as a chain of reflex responses strung together. Sensory input was considered as necessary for movement initiation. However, the theory fails to explain voluntary movement in the absence of external stimuli, ballistic movements as well as different patterns of motor responses evoked by the same stimulation.

4.2 Hierarchical theory (Hughlings-Jackson, 1932; *Fig.1)*, proposes that control of movement "flows" in a top-down direction and it is distributed among three different levels of the nervous system: The *upper* level, located in the association cortex participates in the evaluation of perceptions and planning of response strategies. The *middle* level, located in the sensorimotor cortex, basal ganglia, brainstem and cerebellum converts motor plans into motor programs (motor commands, which produce coordinated motor actions). The *lower* level located in the spinal cord is where the motor programs are executed and transferred to the corresponding muscles. Brown (1911) proposed the existence of motor programs (called central pattern generators – CPGs) in the spinal cord and the brainstem which control reflex activity. CPGs are groups of neuronal circuits controlling habitual, repetitive movements requiring little

attention or concentration. Some of the CPGs exist congenitally while others are initiated and formed by external stimuli during motor learning



- Fig. 1 -

Description of distribution of the motor control in the CNS according to hierarchical theory.

and maturation. The hierarchical theory assumes that the control of reflex activity is integrated into voluntary movements during maturation and is taken over by lower levels of the CNS.

A pathologic event such as a CVA could release primary reflex activity evidenced by the appearance of the crossed flexion reflex, asymmetric reflexes, etc.

4.3 Systems theory (Bernstein, 1967) assumes that the CNS does not function as a fixed system as had been previously thought, but rather, central commands operate via numerous feedback loops at various levels. The basic concept of the theory is that units of the CNS are organised around specific task demands (task systems). Thus, the CNS as a whole can be involved in complex task execution but many simple and repetitive movements and activities (such as walking for example) are triggered and "monitored' by the higher levels while being maintained at the lower levels (such as the spinal cord). The *system's theory* suggests a general explanation of movement production on which most of the contemporary motor control models are based. Whereas the basic principles of motor control are outlined theoretically, the questions of how exactly movement is designed, produced and controlled remain unsolved. These different aspects of motor behaviour are partially explained by current theories and models.

- ***Velocity control model*** – According to this model, there are two types of motor strategies: *velocity dependant* and *velocity independent*. Gottlieb et al. (1989) proposed that the *velocity independent strategy* is applied during execution of rapid and precise movements. It is used when there is no velocity constraint and its general feature is modulated duration of neural pool excitation. The v*elocity dependent strategy* is applied when velocity is important for task achievement. The modulation of the excitation of agonists and the latency of antagonist

activation characterises this control strategy, which has been largely described for the single-joint system.

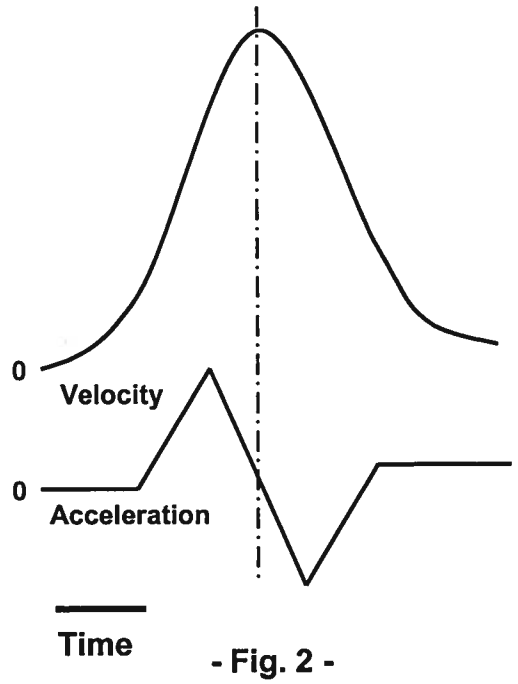
- ***Force control model*** - the central idea of this model is that for the production of voluntary movements, the CNS calculates the forces necessary to execute the task. Gottlieb et al. (1989, 1990) have shown that there is a direct correlation between elbow torque, kinematics, and EMG activity around the elbow joint. The force control model is based on two notions: inverse dynamics model and forward internal model.

- inverse dynamics model (Hogan, 1990; Kawato, 1999) – This model is based on the assumption that fast and coordinated arm movements cannot be executed under feedback control, since the feedback is too slow to influence the ongoing movement. In the inverse-dynamics model, movement trajectories are planned first in kinematic coordinates and then transformed into the required forces and torque. The CNS calculates the coordinates of each point of the desired movement and based on this data, in an inverse way, estimates the necessary torques and forces, to reach these points.

- feedforward internal model - in order to achieve the desired movement, the CNS uses anticipated sensory information without using long-loop sensory feedback (Jordan and Rumelhart, 1992). Under this hypothesis, the CNS learns internal models that simulate the dynamics of the musculoskeletal system and external environment and generate the required feedforward motor commands (Bizzi and Mussa-Ivaldi, 1998; Kawato et al. 1987). In feedforward mode the elements

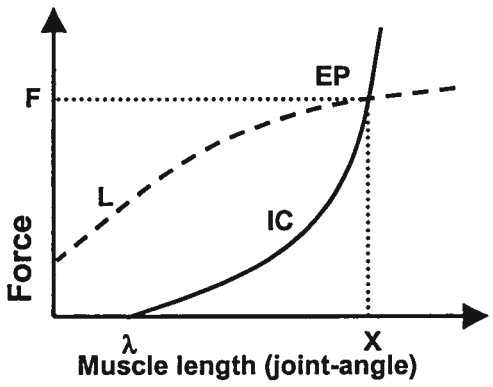
forming the motor action are specified based on previous experience and not on immediate sensory feedback. This model has been used to describe the control and learning of voluntary movement (Kawato et al., 1987).

- ***Minimum jerk model*** – This model is based on the assumption that the neuro-muscular system tends to produce a linear trajectory between the initial and final position (Hogan, 1984; Bizzi et al., 1984) and the velocity profile of the endpoint is bell-shaped *(Fig. 2).* Flash and Hogan (1985) showed that with repetition, the CNS optimises the movement, a phenomenon called "maximizing smoothness". With practice, movement becomes more direct and precise. Once the target is reached, the CNS is informed about any error in position and new coordinates are assigned for the next movement. Another distinguishing feature of the model is that the CNS plans movement in terms of external space and not in terms of internal joint space. This model also assumes burdensome calculation and delays in motor responses. Recently, Gribble and Ostry (1996) showed



- Fig. 2 -

Bell shaped velocity profile with acceleration profile for the end-point during movement. Adapted from Hogan N (1984) An organizing principal for a class of voluntary movements. J Neurosci 11: 2745-2754.



-Fig. 3-

Example of torque/angle interaction (IC) and a single equilibrium point.
Adapted from Feldman A (1986) Once more on the Equilibrium - Point Hypothesis (lambda Model) for Motor Control. J Mot Behav 18: 17-54

that minimal spatial deviation of the hand trajectory could entirely be a property of the biomechanical system without implication of the CNS.

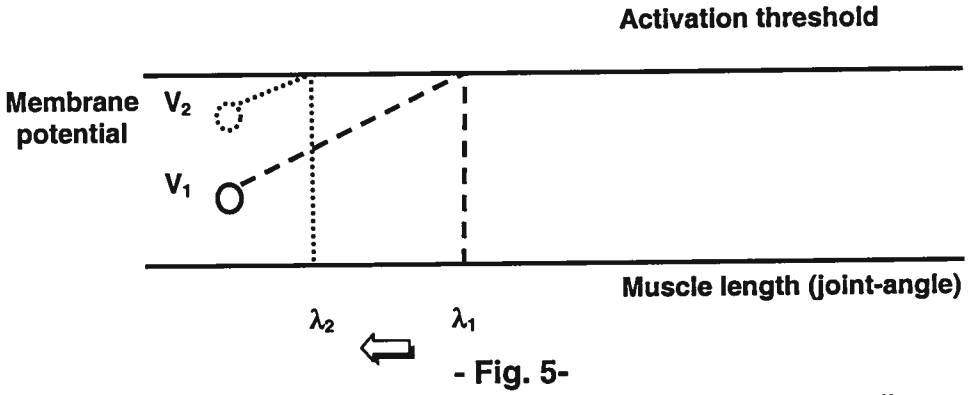
- ***Equilibrium point (EP) hypothesis.*** The α version (Bizzi et al., 1984) of the EP hypothesis suggests that the CNS directly controls electromyographic patterns and thus movements by programming the activity of α-motoneurons. This version proposes that for movements made at moderate speed, the CNS specifies a series of equilibrium positions throughout the movement, thus creating a virtual trajectory of the movement. However the authors attribute movement control entirely to the α-motoneurons. The role of sensory feedback and γ-motoneurons is to control the stiffness around the end-point. Since the stiffness is related to muscle force, this version suggests that CNS preprograms muscle force. It is more likely that force is generated as a concequence of the interaction of the human body with the external environment, which is consistent with the λ version of the EP hypothesis. Since our work is based on the λ model of the EP hypothesis, it will be explained in greater detail below.

4.3.1 λ-model of the equilibrium point (EP) hypothesis. The equilibrium point model hypothesizes that the CNS controls the limbs and trunk by specifying a series of equilibrium positions aligned along the desired trajectory (Asatryan and Feldman, 1965; Feldman, 1966). The EP is a combination of the muscle force 'F' (torque) and length 'X' (or joint angle) where opposing forces (muscle force, F, and load force, L) are at equilibrium and no movement is produced (*Fig.3*). During his experiments in 1966, Feldman observed that if the subject maintains a position of the hand against an external load (i.e. elbow flexion against a weight) and the external weight is suddenly partially removed, an involuntary flexion movement will occur. If the hand is loaded again with the same weight, the hand returns to the initial position. Feldman concluded that if a given central command is held constant by asking the subject not to make corrections ("Do not intervene" paradigm), there would be an infinite number of equilibrium points in space along a specified curve for the arm related to different final external loads. These EPs draw a curve called an invariant characteristic (IC in *Fig. 3*). It is called invariant since the underlying assumption is that the curve is produced under conditions in which the central commands do not change. The IC describes the force (torque) / length (angle) relationship and its slope which represents the stiffness of the system. As a consequence, the EP can be represented as the point of intersection between the IC and the load characteristic (L) so that EP = (X, F), where X is the position (muscle length or joint angle) and

F is the muscle or load force at the equilibrium state. To a certain degree, the λ-model significantly releases the CNS from the necessity of making intense calculations. According to the model, the variable controlled by the CNS is the threshold of activation of motoneurons or the tonic stretch reflex: λ (Feldman and Levin, 1993). λ can be expressed in terms of muscle length or angular degrees. The activation threshold is the intersection point of the IC with the muscle length / joint angle line (λ on *Fig.3*) where the external load is 0. Thus, the nervous system assigns a

- Fig. 4-

Silent and active zones of muscle activity set according to specification of the activation threshold.



- Fig. 5-

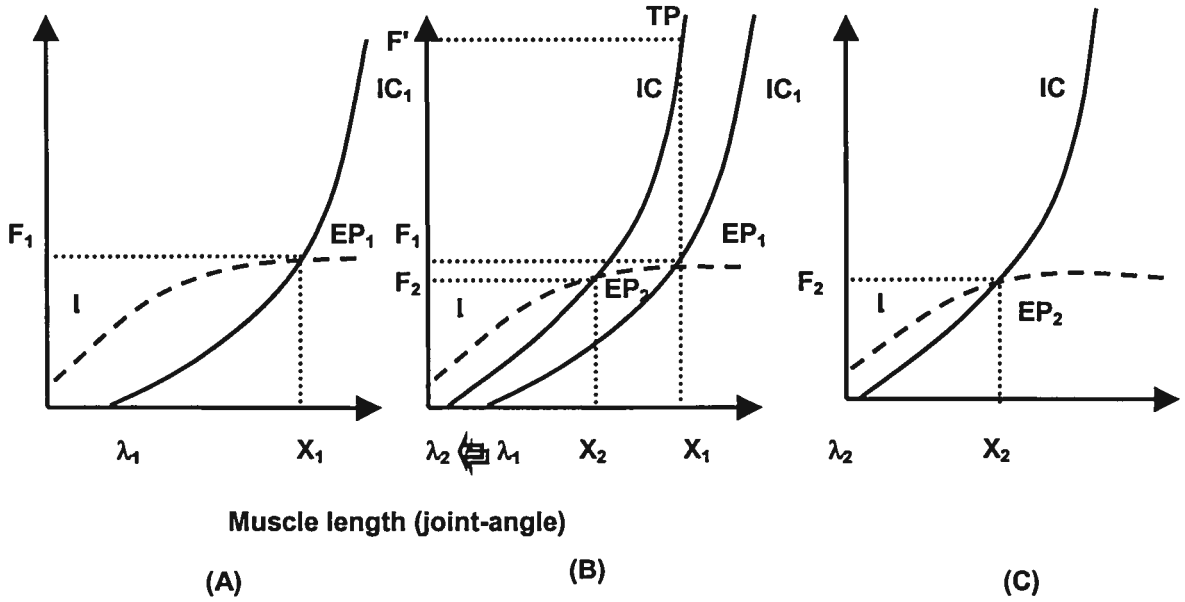
Displacement of the membrane potential and its corresponding muscle activation threshold.

Adapted from Feldman A (1986) Once more on the Equilibrium - Point Hypothesis (lambda Model) for Motor Control. J Mot Behav 18: 17-54

muscle length, beyond which muscular activation begins. In *Fig. 4,* if we have a given threshold muscle length λ, all muscle lengths shorter than this length (to the left of λ) will belong to the silent zone were no activation

occurs. A stretch bringing the muscle length beyond the assigned limit will cause the activation of motor units. Hence, the condition of recruitment is: **$X - \lambda > 0$.** Electrophysiologically, a change in λ results in a shift of the membrane potential (hyperpolarization or depolarization) bringing it further away from or closer to the threshold of the membrane (*Fig. 5*). The CNS changes motoneuronal excitability by altering the distance between the current level and the activation level of the motoneurons.

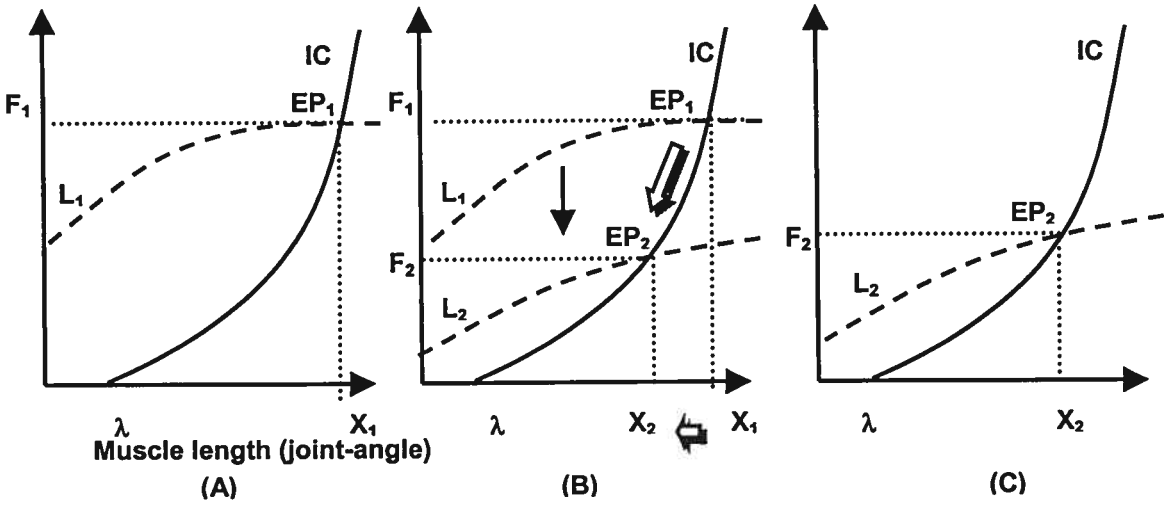
Changes in λ result in shifts of the IC and EPs which in turn produce voluntary movement. Thus the CNS assigns a new muscle length (different from the initial one) at which external and muscle forces will be at equilibrium. The transition from one position to another occurs gradually due to changes in λ. Muscle forces and EMG patterns originate from the deviation between the initial and desired position. This is illustrated in *Fig.*



- Fig. 6-

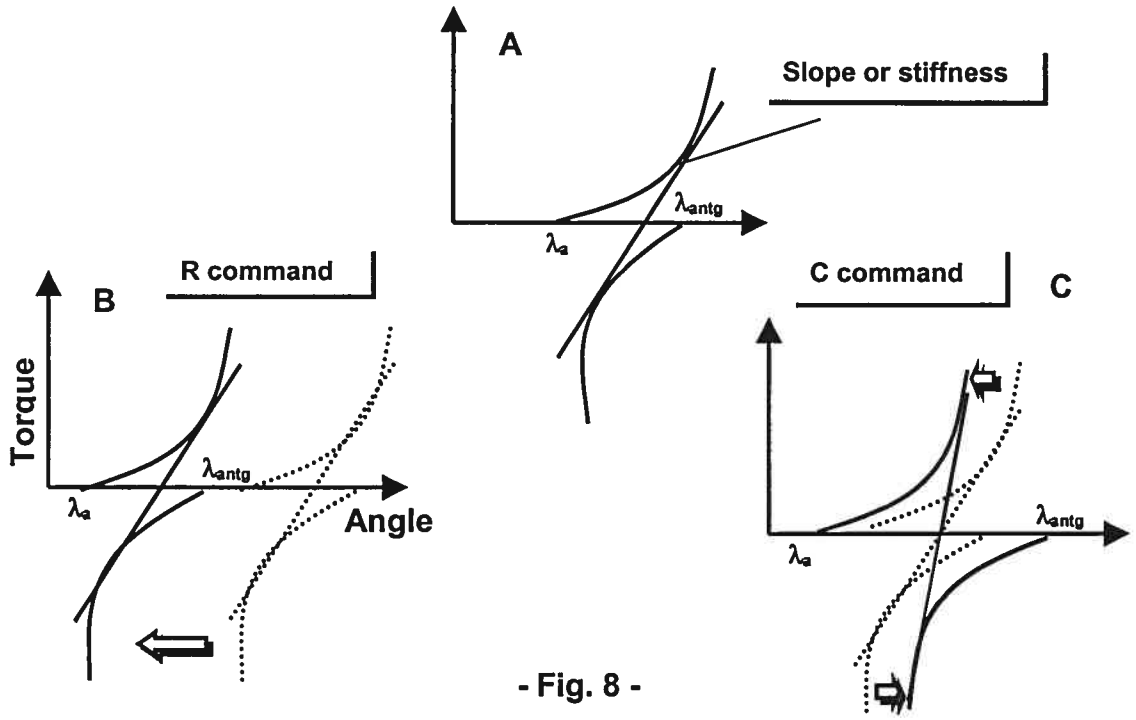
Voluntary shift of the threshold of activation with change of the position of the IC. Adapted from Feldman A (1986) Once more on the Equilibrium - Point Hypothesis (lambda Model) for Motor Control. J Mot Behav 18: 17-54

6A the load (L) and muscle force (F_1) are at equilibrium at point EP_1 (F_1 - L = 0) at muscle length X_1. When the central command changes and the IC is shifted, the muscle force at the initial position X_1 becomes greater than the load (*Fig. 6B*, $F^I > F_1$ at the transition point TP). The initial position becomes unstable because it is no longer at equilibrium, which results in movement to the new equilibrium-point, EP_2 (*Fig. 6C*). In this case the muscle will contract and the limb will move to another position in order to re-establish the balance between the load and the force at EP_2 (*Fig. 6C*; Feldman, 1986). The involuntary movement already mentioned in Feldman's experiment could be explained in a similar way. If the central command is kept constant but the external load is changed (for example – diminished as in *Fig. 7*) once again there will be a loss of equilibrium between the initial position and the new external conditions. In this case, the system will perform an involuntary movement in order to reestablish equilibrium – the muscle will decrease in length to balance the load at a new position (*Fig. 7B and 7C*).



- Fig. 7 -

Change of the position of the EP following changes in the external load. Adapted from Feldman A (1986) Once more on the Equilibrium - Point Hypothesis (lambda Model) for Motor Control. J Mot Behav 18: 17-54



- Fig. 8 -

Central specification of the R command (panel B) by shifts of the activation thresholds for agonist (λ_a) and antagonist muscles (λ_{antg}) in the same direction resulting in a change in position of the joint.
Panel C - specification of the C command by shifts of the activation thresholds in opposite directions resulting in increased of the joint stiffness.

Feldman and Levin (1993) outlined that the control variable (CV) λ is not only dependent on the external environment and sensory feedback but also on a central component specified by the CNS independent of sensory feedback, or: $\Delta\lambda = \lambda_{cv} + \lambda_{feedback}$. This statement suggests that the motor response may be influenced by either central or peripheral influx or both. At the single-joint level, the CNS controls thresholds of activation for both agonist and antagonist muscle groups acting around the joint. The nervous system uses two main types of commands to control the movement: The R (*reciprocal*) command, decreases the activation

threshold (facilitates) of the agonists and increases the threshold (inhibits) of the antagonists *(Fig. 8B)* or vice versa. The C (*co-activation*) command, moves both thresholds in order to increase the activation in all muscle groups around the joint *(Fig. 8C)*.

While the λ - model describes in detail and very explicitly single-joint movements, the explanation of more complicated movements involving many uni and bi-articular muscles remains less clear. Difficulties in resolving the problems of multi-joint movements were first described by Bernstein (1967). He first introduced the idea of motor "redundancy" and multiple degrees of freedom in neuro-muscular systems. The problem arises from the lack of an explanation of how the CNS finds a unique combination of joint movements to perform a specific task from amongst an infinite number of possible combinations. A multi-joint explanatory model (Berkinblit et al., 1986) proposes that the CNS calculates the final equilibrium positions of the joints based on their initial positions and assigns these positions separately for each joint. Others (Flash, 1987, 1989; Latash, 1993) suggest similar multi-joint principles based on the EP hypothesis. In this formulation, the EP hypothesis postulates that the system produces movements by specifying the equilibrium trajectory of the effector. In other words, the CNS considers the movement only of the endpoint. According to this hypothesis, multi-joint arm movements are produced by gradually shifting the hand equilibrium positions from the initial to the final position. In the two-joint model, for example, the

trajectory described by the EP of the hand is a straight line (Flanagan et al., 1993). The model has been further elaborated by the introduction of the concept of the frame of reference for muscle activation. A frame of reference (a positional activation threshold for the whole body) principle was introduced (Feldman and Levin, 1993). According to this principle, motoneuronal recruitment is governed by shifts of this frame of reference (Feldman and Levin, 1995; Feldman et al., 1998; Pigeon and Feldman, 1998; Archambault et al., 1999; Ghafouri and Feldman, 2001). The frame of reference hypothesis is an extension of the λ - model and considers that a common threshold configuration is produced by the summation of all λs of the body. This configuration is considered as a geometrical representation of the body that forms a referent (or virtual) body configuration RC. At the same time, external forces (such as gravity), deflect the body away from the threshold RC configuration to an actual configuration Q. The difference between RC and Q will generate EMG patterns, muscle forces and joint movement. Thus the nervous system does not directly control EMG and forces, which are seen as emergent properties of shifts in RC. Voluntary movements are produced by specification of a new RC configuration, which again will cause a deflection and will trigger modifications of kinematic and dynamic characteristics. On the other hand matching of RC and Q would cause a decrease in electromyographic activity – a condition called a global EMG

minimum that has been observed and tested in fast repetitive movements (Lestienne et al., 2000; Coté et al., 2002; Ghafouri et al., 2002).

In the double joint system (in the present project - shoulder and elbow) the common activation angular threshold, λ, of the two is composed of the λs of each individual joint:

$$\lambda_{common} = \lambda_{elbow}, \lambda_{shoulder}$$

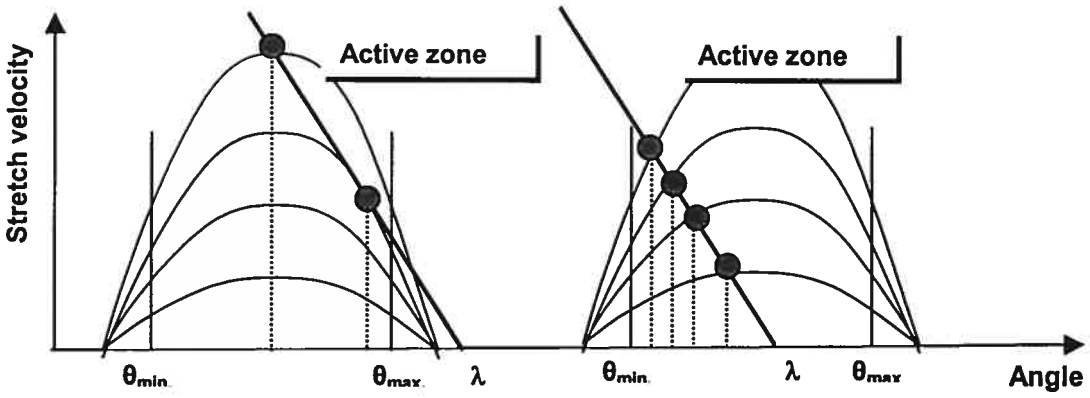
These λ_s are related to the length thresholds of the muscles surrounding the joints. While the threshold of the single-joint muscles is related to the position of the joint that it serves, in the case of the double joint muscles (such as biceps brachii), the threshold length is obtained by a specific combination of position of the two joints (Feldman, 1998; Flanagan et al., 1993). The common joint angle determined by the angles of each joint serves as a frame of reference that is controlled by the central R command. Thus the R command represents the threshold for recruitment of the muscles around both joints.

The feasibility of the λ-model and its capacity to explain a number of problems (e.g., optimisation in motor learning and patterns of reciprocal inhibition and co-activation), make it an attractive and convenient model in studies of motor control and in disordered motor control.

5. Motor deficit explanations based on the λ-model of motor control

The λ-model has been used not only to explain the control of movements in the healthy nervous system but also to explain disordered motor control. For example, Levin and Feldman (1994) proposed that motor control could be disrupted by the inability of the nervous system to specify and control the activation thresholds of motoneurons that is manifested by deficits in several movement parameters and characteristics:

- ***Change in velocity sensitivity of the threshold*** – In *Fig. 9* are presented velocity profiles of stretches of the passive elbow flexors at different velocities. In healthy subjects no responses to slow stretches are observed and only high stretch velocities (above 300°/s) may elicit reflex EMG activity in the stretched muscle (circles *Fig. 9A*). The joint angle at which muscle activation arises in response to stretch at a given velocity represents the dynamic stretch reflex threshold for that velocity. By extrapolating a line through the dynamic thresholds, the tonic SR threshold, which by definition occurs at zero velocity, can be determined. Previous studies have shown that at rest, the tonic SR threshold lies outside the biomechanical range of the joint in healthy individuals (Levin et al., 2000) and full voluntary relaxation can be achieved at all angular ranges of a given joint ($\theta_{min.}$ and $\theta_{max.}$).

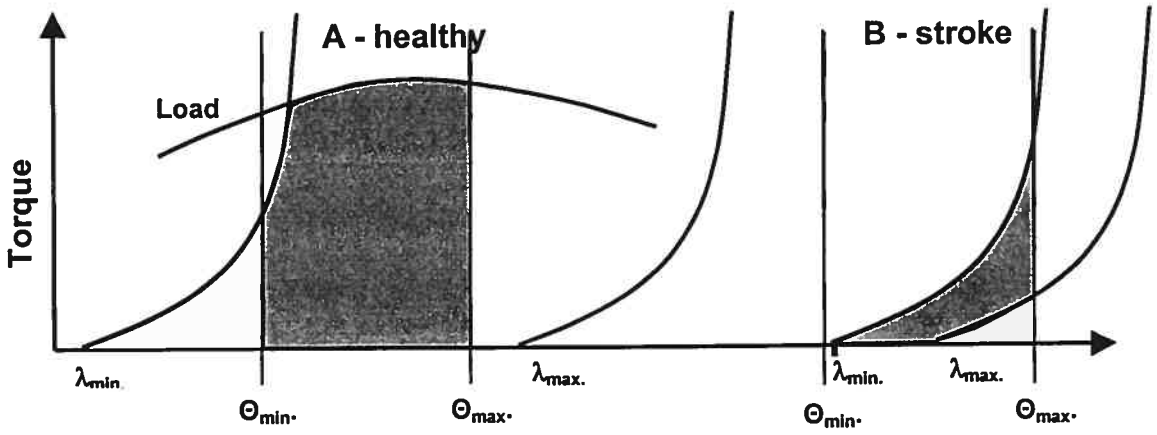


-Fig. 9-

Zone of regulation of the stretch reflex in one healthy (panel A) and one stroke subject (panel B) with dependence of the muscle activation on the velocity of the stretch in stroke patient (panel B). Adapted from Levin MF, Selles RW, Verheul MH, Meijer OG (2000) Deficits in the coordination of agonist and antagonist muscles in stroke patients: implications for normal motor control. Brain Res 853: 352-69.

It has been suggested that the intact nervous system produces movements by regulating the tonic stretch-reflex thresholds throughout and beyond the biomechanical range of the joint. On the other hand, in patients with hemiparesis, stretch at even low velocities will evoke muscle activation at smaller joint angles (*Fig. 9B;* Levin et al., 2000). This suggests that, at rest, the muscle is unable to relax since the muscle activation threshold lies within the biomechanical limits of the joint, which has been demonstrated for elbow flexors and extensors in adult stroke patients (Levin et al., 2000) and in children with cerebral palsy (Jobin and Levin, 2000).

- ***Decrease in range of regulation of λ*** - as mentioned above the condition for eliciting muscle activation is that the muscle length should be greater than the threshold length specified by the CNS. The activation threshold ($\lambda_{min.}$ and $\lambda_{max.}$ in *Fig. 10A*) should be freely regulated within and

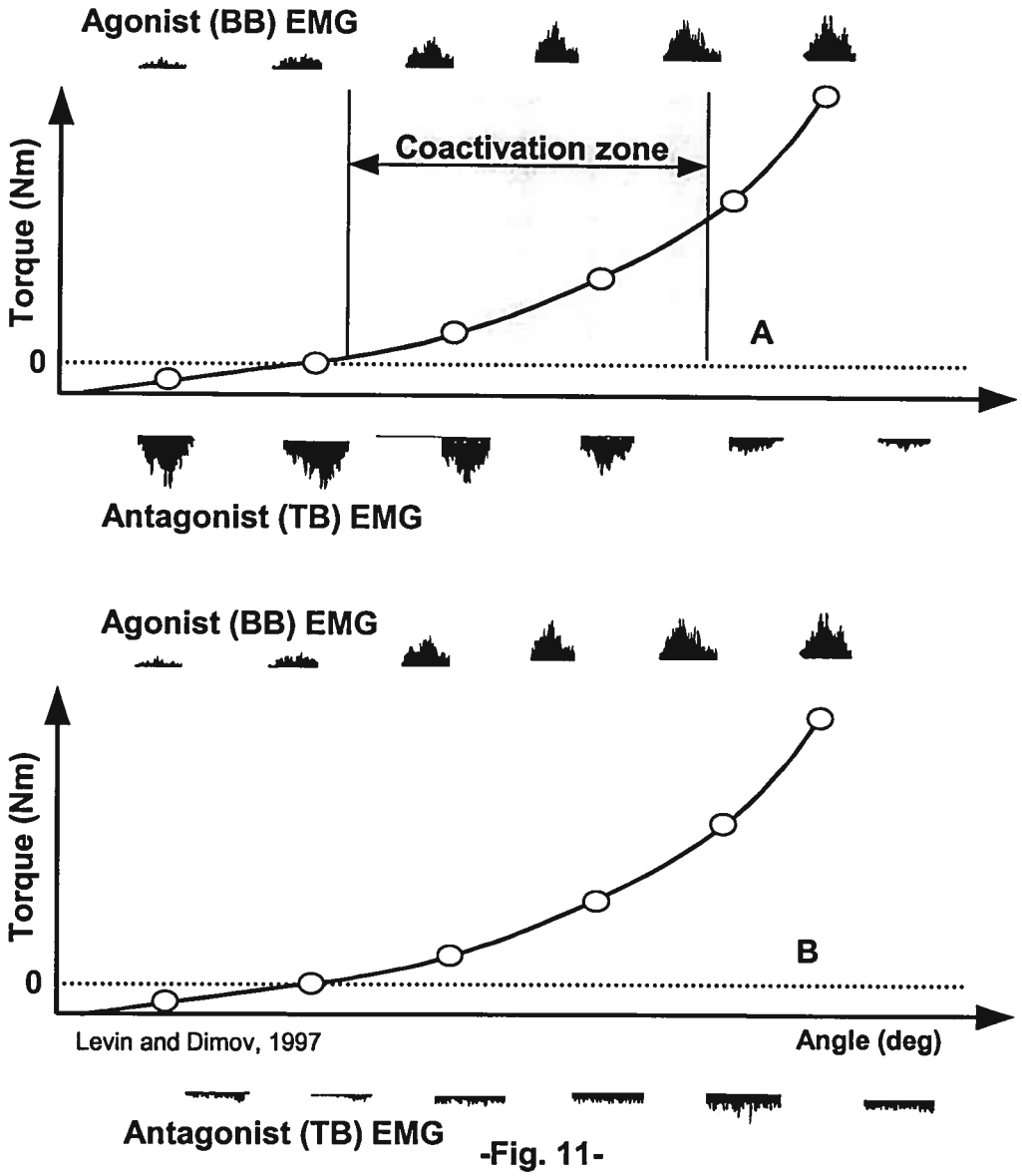


-Fig. 10-

Normal range of regulation of threshold of activation of a single muscle in healthy subject (panel A) and decreased range of regulation in patient with stroke. Adapted from Levin MF, Selles RW, Verheul MH, Meijer OG (2000) Deficits in the coordination of agonist and antagonist muscles in stroke patients: implications for normal motor control. Brain Res 853: 352-69.

beyond the biomechanical limits of the joint – a condition necessary for full control of movement (grey zone on *Fig.10A*). In patients with stroke, displacement of the threshold is restricted (*Fig. 10B*), which results in an inability to initiate voluntary movement, the presence of spasticity or abnormal EMG patterns in the zones outside of $\lambda_{min.}$ and $\lambda_{max.}$ (*Fig. 10B*). As a consequence, the controlled zone is diminished (shown for agonist / antagonist muscle groups at the elbow by Levin et al., 2000).

- ***Inability to specify and regulate C commands in an appropriate way.*** Levin and Dimov (1997) described deficits in the specification of C commands in stroke patients leading to impairments in movement stability at the elbow. With their arm in a single-joint manipulandum attached to a torque motor, participants were asked to oppose on external load with their elbow flexors. Then, the load was completely or partially removed without warning and the subjects were instructed not to intervene to this perturbation. Thus they presumably kept their central commands constant. As a consequence of the unloading, the elbow made an involuntary flexion and the hand moved to a new position in space. The EMG activity and the oscillations of the hand around the final position were recorded. ICs of one healthy (A) and one hemiparetic (B) subject along with the muscle activity for the agonist / antagonist (BB / TB) pair around each final position are shown in *Fig. 11*. In the healthy subject, for the unloaded agonist muscle (BB), the muscle activity diminished with increasing unloading, while at the same time, in the antagonist TB, muscle activity increased. The activity of both muscles formed a coactivation zone presumably specified by a C command, where both muscles were active at the same time. On the other hand, the activity of the antagonist TB was not modulated in most of the patients with stroke. This suggests that the muscle activation zones were not correctly specified in a feedforward manner in patients with stroke, which leads to postural instability.



-Fig. 11-

(Reprinted from Brain Research, date of permission 11 February, 2003)

Presence of coactivation zone for agonist/antagonist muscle pair in one healthy subject (panel A) and absence of such zone in stroke subject.

Chapter II. Rationale

Despite recent progress, the explanation of movement is still unresolved. There are controversies about which parameters of movement are controlled by the CNS and how the CNS interacts with the external environment. In addition, there is still some lack of consensus on the use of terminology. The lack of a complete model of normal motor control hampers the understanding of the motor deficits following an event like CVA. On the other hand, investigation of the motor deficit in the context of a model of motor control would permit the clarification of which elements of movement are lost and no longer controlled by the nervous system. The study of movement in patients with stroke permits us to identify such motor elements. Most of the movement control studies in patients with stroke concentrate on the characterisation of deficits in voluntary movement production. Also, many studies have focused on global motor outcomes in patients with hemiparesis such as, abnormal muscle synergies (Brunnström, 1970), changes in spatiotemporal organisation of hand movement manifested in alterations of reaching and grasping strategies (Roby-Brami et al., 2000), trunk compensatory strategies (Cirstea and Levin, 2000; Michaelsen et al., 2001; Levin et al., 2002), reduction in the ability to independently activate muscles out of the pathologic synergies (Reinkensmeyer et al., 2002) etc. Until now few studies have tried to elucidate the physiopathology underlying the disruption in movement and

the postural instability in the arm in post-stroke patients in the context of a motor control model. As a first step in this direction Levin and Dimov (1997) and Levin et al. (2000) investigated motor deficits in a single-joint system according to the λ model. In the present study we extended this approach to more relevant double-joint movements. An important objective of our work was to verify the possibility that patients with hemiparesis preserve the ability to specify a referent configuration of the arm (for double-joint model only) and to control movement through R and C commands. The results of this study may have important implications for the understanding of the motor control deficits following damage to the central nervous system.

Chapter III. Overall objective of the project:

The objective was to test the hypothesis that the control of arm position is produced by changing the virtual (referent) configuration of the arm and specifying the areas in hand and joint space in which muscle activation or co-activation occurs.

The specific objectives were:

- To describe the biomechanical interaction between the multi-segmented arm and external environmental forces;
- To analyze how these interactions are modified following voluntary changes in central commands;
- To investigate the differences in the control of arm movements between healthy and hemiparetic subjects (recruitment of muscle groups – silent zones and zones of activation, deflection of referent position).

Chapter IV. Methodology

1. Subjects

Thirteen patients with stroke (mean age: 51.9 ± 12.1 years) and 10 healthy subjects (49.0 ± 9.0 years) participated in the study after signing an informed consent form approved by the Ethics Committee of the Rehabilitation Institute of Montreal. The patients had right spastic hemiparesis due to cerebro-vascular accident (CVA) in the left hemisphere at least 6 months previously. They had full passive range of movement at the shoulder, elbow and wrist, partial control of the arm and no severe apraxia or comprehensive aphasia. We used Fugl-Meyer assessment score and the patients' clinical files in order to confirm this. Those with bilateral stroke, pain in the arm or trunk, dysmetria and static or dynamic tremor were excluded. Patient demographic and lesion location information are presented in *Table 1*. The healthy group consisted of right-handed individuals with no sensory or motor impairments or orthopedic problems affecting the arm or trunk. Those with uncorrected visual disturbances were excluded – i.e. patients who voiced complaints of disturbed vision and did not wear glasses.

The experimental session consisted of two parts: determination of maximum voluntary effort (MVE) and an unloading experiment. In addition, patients underwent a clinical evaluation to determine the sensorimotor status of their affected arm.

-Table 1- Demographic data and result of maximal voluntary effort (MVE) testing and clinical testing in hemiparetic patients. Demographic data and MVE results in healthy subjects are given for comparison.

	Age	Sex	Site of lesion	Duration[1]	CSI[2](/16)	FM[3] (/66)	BBT[4](R/L)	MVE PUSH (N)	MVE PULL (N)
1	58	F	frontal	41	11	10	0/76	33.3	16.7
2	52	M	temp.-fronto-parietal	48	7	14	0/60	29.6	19.7
3	61	M	ant.-temporal-parietal	36	11	20	0/65	37.73	27.87
4	45	M	fronto-parietal	28	9	22	7/56	123.3	98.64
5	55	M	ant.-temporal-parietal	47	4	30	0/50	24.66	22.19
6	42	F	Fronto-parietal	36	4	38	13/65	51.05	34.52
7	49	F	ant.-temporal	18	9	39	30/56	23.06	18.13
8	41	M	basic artery	28	5	52	21/75	73.98	64.12
9	51	F	parietal-subcortical	72	7	56	51/65	59.2	44.4
10	25	F	fronto-parietal	46	7	58	28/79	35.8	24.7
11	69	M	Parietal	36	6	60	61/75	96.2	107.3
12	69	M	parietal	47	3	61	59/61	120.8	86.3
13	58	M	temporal	60	4	65	55/61	64.1	64.1
MEAN	**51.9 ± 2.1**				**6.8 ± 2.6**	**40.4 ± 19.6**	**23.1 ± 25.0 60.2 ± 20.1**	**59.4 ± 4.9**	**48.4 ± 32.3**
HEALTHY	**49.0 ± 9.0**							**82.6 ± 33.2**	**79.2 ± 51.7**

[1]Time since the AVC in months [3]Fugl-Meyer score
[2]Composite Spasticity Index [4]Box and Blocks Test

2. Clinical evaluation

Patients with hemiparesis were evaluated clinically prior to beginning the experiment by a qualified physiotherapist using a test battery consisting of three measures:

A) Fugl-Meyer assessment scale (Fugl-Meyer et al., 1975) is an objective and reliable impairment scale of motor ability and reflex function. Since the focus of our study was on upper limb motor impairments, we used the arm and hand section of the test, having a maximum score of 66 points corresponding to normal function. The test consists of 7 sections evaluating reflex activity, flexion and extension movement synergies, the ability to perform isolated movements, functional activity of the wrist and hand, coordination and speed of arm movements. According to this scale, 6 (8 to 13) patients had mild motor impairment with scores ranging from 10 to 39 and 7 (1 to 7) patients had moderate to severe impairment with scores from 52 to 65.

B) Composite Spasticity Index – CSI (Levin and Hui-Chan, 1992) Clinical assessment of spasticity was comprised of biceps-brachii tendon jerks, resistance to passive elbow extension applied by the examiner, and the amount and duration of wrist clonus. The evaluation is done using 4-point scales while the one for resistance was doubly weighted since this measure most closely resembles tone. The three scores were summed and total scores ranging from 5-9, 10 -12 and 13 -16 corresponded to mild,

moderate and severe spasticity respectively. Eleven patients had mild, two had moderate and none had severe spasticity.

C) Box and Blocks test - BBT (Mathiowetz et al., 1985) of manual dexterity. BBT measures unilateral gross manual dexterity and has established norms for age groups. Even though our motor task was not related directly to dexterity, this test provided a measure of how much the patients used their hemiparetic hand in their daily activities. The test consists of grasping and moving wooden blocks (2.54 cm^3) from one side of a box to another within 60 seconds. The test was repeated twice for each hand and the results were averaged. The clinical testing procedure required about 15 minutes.

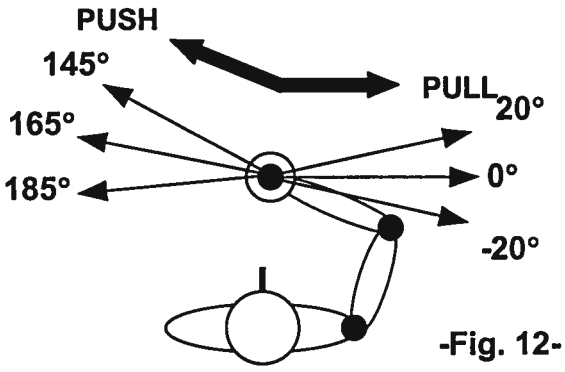
3. Measurement of maximum voluntary effort (MVE N)

The maximum force produced by the right hand in two principal directions was measured and used to determine the initial torques for the unloading experiment. The subject was seated in a chair with a back support with the right shoulder in front of a handle attached to a force transducer (Wheatstone bridge) attached to a shaft of adjustable height. Pulling and pushing forces were measured with the hand at shoulder level and the shoulder in 90° flexion and 45° horizontal adduction (0° is full horizontal abduction defined with the arm in line with the subject's right and left shoulder). The elbow was in approximately 135° extension (full elbow extension was defined as 180°) and the wrist was in the neutral

position between supination and pronation. The left arm was alongside the body. No compensation in the form of body inclination was allowed during the testing. Participants performed three consecutive trials per condition with 1 minute pauses in between. For the pushing direction, the subject pushed the handle to the left, trying to direct the force strictly in the frontal plane. For the pulling direction the subject pulled the handle to the right (the experimental set-up is shown in *Fig. 12*). The participants started to push or pull on a verbal signal and were encouraged vocally by the examiner for 3 s. The force signal was measured on an oscilloscope (Tektronix, type RM 561A). The mean of the three trials in each direction was considered as the mean maximum force of the subject. In some patients with stroke, it was necessary to attach the hand with a Velcro strap to the handle.

4. Unloading experiment

The lengths of the right arm from the acromion to the lateral epicondyle of the elbow and of the forearm from the lateral epicondyle to the distal end of the first metacarpo-phalangial joint were measured in order to calculate the torques in elbow and shoulder



-Fig. 12-

Experimental setup: position of the subjects with respect to the arm of the manipulandum's. Two principle combinations of unloading and final directions of the external force were measured (PUSH and PULL).

joints in later analysis. For the unloading experiment the subject was seated on an adjustable chair in front of a computer screen. The chair orientation as well as the distance between the sternum and the handle of the manipulandum (30 cm) were measured prior to the experimental procedure in order to reproduce the desired initial torques (30% of the MVE, measured in the previous part of the experiment). The trunk was attached by 10 cm wide Velcro straps to the back of the chair to avoid any trunk movements often observed in patients with hemiparesis during arm movements (Roby-Brami et al., 2000; Cirstea and Levin, 2000; Michaelsen et al., 2001). The right hand was placed in a polypropylene bi-valve splint attached to the handle of the manipulandum. The splint prevented wrist, forearm and hand motion leaving only the shoulder and elbow to participate in the movement. In addition, it assured a firm grip on the handle for those patients who had disturbed control of grasping. The double-joint manipulandum was controlled in the horizontal plane by two torque motors (Mavilar motors MT - 2000), each motor creating torque at one joint. Torque could be produced independently at each joint of the manipulandum (maximal torque of 60 Nm per motor, resembling a maximal force of approximately 165 N at the level of the handle for the manipulandum configuration used in this experiment). For safety of the participants the total output of the two motors was reduced to 30% of their maximum or 49.5N. A software procedure was developed to reduce the effects of the manipulandum's inertia on hand movement. Specifically, positive feedback was introduced in the torque output, based on the

acceleration (recorded through accelerometers) and the moment of inertia of each of the manipulandum's segments. The feedback factor was selected by trial and error, so as to reduce inertia without introducing oscillations. Software for the control of the experiment and the correction of inertia was developed in Labview (National Instruments, Texas, USA).

Three safety features were incorporated into the apparatus. First, the two arm segments of the manipulandum were supplied with interrupters placed near the limits of their movement (about 110° in the horizontal plane for each articulation) which automatically switched off the motors when the segment reached these limits. Second, the motors were automatically turned off if the speed of the manipulandum approached the limit of the natural speed of the arm and hand (2 m/s). Third, subjects held a "Panic" button, which interrupted the electrical supply to the torque motors instantly when pressed. In addition, the total output of the two motors was reduced to 30% of their maximum or 49.5N.

The subject moved the handle of the manipulandum until the position of the hand, indicated by a cursor on the computer screen, reached a fixed target (within a 2 cm red circle at the center of the screen). As the hand approached the target position, the resistance force applied to the manipulandum by the torque motors increased linearly with a constant direction until it reached a peak (30% of the MVE) when the cursor reached the centre of the target. The external load started at a value of 0 N at a distance of 10 cm from the target. Once the target was attained the

subject maintained the cursor within its borders. Initial resistance torques were applied to the handle and the arm by the motors according to two conditions: For the PUSH condition the force vector generated by the motors moved the subject's arm to the right and for the PULL condition it moved the arm to the left.

The subject was required to match the initial force at the handle in order to maintain the hand at the target position. After a randomized delay period of 2 to 4 s, the torque from the manipulandum was unexpectedly decreased. The subject was instructed not to intervene (react) to this perturbation (i.e. let the arm go in a natural way to a new final position in space without trying to intentionally modify the movement or to relax completely). The participants had complete vision of their arm and of the computer monitor during the experiment. However, the position of the cursor on the screen was not updated after removal of the load. An important assumption in the "do not intervene" paradigm is that the subject is capable of maintaining a constant pattern of motor signals. We did not use the combination of loading and unloading of already active muscles because of the likelihood of inequifinality in the final position occurring due to the tendency to voluntarily or automatically change the central commands under these conditions (Feldman, 1975; Feldman and Levin, 1995). Several training trials (usually 5 to15) were done before data were recorded. Practice ended when subjects reliably produced movements without corrections and containing a single peak in the hand velocity profile for at least 3 consecutive trials. The initial force direction for the

"PUSH" condition was 165° with respect to the subject's hand position so that the initial force was directed to the right of the individual approximately in the frontal plane and 0° for the "PULL", which directs the initial force in the same plane but in the opposite direction. We hypothesized that joint torques and angles would be related in the form of a smooth function and therefore we used a large number of unloading conditions in order to better characterize this surface. For both initial directions there were three different directions of unloading: 0°, +20° and -20° with respect to the initial direction. Six different levels of unloading (60%, 40%, 20%, 10%, 0% and –10% from the initial torque) were applied with zero deviation from the initial load direction. Three different levels of unloading (40%, 20% and –10% from the initial torque) were used for two other directions (+20° and -20°) of the initial force (*Table 2,* see *Fig. 12*).

-Table 2- description of all conditions of unloading.

Condition	1	3	6	8	9	11	4	7	12	2	5	10
Push	165°	165	165°	165°	165	165°	185°	185	185°	145	145°	145°
Pull	0°	0°	0°	0°	0°	0°	+20°	+20	+20°	-20°	-20°	-20°
Force	60%	40%	20%	10%	0%	-10%	40%	20%	-10%	40%	20%	-10%

For each of the 12 combinations, there were 6 trials for a total of 72 trials per condition. The 12 different combinations were randomized for each condition. The experiment thus consisted of 144 trials and lasted 2.5 - 3 hours. To avoid fatigue , subjects were allowed to rest in between trials

whenever necessary and there was a 15 min. pause between experiments for PUSH and PULL.

5. Recorded variables

- ***Electromyographic activity (EMG).*** Six pairs of bipolar surface electrodes were used to record the electromyographic (EMG) activity of brachioradialis (BR), anconeus (AN), biceps brachii (BB), lateral head of triceps brachii (TB), deltoideus posterior (DP), clavicular portion of pectoralis major (PM) after standard skin surface preparation. Electrodes were placed so that cross-talk contamination was eliminated by observing the response to isolated and associated contractions of the target muscle. EMG signals (16 channel Grass electromyograph) were amplified (gain 10-20), filtered (5 – 500 Hz) and sampled at a rate of 1500 Hz. The signals were filtered offline using high-pass filters (cutoff = 35 Hz) to remove motion artifacts.

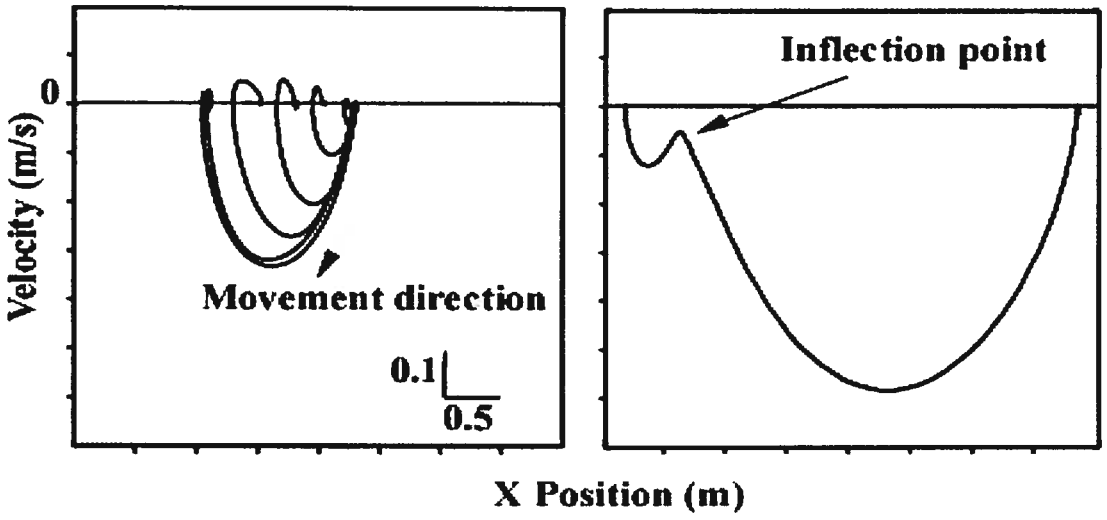
- ***Movement kinematics*** were recorded with four active infrared emitting diodes (IREDs) placed on the acromion processes of the two shoulders, lateral epicondyle of the right humerus and the handle of the manipulandum. Data were collected (sampling frequency, 100 Hz) for 3 seconds with an Optotrak Motion Analysis System (Northern Digital, model 3010, Waterloo, Ont).

6. Data Analysis

- ***Kinematic data*** - The changes in arm joint angles of the shoulder and elbow in the horizontal plane were calculated based on the scalar products of the vectors joining the appropriated IREDs. The angles of the elbow (flexion / extension) were calculated by connecting the vectors of the forearm and arm. Horizontal adduction of the shoulder was computed from the vector of arm and the horizontal line between markers on the two shoulders. The horizontal projection of the line between the two shoulder markers was considered as 0° for the shoulder, complete extension of the elbow was considered as 180°. Angular velocities of the manipulandum for each of the two segments were measured by two axial resolvers. Velocity and acceleration of the handle were computed by time derivates of XY positional data. Torques created by the manipulandum were measured by strain gauges incorporated into the manipulandum. Elbow and shoulder torques were computed from the initial force applied by the subject at the hand in consideration of the limb segment length. Data were recorded for 3 seconds, as the initial point was considered 0.5s before unloading. For each participant, pre-unloading mean values of all variables were acquired at the 0.2 and 2.5 seconds after the initiation of the record for a period of 100 ms each, which corresponds to the times before and after unloading. Movement onset and offset were defined as the times at which hand velocity exceeded or fell below 20 m/s

respectively. Trials in which subjects reacted to the unloading perturbation were repeated.

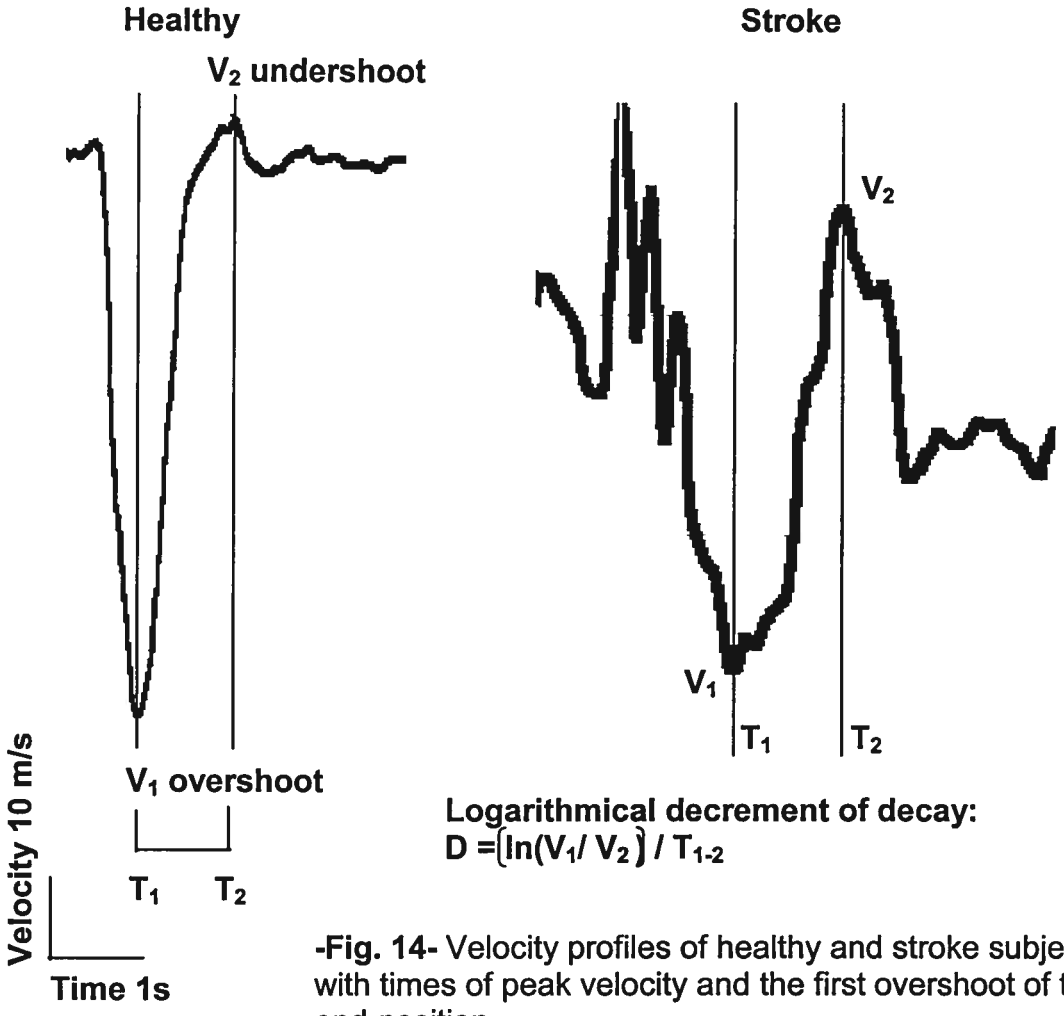
Such intervention was easily identified as points of inflection in the velocity/position phase diagrams (Fig. 13B, arrow), plotted as the velocity of movement against the end-point displacement on the x (frontal) axis.



-Fig. 13-

Phase diagrams for 6 different combinations of unloading without voluntary correction (panel A). Voluntary correction of the final position of the arm (panel B).

Tangential velocity profiles of the hand were used to analyze oscillations around the final hand position after unloading as a measure of postural stability and to determine instability indexes for each subject. For the latter, we used the inverse of the logarithmic decrement of decay of the oscillations of the arm: $D=\left[\ln(V_1/V_2)\right] / T_{1\text{-}2}$ (Levin and Dimov 1997), where V_1 and V_2 are peak velocities of the hand and the first overshoot of the target respectively and $T_{1,2}$ is the time between them (*Fig. 14*). The decrement is related to the system's damping and stiffness, as a higher value of the index indicates decreased damping or greater oscillations.



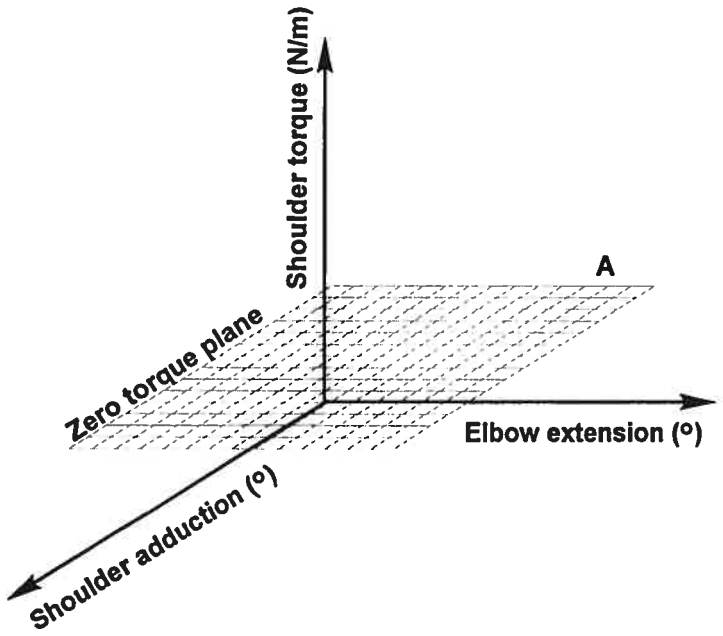
-Fig. 14- Velocity profiles of healthy and stroke subjects with times of peak velocity and the first overshoot of the end-position.

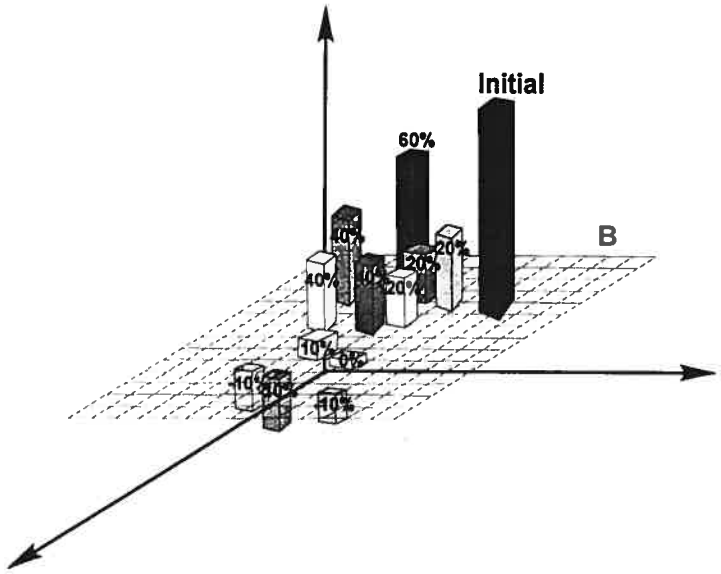
- ***Invariant characteristics (ICs) of the double-joint system.*** The invariant characteristic of the double-joint system differs from that of single joints because of the dependency of the torque of one joint on the positions of the same and the neighbouring joint. For the creation of ICs, we used the same procedure as in Archambault et al. (2003). We averaged the torques for each of the 12 conditions and plotted them versus the position of the two joints (shoulder and elbow). The regression surface (Fig. 14.1) which passed trough the averaged torques represents

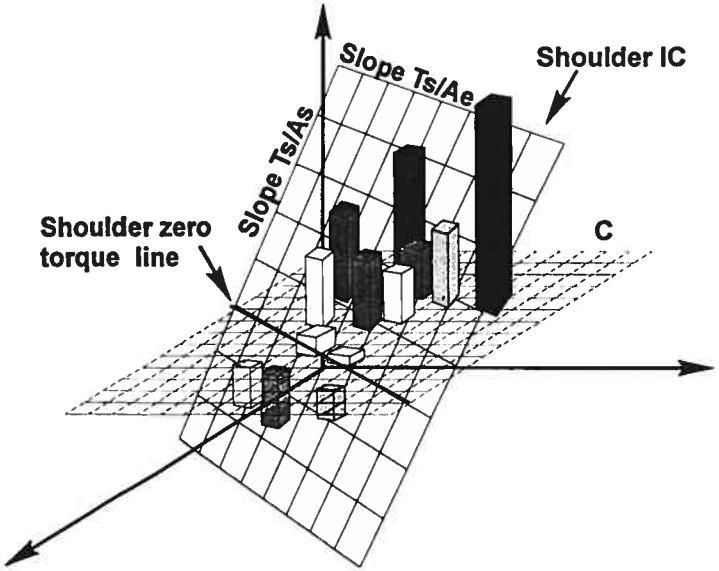
the double joint IC. The surface is limited by the two single joint ICs. For example the IC surface of the torque in shoulder is limited by the IC curve of the torque in the shoulder and the angle in the shoulder from one side and the torque in the shoulder and the angle in the elbow. We hypothesised that joint angles and torques would be related as a smooth function. That is why a large number of unloading conditions were used. The ICs of the arm were determined by plotting the regression surfaces of the interaction between displacement of the 2 joints with respect to the torque of the elbow and the shoulder. The two surfaces, of which the slope represents the stiffness of each torque/angle interaction, were used to describe the position of the double-joint system. Each regression surface (IC) had two slopes with respect to the angular plane: one for the elbow and one for shoulder. Thus each subject had 4 slopes per condition:

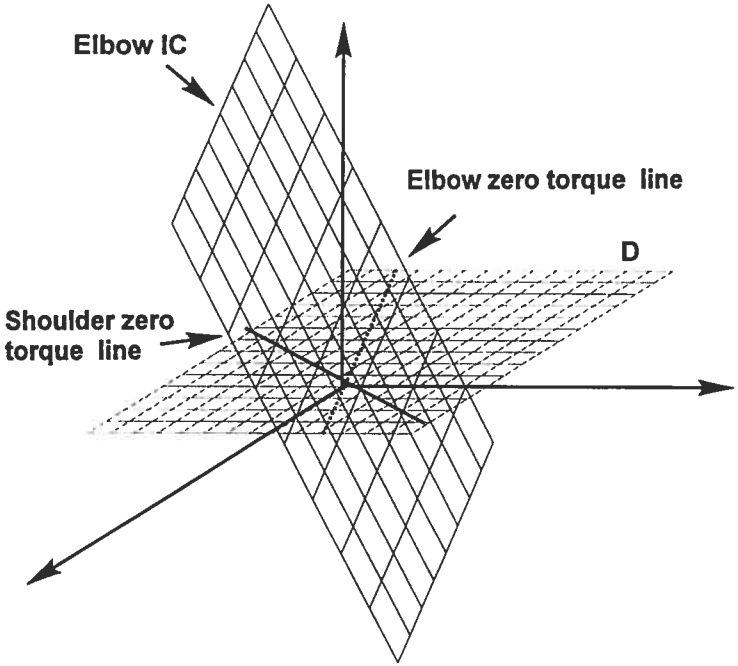
- $Torque_{elbow}$ / $Angle_{elbow}$ (T_e/A_e);
- $Torque_{elbow}$ / $Angle_{shoulder}$ (T_e/A_s);
- $Torque_{shoulder}$ / $Angle_{elbow}$ (T_s/A_e);
- $Torque_{shoulder}$ / $Angle_{shoulder}$ (T_s/A_s).

The site of intersection of the invariant surface with the angle plane (where torque is zero), represents a zero-torque line. The point of intersection of the two zero-torque lines, one for each joint, represents the referent configuration (RC) of the arm or the referent point assigned by the central command in terms of angular coordinates.

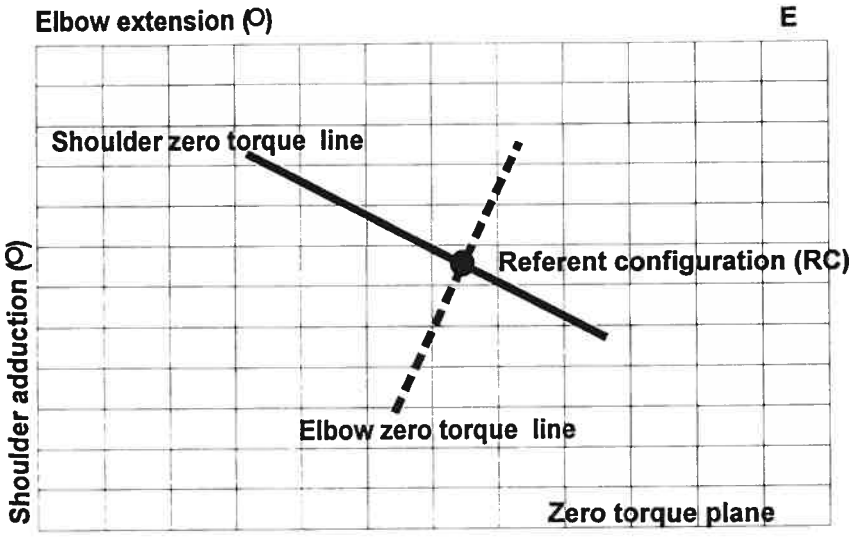

Shoulder torque (N/m)
A
Zero torque plane
Elbow extension (°)
Shoulder adduction (°)


Initial
60%
B


Slope Ts/Ae
Shoulder IC
Slope Ts/As
Shoulder zero torque line
C


Elbow IC
Elbow zero torque line
D
Shoulder zero torque line

-Fig. 14.1-



-Fig. 14.1-

-Fig. 14.1- Panel A represents the system of co-ordinates system consisting of joint torque (shoulder) versus the angular positions of the shoulder and elbow. The zero torque plane is shown as an open grid. In B, the mean initial torque versus joint positions before unloading is plotted (solid black bar) the mean shoulder torques are plotted. Combinations of joint torque and joint positions for each level of unloading are plotted and different shades of grey represent the three directions of unloading (white = 0; medium grey = +20; light grey = - 20). In C, the regression surface passing through the mean torques represents the invariant characteristic (IC) of the shoulder. This surface has two slopes ($Torque_{shoulder}$ / $Angle_{elbow}$ and $Torque_{shoulder}$ / $Angle_{shoulder}$). The intersection of the the IC with the zero torque plane represents the infinite combination of shoulder and elbow angles for the given central command at which shoulder torque is zero. In D, the IC and zero torque line for the elbow is shown. The intersection of the two zero torque lines (E) represents the position determined by the central command for the double joint system or the referent configuration (RC).

- ***EMG signals.*** To determine the level of tonic activity before and after unloading, we calculated the root-mean square value of the EMG activity during an 100 ms window centered at the 0.2 and 2.5 s marks of the recorded trials respectively. The role (agonist – AG, antagonist - ANT) of each muscle during the task was determined by examining its response to unloading and agonist / antagonist EMG patterns were identified for each joint (elbow and shoulder). Coactivation ratios for shoulder and elbow AG and ANT pairs were calculated before and after unloading, according to the formula ANT/(AG+ANT). These ratios were correlated with instability indexes described above.

7. Statistical analysis

Student t-tests were used to investigate the following data: differences between the initial torques and angles for the two conditions within each group and between groups; dispersion of the trajectories; differences in angular velocity in both groups; R^2 of the regression surfaces; slopes of the invariant characteristics. ANOVAs were used to investigate the differences in joint-torques and angles among different combinations of unloading. The instability indexes were investigated by Kruskal-Wallis ANOVA. We used Pearson correlation matrices to investigate the link between coactivation ratios, levels of instability and levels of clinical impairment.

Chapter V. Article

Control of double-joint arm posture in patients with unilateral brain damage.

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1. Abstract

According to the λ model for motor control, multi-joint movements result from the specification of an internal referent body configuration. The activity of muscles and force required for movements emerge following deviation of the actual body configuration from the referent one. We identified the referent arm configurations specified by the nervous system that provide responses of the arm to sudden unloading, both in healthy individuals and in those with arm motor paresis due to stroke. From an initial position of the right hand, subjects matched the force produced at the handle of a double-joint manipulandum by two torque motors by pushing the hand to the left (165°) or pulling it to the right (0°). For both initial conditions, 3 directions of the final force: 0°, +20° and -20° with respect to the direction of the initial force were used. Subjects were instructed not to intervene when the load was unexpectedly partially or completely removed. Both groups of subjects produced similar responses to unloading of the double-joint arm system. Partial removal of the load resulted in distinct final hand positions associated with unique shoulder-elbow configurations and joint torques. The net static torque at each joint before and after unloading was represented as a function of the two joint angles describing a planar surface in 3D torque-angle coordinates, or invariant characteristic. For each initial condition, the referent arm configuration was identified as the combination of elbow and shoulder angles at which the net torques at the two joints were zero. These

configurations were different for different initial conditions. The identification of the referent configuration was possible for all healthy participants and for most of those with hemiparesis. This indicates that most individuals with stroke-related brain damage and hemiparesis preserved the ability to adapt their central commands - the referent arm configurations - to accommodate changes in the external conditions. Despite the preservation of the basic patterns of responses, individuals with stroke damage had a more restricted range of hand trajectories following unloading, an increased instability around the final endpoint position and differences in the dispersion of referent configurations in elbow-shoulder joint space compared to healthy individuals. Moreover, in 4 out of 12 patients, referent configurations of the arm could not be identified, suggesting deficits at a higher level of motor control. The deficits in the specification of referent arm configurations may affect the ability of patients with stroke to produce coordinated multi-joint movements.

2. Introduction

Postural stability of the arm results from the specification of a coordinated pattern of agonist and antagonist muscle activity around appropriate joints. In particular, coactivation of the opposing muscle groups increases joint stiffness and thus improves stability of posture and movement (Asatryan and Feldman 1965). Levin and Dimov (1997) showed that after sudden unloading of the pre-activated elbow flexors, the forearm moves to a new position at which it is stabilised due to the appropriate feed-forward specification of agonist and antagonist muscle coactivation within a spatial zone surrounding the final position. The location of the coactivation zone (CZ) in the biomechanical range of the joint is determined by the position called the referent (R) position of the joint. When the extent of the CZ is zero, R coincides with the threshold position for all muscles of the joint. In other words, at this position EMG activity of all muscles is zero but when there is a deviation from this position, appropriate muscles are activated to resist the deviation. It has been shown experimentally that the activation threshold is identical to the threshold of the stretch reflex (SR) and changes in the threshold may underlie voluntary movements at the elbow joint (Asatryan and Feldman 1965; Levin and Dimov 1997). Since the localization of the CZ in joint space is determined by the R command, the CZ moves with the R. In the control of posture and movement, changes in R can also be associated with task-related changes in the size of the CZ (Levin and Dimov 1997).

Archambault et al. (2001) recently showed that the notions of referent position and coactivation zones are applicable to motor tasks involving the whole arm, in which case, neural control levels would specify referent positions of all the joints of the arm. These positions represent the referent configuration (RC) of the arm. As for a single-joint, when the extent of the CZ is zero, the RC is the configuration of the arm at which all arm muscles simultaneously reach their threshold of activation. When the actual configuration of the arm (Q) differs from RC, muscular activity is generated in proportion to this difference between Q and RC. This difference is thus a global factor influencing the activity of all arm muscles. This factor is combined with individual factors influencing the activity of motoneurons (anatomical arrangement, afferent feedback, etc.). As for a single joint, the RC also determines the spatial location of the CZ if such a zone is present and the RC remains the same when the extent of the CZ is changed. As a result, the net joint torques generated at this configuration remain zero. It has also been demonstrated (Archambault et al. 2001, 2003) that active movements of the arm result from task-related changes in RC, usually accompanied by a change in the extent of the CZ. In this formulation, the nervous system does not directly control EMG and forces, but these are seen as emergent properties of shifts in the RC, as well as changes in the location and extent of the CZ. Experimentally, it has been demonstrated that the RC and CZ concepts can be applied to whole body movements (Coté et al. 2002; Lestienne et al. 2000; St-Onge and Feldman, 2003).

The concepts of the RC and CZ have been introduced in the framework of the λ model for motor control (Levin and Feldman 1994). The model is applicable not only to posture and movement in healthy subjects but also in individuals with lesions in the central nervous system. In particular, it has been shown that motor impairments at the elbow, including muscle weakness and spasticity in hemiparetic patients recovering after stroke as well in children with cerebral palsy result from deficits in the range of regulation of activation thresholds of flexor and extensor muscles acting around the elbow joint (Jobin and Levin, 2000; Levin and Feldman 1994; Levin et al. 2000).

Motor control studies in individuals with stroke have described numerous deficits such as abnormal movement synergies (Brunnström 1970), reduction in the ability to independently activate muscles out of the pathological synergies (Reinkensmeyer et al. 2002), changes in the spatiotemporal organization of hand movement manifested in alterations of reaching and grasping strategies (Roby-Brami et al. 1997; 2003) as well as use of the trunk as a compensatory strategy for extending the reach of the hemiparetic arm (Cirstea and Levin 2000; Levin et al. 2002; Michaelsen et al. 2001).

Several studies have examined multijoint control of the arm in individuals with stroke-related brain damage (Beer et al. 2000; Dewald et al. 1995; Kamper et al. 2002; Levin 1996; Roby-Brami et al. 1997; Trombly 1993; Wing et al. 1990). Deficits in multi-joint movement in patients with

stroke has been alternatively attributed to disruptions in interjoint coordination (Cirstea et al. 2003; Levin 1996), the presence of pathological movement synergies (Dewald et al. 1995; Reinkensmeyer et al. 2002) and impaired feedforward control of the passive interaction torques at the elbow joint (Beer et al. 2000). Investigation of motor deficits after stroke-related brain damage in the context of a physiologically feasible model of motor control permits the clarification of which elements of the control of movement are disrupted. As a first step in this direction, Levin and Dimov (1997) and Levin et al. (2000) investigated motor deficits in a single-joint system according to the λ model by using the unloading method. In the present study we extended this approach to the double-joint system by analyzing the regulation of referent arm configurations in posture and movement production following unloading (the regulation of the CZ will be the subject of a separate report). We specifically focused on the role of centrally specified RCs in determining the patterns of the interaction between the multi-segmented arm and external environmental forces, as well as on the changes of the RC in the accommodation of the neuromuscular system to different environmental conditions. In this context, we analyzed differences in the control of arm posture and movement between healthy subjects and individuals with hemiparesis. Some of the results have appeared in abstract form (Mihaltchev et al. 2002).

3. Material and methods

3.1 Subjects

Thirteen patients with stroke (mean age: 51.9 ± 12.1 years) and 11 healthy subjects (49.0 ± 9.0 years) participated in the study after signing informed consent forms approved by the Ethics Committee of the Rehabilitation Institute of Montreal. The patients had right spastic hemiparesis due to cerebro-vascular accident (CVA) in the left hemisphere at least 6 months previously. Only patients with lesions in the dominant left hemisphere were studied in order to control for response variability due to differential control of specific components of a variety of motor tasks performed by the arm and hand (Winstein and Pohl 1995; Chen et al. 1997). Participants had full passive range of movement at the shoulder, elbow and wrist, partial control of the arm and no severe apraxia or receptive aphasia. Those with bilateral stroke, pain in the arm or trunk, dysmetria and static or dynamic tremor were excluded. Patient demographic and lesion location information is presented in Table 3. The healthy group consisted of right-handed individuals with no sensory or motor impairments or orthopedic problems affecting the arm or trunk. Those with uncorrected visual disturbances were excluded.

The experimental session consisted of two parts: determination of maximal voluntary effort (MVE) and an unloading experiment. In addition, patients underwent a clinical evaluation to determine the sensorimotor status of their affected arm.

-Table 3- Demographic data, maximal voluntary effort (MVE) and initial angular and torque values for the PUSH and PULL conditions in patients with stroke and in healthy subjects. Results of clinical testing of patients with stroke are also given.

Subjects	Age/Sex	Lesion	Time[1]	CSI[2](/16)	FM[3](/66)	BBT[4](R/L)	MVE PUSH (N)	MVE PULL (N)
1	58/F	frontal	41	11	10	0/76	33.3	16.7
2	52/M	temp.fronto parietal	48	7	14	0/60	29.6	19.7
3	61/M	ant.temporalpar.	36	11	20	0/65	37.73	27.87
4	45/M	fronto parietal	28	9	22	7/56	123.3	98.64
5	55/M	ant.temporalpar.	47	4	30	0/50	24.66	22.19
6	42/F	fronto parietal	36	4	38	13/65	51.05	34.52
7	49/F	ant. temporal	18	9	39	30/56	23.06	18.13
8	41/M	basilar artery	28	5	52	21/75	73.98	64.12
9	51/F	parietal subcortical	72	7	56	51/65	59.2	44.4
10	25/F	fronto parietal	46	7	58	28/79	35.8	24.7
11	69/M	parietal	36	6	60	61/75	96.2	107.3
12	69/M	parietal	47	3	61	59/61	120.8	86.3
13	58/M	temporal	60	4	65	55/61	64.1	64.1
Mean ± SD	51.9 ± 12.1			6.8 ± 2.6	40.4 ± 19.6	R 23.1± 25.0 L 60.2± 20.1	59.4 ± 34.9	48.4 ±32.3
Healthy	49.0 ± 9.0						82.6 ±33.2	79.2 ±51.7

1. Time since the AVC in months; 2. Composite Spasticity Index; 3. Fugl-Meyer score; 4. Box and Blocks Test

-Table 3 – continued

	PUSH Initial Torques and Angles				PULL Initial Torques and Angles			
Subjects	$T_{elbow\ (N/m)}$	$T_{shoulder\ (N/m)}$	$\theta_{elbow\ (°)}$	$\theta_{shoulder\ (°)}$	$T_{elbow\ (N/m)}$	$T_{shoulder\ (N/m)}$	$\theta_{elbow\ (°)}$	$\theta_{shoulder\ (°)}$
1	-0.19	2.10	92.0	73.8	-0.85	-1.65	107.2	92.5
2	-0.74	3.23	78.1	66.2	-0.22	-1.67	73.8	82.5
3	-0.59	2.96	90.2	73.6	-0.95	-1.05	79.4	61.6
4	0.55	11.19	100.5	82.4	-1.90	-8.79	101.2	89.9
5	-0.40	2.30	78.0	60.1	-0.96	-1.99	74.7	62.5
6	0.00	3.94	97.1	75.2	-1.42	-2.83	86.9	76.5
7	-0.06	1.57	74.4	52.5	-0.85	-1.09	70.7	48.3
8	-2.64	5.14	90.1	66.9	-2.64	-5.14	87.2	68.9
9	0.47	5.37	93.7	73.6	-1.99	-3.00	78.2	63.7
10	-0.94	3.30	74.8	66.5	-1.04	-1.49	73.0	58.5
11	-0.20	8.31	89.9	73.6	-3.67	-7.37	77.6	56.5
12	1.46	11.91	87.8	67.4	-3.64	-6.35	86.1	62.0
13	-0.89	6.43	82.6	73.5	-1.97	-4.96	77.6	61.8
	-0.3 ± 1.0	5.2 ± 3.4	86.8 ± 8.5	69.6 ± 7.5	-1.7 ± 1.1	-3.6 ± 2.6	82.5 ± 11.0	68.0 ± 13.3
	0.03 ± 1.0	7.9 ± 3.6	84.0 ± 4.0	63.4 ± 10.5	-3.0 ± 1.4	-4.1 ± 2.0	76.4 ± 5.9	51.8 ± 11.0

The clinical evaluation was administered by a qualified physiotherapist using a test battery consisting of three measures:

A) Arm motor impairment was measured with the valid and reliable Fugl-Meyer scale (Fugl-Meyer et al. 1975). Since we focussed on arm motor impairments, we used the arm and hand section of the scale that has a maximum score of 66 points corresponding to normal function. The test consists of 7 sections evaluating reflex activity, flexion and extension movement synergies, the ability to perform isolated movements, the functional activity of the wrist and hand, coordination and speed of arm movements. According to this scale, 6 (8 to 13) patients had mild motor impairment with scores ranging from 10 to 39 and 7 (1 to 7) patients had moderate to severe impairment with scores from 52 to 65.

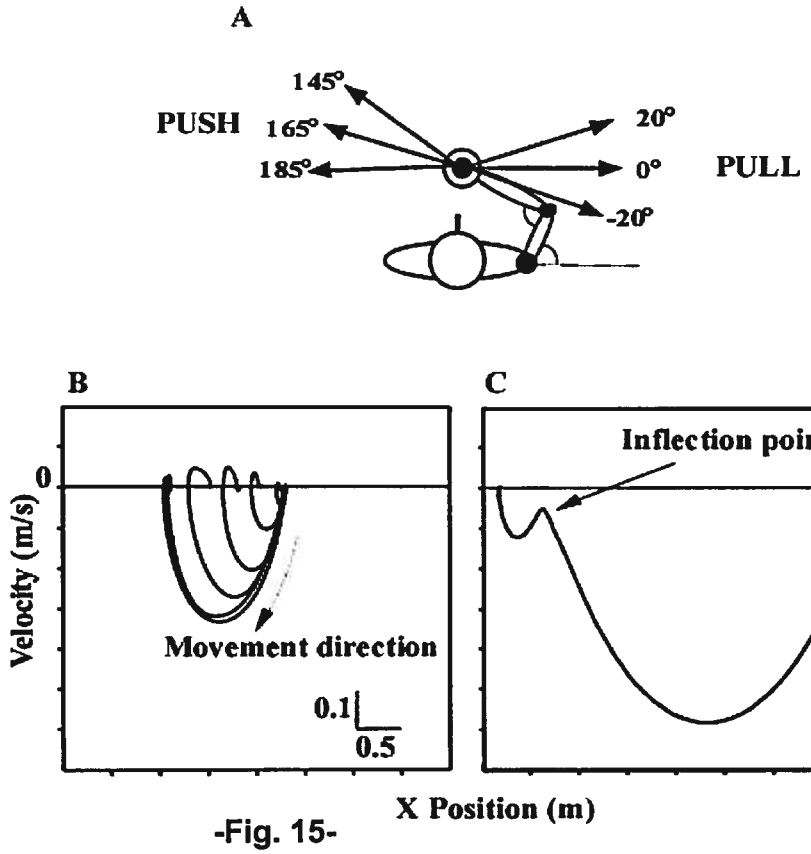
B) Spasticity of the elbow flexor muscles was assessed with the valid and reliable Composite Spasticity Index (CSI; Goulet et al. 1996; Levin and Hui-Chan 1992; Nadeau et al. 1999). The CSI rates the excitability of biceps-brachii tendon jerks, the resistance to passive elbow extension applied at a moderate speed, and the amount of wrist clonus. Each sub-test is rated on 4-point scales while the one for resistance is doubly weighted since this measure most closely resembles tone. The three scores were summed. Based on clinical experience and results of previous studies, total scores ranging from 5 to 9, 10 to 12 and 13 to 16 corresponded to mild, moderate and severe spasticity respectively. According to this scale, 9 participants had mild and 4 had moderate spasticity in the arm.

C) Manual dexterity was assessed with the Box and Blocks test (BBT, Mathiowitz et al. 1985) that measures unilateral gross manual dexterity and has established norms for age groups. Even though our motor task was not directly related to dexterity, this test provided a measure of how much the patients used their hemiparetic hand in their daily activities. The test consists of grasping and moving wooden blocks (2.5 cm^3) from one side of a box to another in a period of 60 seconds. The test was repeated twice for each hand and the results were averaged. The clinical testing procedures required about 15 minutes.

3.2 Determination of maximal voluntary effort (MVE)

The maximal force produced by the right arm in two principal directions was measured and used to determine the initial torques for the unloading experiment. The subject was seated in a chair with a back support with his right shoulder in front of a handle connected to a force transducer fixed to a shaft of adjustable height (Fig. 15A). Pulling and pushing arm forces in the frontal plane were measured at the hand. The hand was placed at shoulder level and the shoulder was in 90° flexion and 45° horizontal adduction (0° is full horizontal abduction defined with the arm in line with the subject's right and left shoulder). The elbow angle was about 45° (full elbow extension was defined as 180°) and the wrist was in the neutral position between supination and pronation. The left arm was alongside the body. No compensation in the form of trunk forward or lateral inclination was permitted during the testing. Participants performed

three consecutive trials per direction with 1 minute pauses in between. For the pushing and pulling directions, the subject pushed the handle to the left or the right respectively, trying to direct the force strictly in the frontal plane. The participants started to push or pull on a verbal signal and were encouraged vocally by the examiner for 3 s. The force signal was



-Fig. 15-

Schematic diagram of the experimental set up and examples of non-corrected and corrected responses to unloading. A: The subject sat in front of a computer screen with their trunk strapped to the back of the chair and their right forearm and hand supported by a splint attached to the handle of a double-joint manipulandum controlled by two torque motors. Subjects were required to match the force of the motors by pushing the handle to the left (165°) or pulling it to the right (0°). For both initial conditions, 3 directions of the final force: 0°, +20° and -20° with respect to the direction of the initial force were used. B, C: Velocity / position (phase) diagrams showing uncorrected (B) and corrected (C, arrow) responses to unloading in one healthy subject.

monitored on an oscilloscope (Tektronix, type RM 561A) and the means of the three trials in each direction were used in later calculations. In some patients with hand paresis, it was necessary to attach the hand to the handle of the force transducer with a Velcro strap.

3.3 Unloading experiment

The lengths of the right arm from the acromion to the lateral epicondyle of the elbow and of the forearm from the lateral epicondyle to the distal end of the first metacarpo-phalangial joint were measured in order to calculate the torques in elbow and shoulder joints in later analysis. For the unloading experiment, the subject was seated on an adjustable chair in front of a computer screen. The chair was placed so that in the initial position, the hand was at a distance of 30 cm from the subject's sternum. The trunk was attached to the back of the chair by two 10 cm wide Velcro straps to avoid trunk movements often observed in patients with hemiparesis during arm movements (Cirstea and Levin 2000; Michaelsen et al. 2001; Roby-Brami et al. 1997). The right forarm was placed in a polypropylene bi-valve splint attached to the handle of a manipulandum. The splint prevented wrist, forearm and hand motion leaving only the shoulder and elbow free to participate in the movement. In addition, it assured a firm grip on the handle for those patients who had disturbed control of grasping. The double-joint manipulandum was controlled in the horizontal plane by two torque motors (Mavilar motors MT - 2000), each motor creating torque at one joint of the manipulandum independently of the torque produced at the other joint. A software

procedure was developed to reduce the effects of the manipulandum's inertia on hand movement. Specifically, positive feedback was introduced in the torque output, based on the acceleration (directly recorded with accelerometers) and the moment of inertia of each segment of the manipulandum. The feedback factor was selected by trial and error, so as to reduce inertia without introducing oscillations. Software for the control of the experiment and the correction of inertia was developed in Labview (National Instruments, Texas, USA).

Three safety features were incorporated into the apparatus. First, the two arm segments of the manipulandum were supplied with interrupters placed near the limits of their movement (about 110° in the horizontal plane for each articulation), which automatically switched off the motors when the segment reached these limits. Second, the motors were automatically turned off if the speed of the manipulandum approached the limit of the natural speed of the arm and hand (2 m/s). Third, subjects held a "Panic" button that instantly interrupted the electrical supply to the torque motors when pressed. Finally, the maximal torque in each motor was limited to about 30% (15Nm) of its maximal output.

The subject moved the handle of the manipulandum until the position of the hand, indicated by a cursor on the computer screen, reached a fixed target (within a 2 cm red circle at the center of the screen). When the distance between the hand and the target reached 10 cm, the torque motors began to create a constant-direction force applied to the

hand that increased linearly with decreasing distance to the target until it reached a plateau (30% of the MVE) when the cursor arrived in the centre of the target. Once the target was attained, the subject maintained the cursor within its borders. Two conditions were used. For the PUSH condition, the load force was applied so that if would have moved the subject's arm to the right if it were not opposed (0° with respect to the frontal plane) and for the PULL condition, it was applied so that it would move the arm to the left (165° to the plane; Fig. 15A). The subject was thus required to match the initial force in order to maintain the hand at the target position.

After a randomized delay (2-4 s), the load force was unexpectedly decreased, resulting in motion of the arm to a new position. The subject was instructed not to intervene, i.e. let the arm go in a natural way to a new position without trying to intentionally modify the position. The participants had full vision of their arm and of the computer monitor during target attainment. However, the position of the cursor on the screen was not updated after changes in the load. It has been shown that, with this instruction, the changes in the EMG activity, muscle forces and arm position are reflex-like reactions to changes in the load, typically not involving modifications of the central control signals in the sense defined in the λ model (see above; Feldman and Levin 1995). Several training trials (5 – 15) were done before data were recorded. Practice ended when subjects reliably produced uncorrected movements characterized by a

smooth transition to a new position with a single peak in the hand velocity profile for at least 3 consecutive trials. Trials in which subjects corrected the effects of unloading could be identified by visual inspection of inflections in phase (velocity-position) diagrams displayed on the monitor after each trial (Fig. 15 B, C). Overall, 15% of trials in each group were excluded because of voluntary or involuntary corrections identified in this manner. As comparison in the single-joint experiments (Levin and Dimov, 1997) the number of excluded trials is about 1%. The corrections probably increased because of the double-joint movement. Excluded trials were repeated so that, despite the exclusion, the total number of analyzed trials remained the same in all subjects. It has also been shown that stretching of active muscles produced by sudden loading elicits protective voluntary or triggered reactions associated with changes in central commands (Feldman and Levin 1995). In the present study, it was essential to reduce corrections of responses to changes in load to a minimum and therefore we did not employ sudden loadings.

We anticipated that, for each initial condition, the relationship between torques and joint angles in steady states resulting from unloading would be described by a smooth surface in the torque-angle coordinates. To better characterize this surface, we used different levels of unloading with or without changes in the direction of the load force. For both initial conditions, we used three different directions of the final force: 0°, +20° and -20° with respect to the direction of the initial force. Six different levels

of unloading (60%, 40%, 20%, 10%, 0% and –10% from the initial force) were applied with zero deviation from the initial load direction. Three different levels of unloading (40%, 20% and –10% from the initial torque) were used for the two other directions (+20° and -20°) of the initial force (see Fig. 15, 17). For each of the 12 combinations, there were 6 trials for a total of 72 trials per condition. The 12 different combinations were randomized for each condition. The experiment thus consisted of 144 trials and lasted 2.5 to 3 hours. Thus, only one referent configuration per condition was constructed due to the lengthy recording procedure. To better characterize this behavior however, it would be desirable to measure RCs from several different initial conditions (in this case different initial forces and positions). However, the large number of repetitions and the long length of the experiment made such an experiment in patients with hemiparesis impractical.

3.4 Data recording and analysis

The position of four active infrared emitting diodes (IREDs) placed on the acromion processes of the two shoulders, the lateral epicondyle of the right humerus and the handle of the manipulandum were recorded using an Optotrak Motion Analysis System (Northern Digital, model 3010, Waterloo, Ont., sampling rate 100 Hz, 3 s/trial). The angular position and velocity of the segments of the manipulandum were measured with two axial resolvers. The shoulder and elbow angles in the horizontal plane were calculated based on the scalar products of the vectors joining the

appropriated IREDs. Velocity and acceleration of the handle were computed using X-Y positional data and a 2nd order Butterworth filter. Torques generated by the motors were measured by strain gauges incorporated into the axes of rotations. Using these torque values and basic geometry of the arm-manipulandum system, we computed shoulder and elbow torques and the force applied to the handle. For each trial, kinematic and kinetic data were measured as mean values occurring in 100 ms windows centred around two time epochs. For pre-unloading data, the window was centred at 0.3 s before unloading and for post-unloading data, the window was centred on the 2 s mark after unloading. Movement onsets and offsets were defined as the times at which the hand velocity exceeded and remained above and then fell below and remained below 20 m/s, respectively.

3.5 Torque-angle characteristics of the double-joint system.

For each initial condition (PULL or PUSH), the net static torque generated at each joint was considered as a function (called invariant characteristic or IC) of the two (elbow and shoulder) joint angles. Thus for the two joints, we determined two ICs for each condition. We defined torques that flexed the elbow and adducted the shoulder as being positive. The ICs of the double-joint system thus differ from those of the single-joint system since in the latter, the muscle torque is a function of only one joint angle. For the reconstruction of double-joint ICs, we used the same procedure as that described in Archambault et al. (2001). For each initial condition, all combinations of net static torque and joint angles measured

before and after unloading were plotted as points in a Cartesian 3D space. Using regression analysis, we determined each IC as a 3D surface best fitting the set of points associated with torques of the respective joint. These surfaces were characterized by slopes or stiffness coefficients (2 per each IC). For example, each elbow IC was characterized by two stiffness coefficients, S_{ee} and S_{es} (Nm/rad), where the first subscript (e) refers to the elbow joint from which the net joint torque was measured, and the second subscript refers to the angle (e, elbow or s, shoulder) that changed to influence this torque. As in the convention used to define the direction of torque the sign of the stiffness coefficient for the shoulder (for example) was negative when the shoulder torque was in the counter-clockwise direction. The intersection of each IC with the elbow-shoulder plane (where torque is zero), represents a zero-torque line. For each initial condition there were two ICs and thus two zero-torque lines. The point of intersection of these lines represents the configuration of the arm at which all joint torques are zero. By definition, this point is a referent configuration (RC) of the arm (see Introduction). These RCs were compared across initial conditions and groups of subjects.

3.6 Statistical analysis

Student t-tests were used to compare variables between groups or conditions (initial arm positions and torques, peak velocities, spatial dispersions of the trajectories). Final positions and torques of the hand before and after unloading between- and within- groups were tested using repeated measures ANOVAs to verify if the combinations were different

from each other. Three-dimensional torque vs angle surfaces were constructed using regression analysis and their correlations and slopes were compared between conditions and groups with Student t-tests. To determine the dispersion of referent configurations in Cartesian space, we applied geometrical analysis and compared the slopes and areas of the computed ellipses with Student t-tests. Pearson Product Moment coefficients were used to correlate recorded variables with clinical status scores. The significance level was set at $p < 0.05$ for all tests.

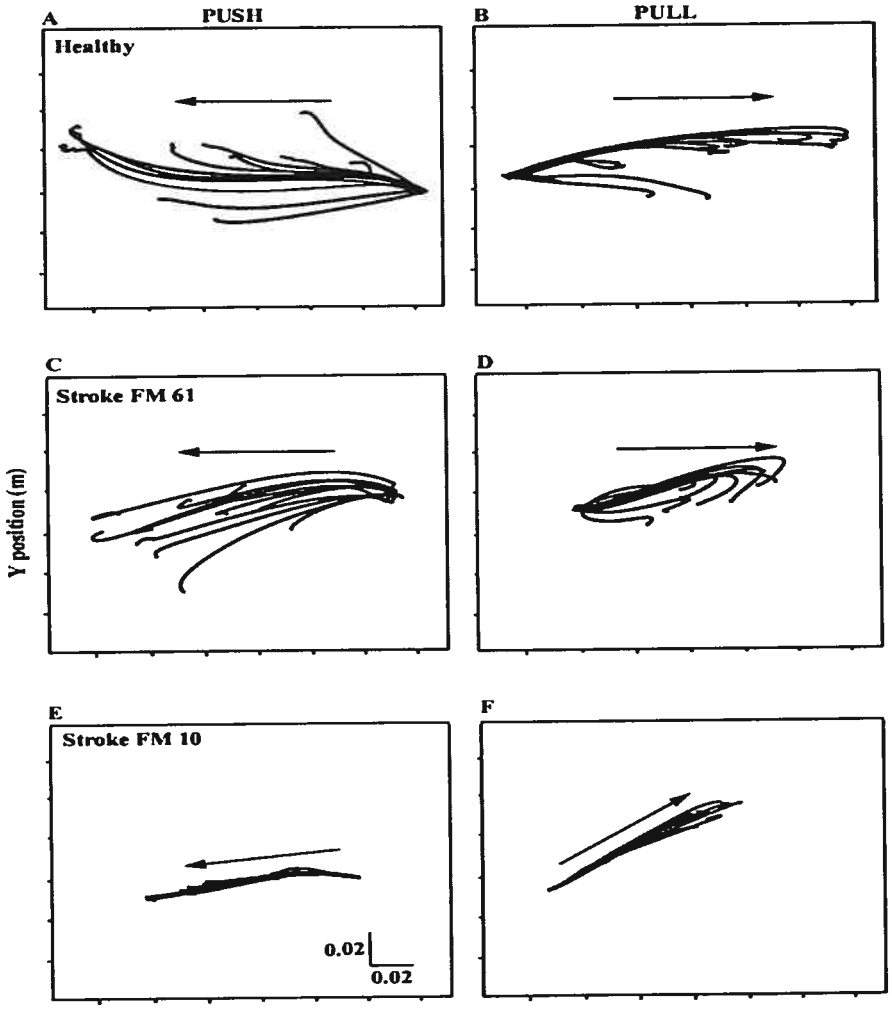
4. Results

4.1 General characteristics of responses to unloading

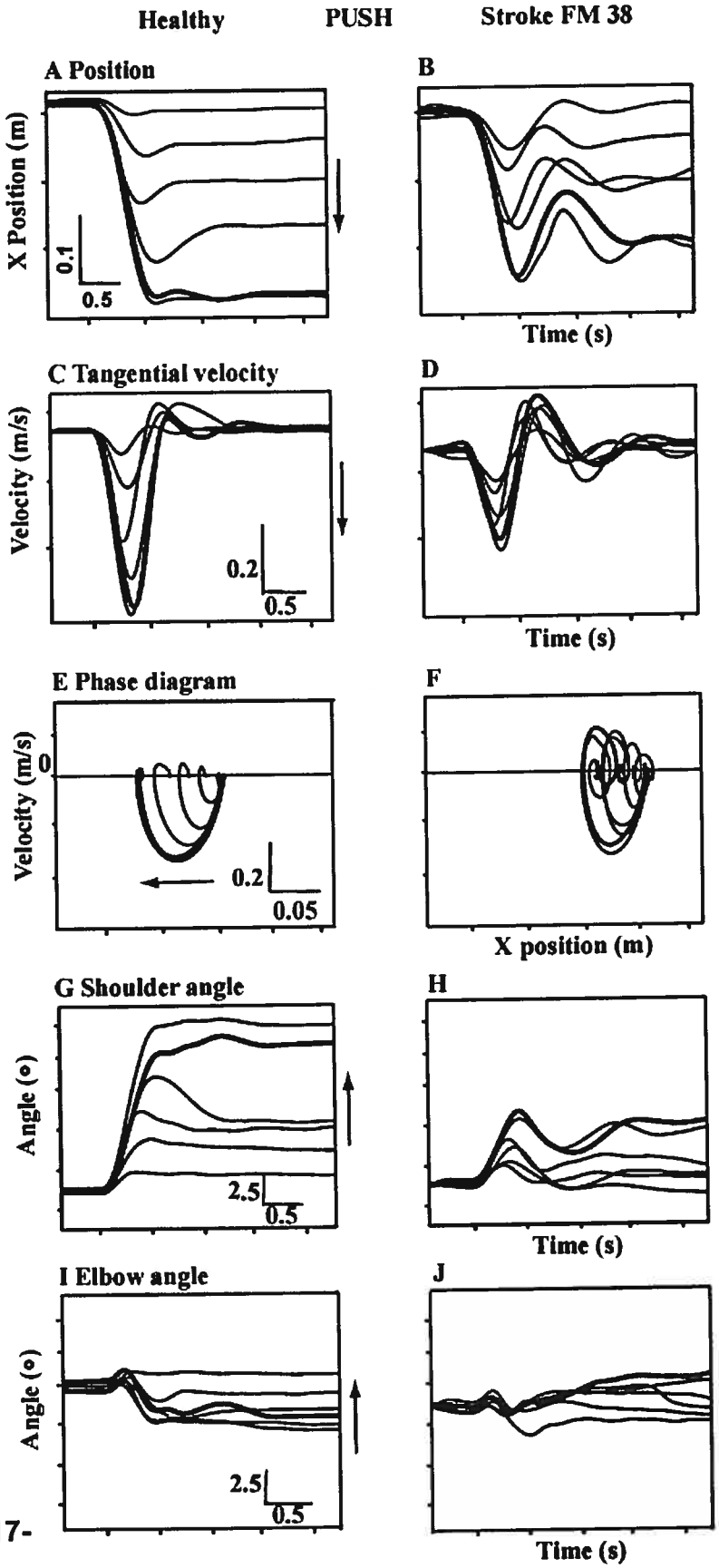
In healthy subjects, after complete or partial removal of the load, the hand began to move and, after a transient overshoot, reached a new position at which the residual load was balanced. The arm remained in the new position until the end of the trial (Fig. 16 A, B; Fig. 17, left panels). Similar responses to unloading were observed in patients but, in the latter, hand trajectories at the end of movements were sometimes hook-shaped (Fig. 16 C, D) and terminal overshoots and oscillations were more marked (Fig. 17, right panels). For each condition, after unloading, the hand stabilized in a new final position typically after less than 1 s in both groups. For complete unloading, the mean maximal peak velocity was 0.425 ± 0.116 m/s for the PUSH and 0.424 m/s ± 0.117 m/s for the PULL condition in healthy subjects compared to 0.279 ± 0.133 m/s and 0.261 ± 0.127 m/s

respectively for patients with stroke. This difference was only significant between groups for the PUSH condition (t_{22} = -2.23, $p < 0.04$).

In healthy subjects, the experimental protocol (see Methods) effectively eliminated voluntary corrections usually present in practice trials (Fig. 15 C). In patients, the terminal hooks in hand trajectories could be interpreted as corrections but these features of hand motion were likely related to the disruption in the interjoint coordination and problems in arm postural stability (Levin 1996; Levin and Dimov 1997).



-Fig. 16-
Examples of mean trajectories for each of the 12 unloading conditions for the pushing (left panels) and pulling (right panels) directions in a healthy subject (A, B) and in participants with mild (C, D) and moderate (E, F) arm motor impairment.



-Fig. 17-

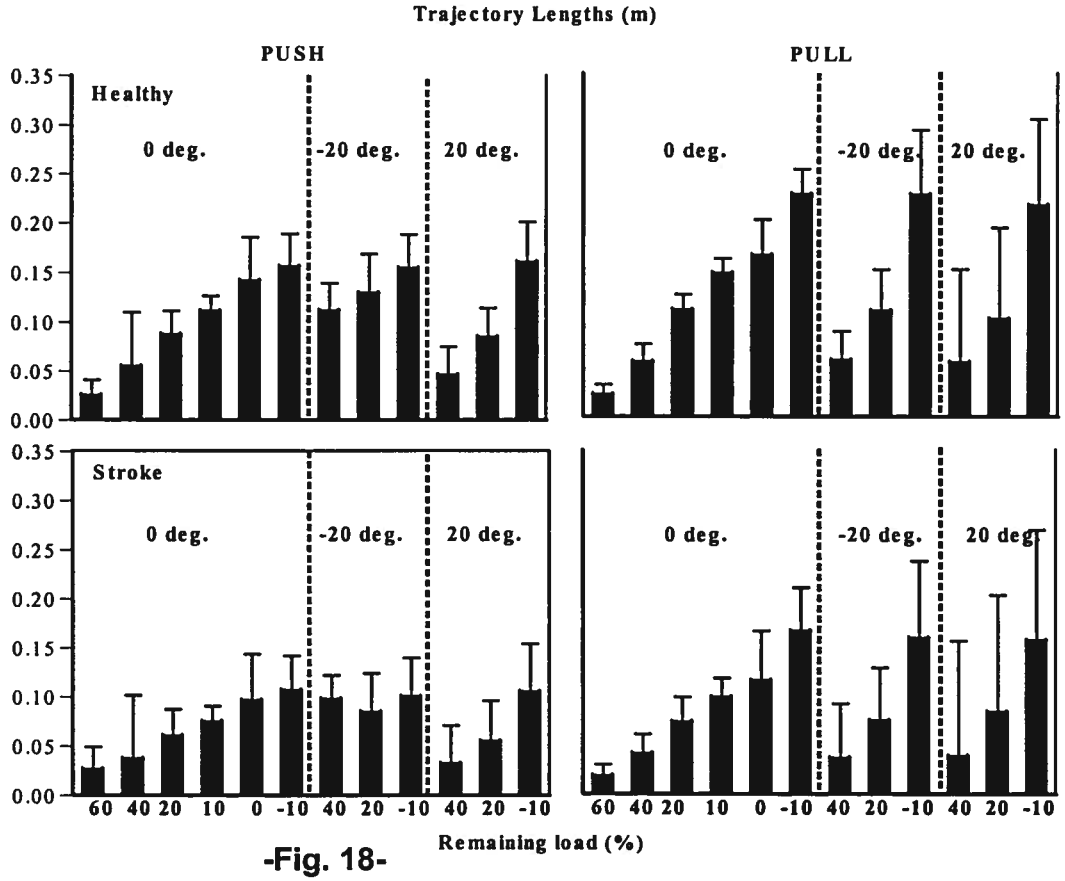
Representative kinematic data (means by condition) showing responses to 6 different combinations of unloading in the push condition (0°) in one healthy subject (left panels) and one participant with moderate arm motor impairment. A, B: hand displacements; C, D: tangential velocities; E, F: velocity / position phase diagrams; G, H: shoulder horizontal adduction angles; I, J: elbow extension angles.

For both groups of subjects, the hand displacement was in the direction opposite to that of the initial load (Fig. 16). The hand movement extents (trajectory lengths) and joint angles changed monotonically with increasing amounts of unloading (Figs. 16 - 18). In both groups, changes in position were observed in response to even the smallest changes in the load corresponding to a decrease in load of 40% (4.7 N in the healthy group and 2.0 N in the patient group), showing that the neuromuscular system was sensitive to even small perturbations. After complete unloading, the mean trajectory length (mean of all 12 combinations) was 143 ± 33 mm (PUSH) and 166 ± 65 mm (PULL) in healthy subjects compared to 97 ± 38 mm (PUSH) and 117 ± 78 mm (PULL) in patients. The change in hand position corresponded to a mean increase of 2° for PUSH and 10° for PULL for healthy subjects compared to 2° for PUSH and 2.5° for PULL in the patient group. Shoulder adduction increased on average by 13° for PUSH and decreased by 10° for PULL in healthy subjects compared to a mean increase or decrease of 9° for PUSH and PULL respectively in the patient group.

In the patient group, only 3 subjects with mild motor deficits (FM > 50) had patterns of trajectories similar to those in healthy subjects (e.g., Fig. 16 C, D). The hand trajectories in the remaining patients were restricted, not in terms of their direction, but in terms of their range of dispersion in the sagittal direction (e.g., Fig. 16 E, F). For each of these patients, the spatial dispersion was smaller for both PUSH and PULL conditions compared to the healthy subjects. For PUSH, the mean spatial

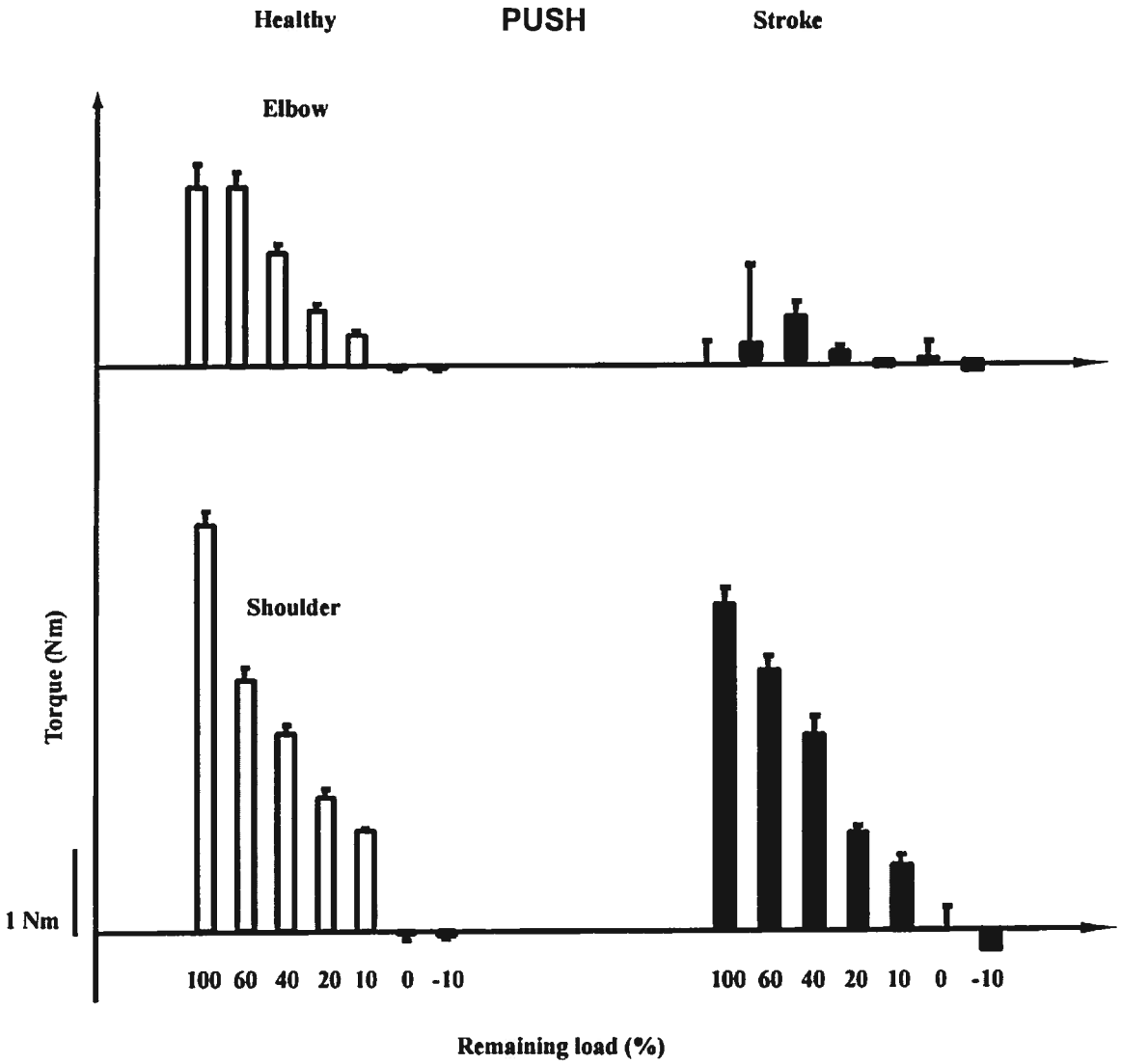
dispersion was 59.5 ± 32.9 mm for all patients with stroke compared to 96.8 ± 42.7 mm for the healthy group ($t = -2.37$, $p < 0.03$). For PULL, the mean dispersion in the stroke group was 43.7 ± 25.4 mm compared to 128.5 ± 56.8 mm for the healthy group ($t = -4.82$, $p < 0.001$).

Shoulder torques also decreased monotonically with increasing amount of unloading for both initial conditions and groups of subjects (Fig. 19). While elbow torques also changed monotonically with the change in the load for the PULL condition, this was not always the case for the PUSH condition in patients with hemiparesis (see Fig. 19).



-Fig. 18-

Mean (+ SD) trajectory lengths for PUSH (left panels) and PULL (right panels) conditions for the 12 combinations of unloading in healthy subjects (top panels) and in participants with hemiparesis (bottom panels). The first six bars in each panel show data for 6 different levels of unloading in the 0° direction. The next 6 bars show data for different levels of unloading in the + 20° (3 bars) and – 20° (3 bars) directions



-Fig. 19-

Mean (+ SD) shoulder elbow (upper panels) and shoulder (lower panels) torques in healthy subjects (left panels) and participants with hemiparesis (right panels). Data for all combinations of unloading for the 0° direction are shown

Unlike healthy subjects whose forearm stabilized in a new final position after one transient overshoot (Fig. 15 B, Fig. 17 A, C, E), the movement in participants with hemiparesis usually terminated after several cycles of oscillations (Fig. 17 B, D, F). In most cases, these oscillations

ceased after 2-3 cycles at which time mean final positions could be measured. In the cases when the hand continued to oscillate until the end of the trial, the mean final positions were measured as the midpoint of the last oscillatory cycle.

In each subject, 12 hand trajectories resulting from unloading were recorded for each initial condition (PUSH, PULL; Fig. 16). For each initial condition, the trajectory length (Fig. 18, upper panels) and peak velocity in healthy subjects increased with the amount of unloading for each of the three directions of the final load. In the patient group, the trajectory length was also monotonically related to the load for each of the final load directions, except for the direction of -20° for unloading from the PUSH condition (Fig. 18, lower panels). For both groups of subjects, repeated measures ANOVAs on the final positions after unloading for each of the 12 combinations for PUSH and PULL conditions showed that all positions were significantly different from each other suggesting that there was a unique relationship between final position and load.

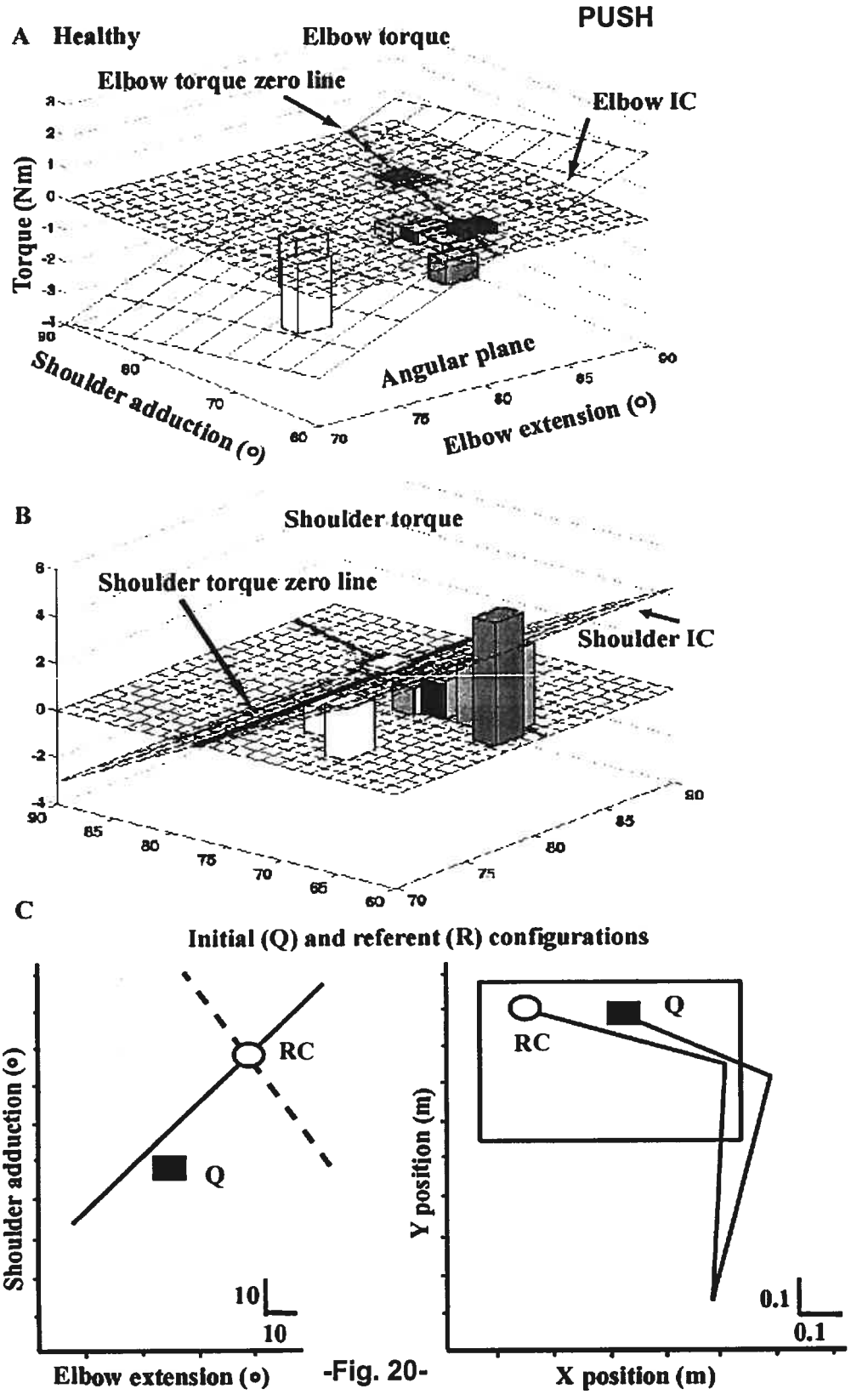
In response to unloading, the shoulder adducted in all subjects for PUSH and abducted for PULL. However, the shoulder movement was combined with elbow flexion and extension differently in each individual with no differences between the groups. The majority of subjects in both groups, combined shoulder horizontal adduction with elbow extension for unloadings in the PUSH condition (67% of the healthy group and 70% of the stroke group) and shoulder horizontal abduction with elbow extension

for unloadings in the PULL condition (83% of the healthy group and 80% of the stroke group).

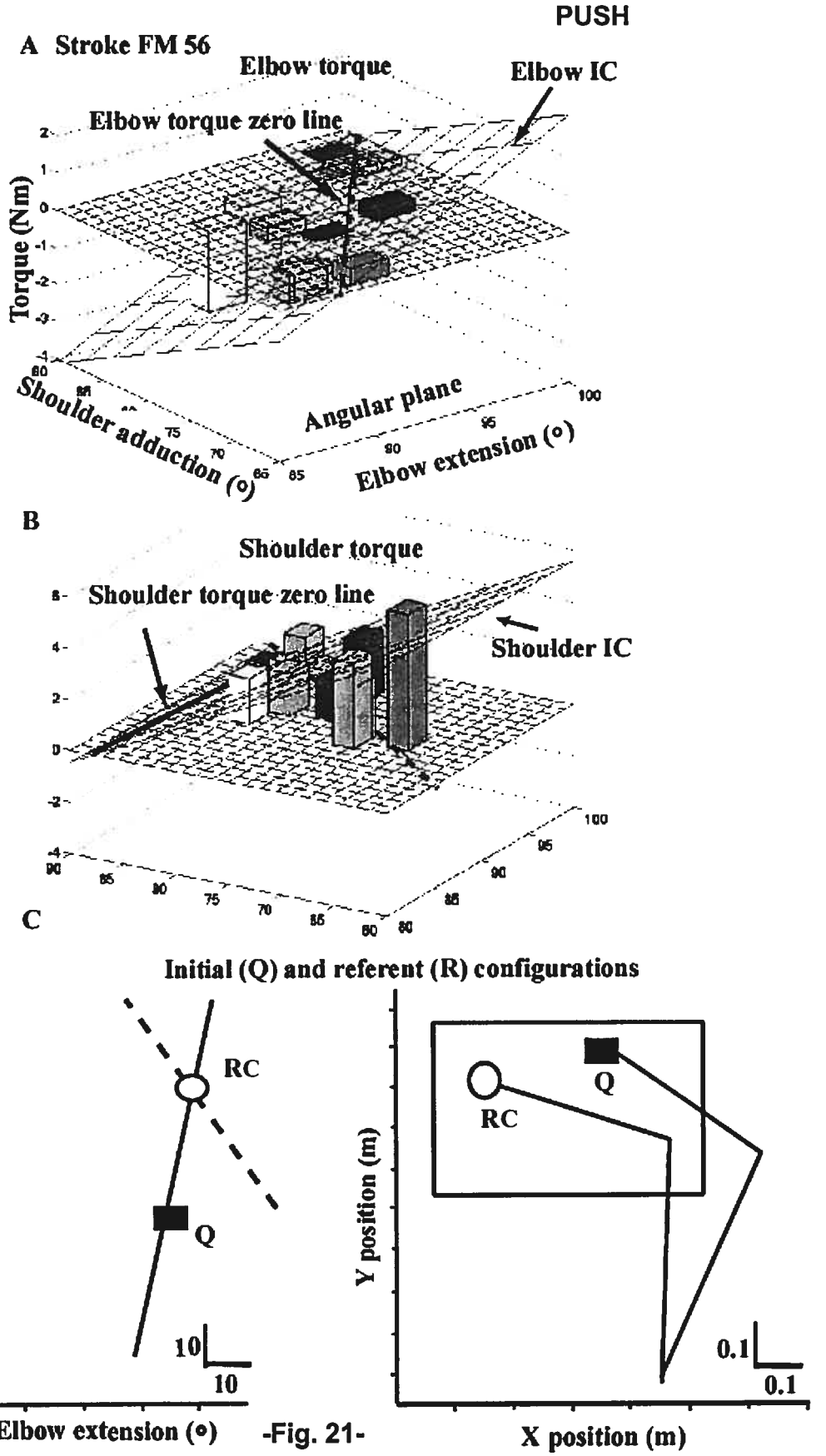
4.2 Torque-angle characteristics and referent configurations of the double-joint arm

The net static muscle torque at each joint was considered as a function of two joint angles (elbow, shoulder) that could be represented as a surface in the three-dimensional torque vs joint angle space. Thus we constructed 2 such surfaces (one for the elbow and one for the shoulder torques for each condition for a total of 4 surfaces). The points forming each surface were obtained by averaging the final joint torques and angles for all trials per condition. Together with the point representing the initial condition, 13 points were used to construct each of the 4 torque-angle 3D-surfaces. Using regression analysis, we approximated each surface by a plane (Fig. 20 and 21). The regression analyses yielded significant R^2 values (0.83 ± 0.12, range 0.48 – 0.98) in all healthy subjects (Table 4). In the participants with hemiparesis, the values were lower (0.70 ± 0.21, range 0.04 – 0.97). Two patients (Patients 7 and 8) had non-significant R^2 values for the elbow torque / angle surface for PUSH (0.33, 0.09) and 2 others (Patients 2 and 3) for the elbow torque / angle surface for PULL (0.04, 0.33). One of these patients also had a non-significant R^2 (0.28) value for the shoulder torque / angle surface for PULL. There was no correlation between clinical severity and R^2 values. Thus, the lack of ability to specify an invariant torque / angle relationship in these four patients was not related to their level of clinical impairment (Table 3). Since these

patients could not reliably specify an IC, their data were excluded from the analyses of the referent configurations.



-Fig. 20-


PUSH
A Stroke FM 56
Elbow torque
Elbow IC
Elbow torque zero line
Torque (Nm)
Shoulder adduction (°)
Angular plane
Elbow extension (°)
B
Shoulder torque
Shoulder torque zero line
Shoulder IC
C
Initial (Q) and referent (R) configurations
RC
Q
10
10
Shoulder adduction (°)
Elbow extension (°)
Y position (m)
X position (m)
0.1
0.1

-Fig. 21-

-Fig. 20 - Torque/angle characteristics and referent configurations of the arm in one healthy subject for the PUSH condition. A: Elbow torque as a function of two joint angles (elbow and shoulder) approximated by a planar surface (wide grids). The intersection of the planar surface with the zero torque plane (narrow grids) forms a line that describes all combinations of elbow and shoulder angles (arm configurations) at which the elbow torque is zero, for this condition. B: Same as in A but for shoulder torque. C: The intersection of two zero torques lines for the elbow (dashed) and shoulder (solid) identifying the unique, referent configuration (open circle, RC) at which all joint torques are zero. The solid square shows the arm configuration at the initial position the hand (Q). D: As in C, but presented in the coordinates of external space. Stick diagrams show actual positions of arm segments at RC and Q configurations

-Fig. 21- Torque/angle characteristics and referent configurations of the arm in one participant with stroke with mild hemiparesis for the PUSH condition whose Fugl-Meyer (FM) score was 56/66. Notations as in Figure 20

-Table 4- R^2 values and slopes of regression lines for both groups of subjects for PUSH and PULL conditions.

	R^2_e	S_{ee}	S_{es}	R^2_s	S_{se}	S_{ss}
PUSH condition						
Healthy	0.88	0.328	-0.044	0.83	0.053	-0.433
	±0.14	±0.183	±0.149	±0.13	±0.213	±0.353
Stroke	0.70	0.193	-0.021	0.72	0.069	-0.405
	±0.26	±0.177*	±0.041	±0.11	±0.235	±0.314
PULL condition						
Healthy	0.86	0.153	-0.120	0.77	-0.143	-0.325
	±0.11	±0.141†	±0.063	±0.10	±.229†	±0.202
Stroke	0.67	0.139	-0.091	0.70	0.113	-0.169
	±0.26*	±0.189	±0.105†	±0.16	±0.397*	±0.128†*

* healthy group versus stroke group; $p < 0.05$ (Student's t-tests)
† PUSH vs PULL; $p < 0.05$ (paired Student's t-tests)

Each regression surface was characterized by two slopes and there were two regression surfaces per condition (see Methods). The slopes for the PUSH conditions were different from those for the PULL conditions,

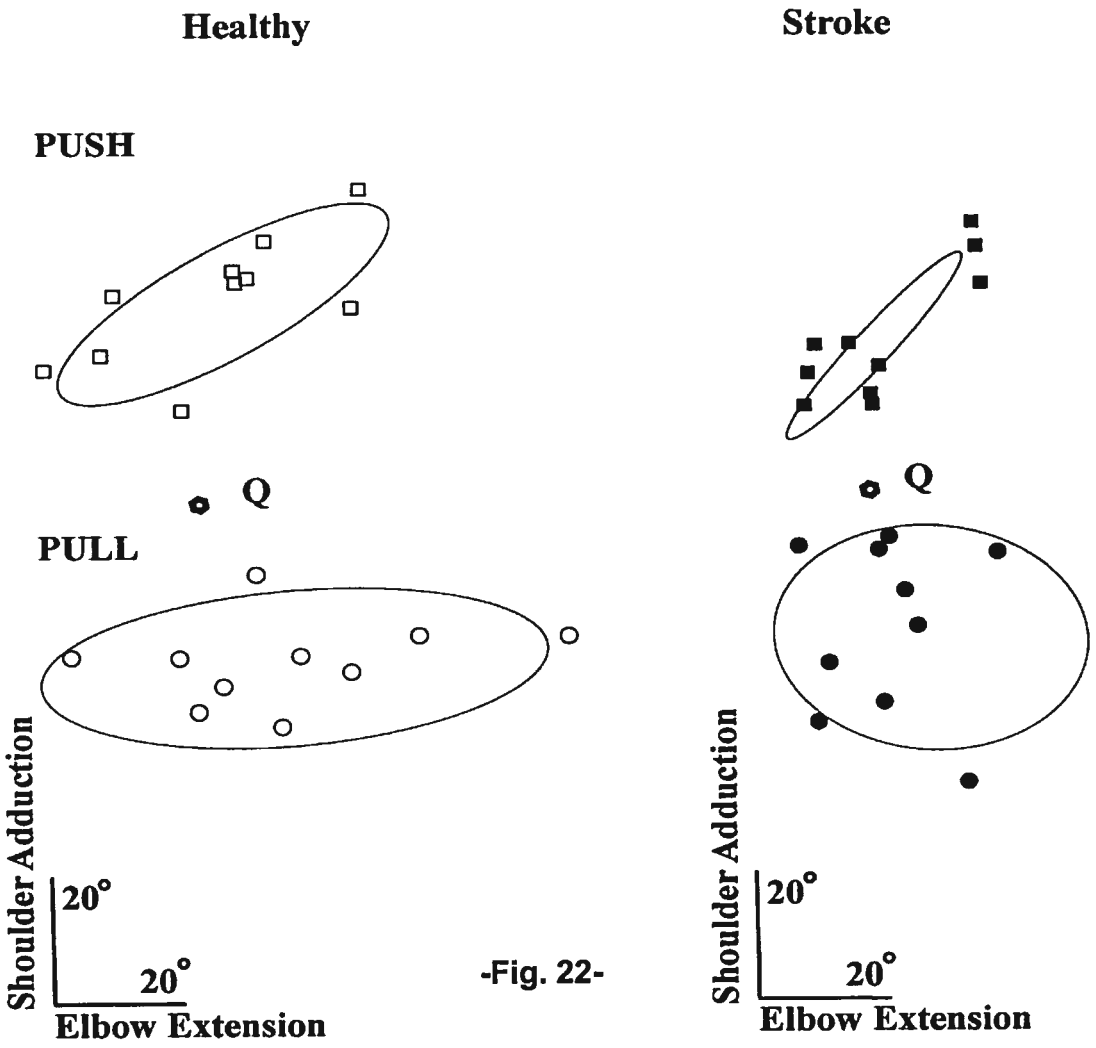
both in healthy subjects and in participants with stroke (Table 4). Between group comparisons revealed significantly lower slopes for patients with hemiparesis compared to healthy subjects for S_{ee} in the PUSH condition and for S_{se} and S_{ss} in the PULL condition (Table 4). The almost zero value of the shoulder slopes in some cases demonstrates the absents of stiffness in this segment. In this case the other segment (the elbow) is stiffer.

The intersection of each regression surface with the angle plane formed a line traversing a range of angles where torque is zero. One zero torque line for each joint was computed (thick straight lines in Figs. 20 A, B and 21 A, B) and the intersection of the two zero torque lines described a point representing the referent configuration (RC) of the arm (Fig. 20 C and 21 C, open circles). For comparison, the actual configurations (Q; Fig. 20 C and 21 C, filled squares) of the arm were plotted on the same coordinate system.

To compare the dispersions of RCs for PUSH and PULL conditions in both groups of subjects, we expressed the position of the RCs of each subject relative to their initial position (Q) and plotted data from all subjects and conditions on the same graph (Fig. 22). The dispersions were described by ellipses of different orientations (slopes) and sizes (areas). In participants with hemiparesis, the area was smaller for PUSH while it was larger and oriented differently for PULL compared to healthy subjects (Table 5).

-Table 5- R^2 values, slopes of regression lines and areas of ellipses describing the dispersion of referent configurations for both groups of subjects for PUSH and PULL conditions.

	R^2	Slope	Area
PUSH condition			
Healthy	0.67	0.46	105.89
Stroke	0.84	0.89	54.24
PULL condition			
Healthy	0.22	0.07	201.86
Stroke	-0.14	-0.23	193.35



-Fig. 22-

Referent arm configurations in elbow vs shoulder angle coordinates for PUSH (squares) and PULL (circles) conditions for healthy (open symbols) subjects and participants with stroke (filled symbols). The slope of the 95% confidence ellipse for each group of points is indicated. Data is shown in reference to the initial configuration of the arm (open hexagon, Q)

4.3 Correlation with clinical data

Although initial torques at the elbow and shoulder were significantly lower in patients compared to healthy subjects, only values of initial torques produced at the elbow for the PULL condition were correlated with the clinical severity of the arm impairment (Fugl-Meyer scores, $r = -0.69$). There were no correlations between arm impairments and slopes of the ICs or their regression coefficients.

5. Discussion

5.1 Basic findings

Both groups of subjects produced similar responses to unloading of the double-joint arm system. Although individuals with arm paresis had lower initial torques (Table 3, Fig. 19), partial removal of the load resulted in distinct final hand positions (Fig. 18) associated with unique shoulder-elbow configurations and joint torques. The net static torque at each joint before and after unloading could be represented as a function of the two joint angles describing a planar surface in 3D torque-angle coordinates, or invariant characteristic (IC; Figs. 20 and 21). Both groups were also able to specify different ICs related to the two different initial conditions (PUSH or PULL). This indicates that individuals with stroke-related brain damage and hemiparesis preserved the ability to adapt their central commands to accommodate changes in the external conditions. One of these commands specifying the referent configuration of the arm was identified based on two criteria: it is measurable by a variable that changes

according to the initial conditions; it remains invariant for the whole set of unloading responses obtained for the same initial condition. The referent configuration of the arm describes the combination of the elbow and shoulder angles at which all joint torques are zero. This analysis thus shows that the net static torque generated at each joint is a function of the difference between the actual and the referent configurations of the arm and that the adaptation to the initial condition was produced by appropriate adjustments in the referent arm configuration. The finding that the slopes of the ICs were different for different initial conditions implies the involvement of an additional central commands.

Despite the preservation of the basic patterns of responses, individuals with stroke damage had a more restricted range of hand trajectories following unloading (Fig. 16), an increased instability around the final endpoint position (Fig. 17) and differences in the dispersion of referent configurations in elbow-shoulder joint space (Fig. 22) compared to healthy individuals. Moreover, in 4 out of 12 patients, ICs could not be identified, implying that these individuals had problems specifying RCs in task-related way.

5.2 "Do not intervene" paradigm

It is known that unloading reactions – smooth transition of the arm to a new position at which equilibrium with the final load is achieved – are very robust and reproducible (Asatryan and Feldman 1965). In non-experienced subjects, they can be observed without any instruction. With repetitions, however, subjects may try to correct unloading responses. The

comparatively short latency of voluntary corrections observed in healthy subjects (about 150 ms) suggests that they could be initiated before the movement offset in the present experiments in which the movement time was about 1 s. Reaction times of individuals with hemiparesis are reportedly longer (later than 400 ms for the elbow, Dickstein et al. 1993) making it less likely that they made corrections of unloading responses. In order to avoid corrective movements in response to unloading, the instruction "do not intervene" is usually given, as was the case in our experiments. In addition, it has been shown that subjects often generate involuntary ("triggered") corrective responses to loading stimuli (Fig. 4 in Feldman and Levin 1995; see also Crago et al. 1976; Newell and Houk 1983). Therefore, only unloading stimuli were used in the present study. Finally, trials in which subjects did not comply with the instruction and made corrections (Fig. 15 C; Fig. 17 E, F) were excluded on-line but the total number of trials determined by the experimental protocol was preserved by repeating these trials.

It has been assumed that non-corrected responses to unloading are produced by the neuromuscular system without changes in the central commands (control variables) as defined by the two criteria formulated in the previous section. In our experiments, muscle torques changed following changes in the load and thus they do not satisfy the second criteria in the definition of control variables. EMG activity of agonist and antagonist muscles also changed following unloading in our experiments (not illustrated), which has been documented in previous studies for

single-joint unloading (Levin and Dimov 1997). In addition, the well-known EMG-force relationship also implies that the EMG activity level could not remain invariant when the load changes and thus, according to our criteria, EMG is not a central command. In previous studies of unloading responses of the elbow joint, it has been shown that control variables determining the shape of the torque-angle characteristic for a given initial condition remain invariant (although they might change with adjustment to a new initial condition). Specifically, it was found that the shape of the characteristic remains the same despite variations in the unloading procedure. Both double-step decreases in the load torque and the use of position-dependent, elastic loads with positive or negative stiffness produced the same type of torque-angle relationship (Asatryan and Feldman 1965; Feldman and Levin 1995). The present experiment examined the relationship between central commands and joint torque and angles of the homologous and adjacent joints for the double-joint system. The results support the suggestion that the referent arm configuration is a control variable for the multi-joint arm system. Specifically, it determines the location of all the torque-angle characteristics of the joints used for a particular task in joint space.

Our results support previous studies showing that each joint torque depends not only on the angle of the homologous joint but also on the angle of the adjacent joint (Beer et al. 2000; Cooke and Virji-Babul 1995; Gribble and Ostry 1998; Hollerbach and Flash 1982). Both mechanical (bi-articular muscles) and neural factors (proprioceptive reflexes between

muscles crossing different joints) are likely responsible for this dependency. These factors make the multi-muscle and multi-joint arm system to function as a coherent whole, rather than as a group of independent elements. Similarly, the concept of the referent configuration implies that this system is also *controlled* as a coherent whole. Moreover, even single-joint control may be organized in the framework of a pre-existing referent configuration.

5.3 Referent arm configurations

Elbow and shoulder joint torques and angles for each level of unloading in three directions for arm pushing and pulling were used to determine the referent configurations (RCs) of the arm (in terms of joint coordinates). The notion that control levels of the nervous system specify an RC and that the neuromuscular system has the capacity to generate EMG activity of multiple muscles depending on the difference between the actual and the referent configurations of the arm to maintain a posture or produce movement are fundamental elements of the λ-model for motor control (Feldman and Levin 1995). Our recordings of two distinct RCs for the two different initial conditions (PUSH and PULL) demonstrate the ability of the nervous system to modify the central command in a task-specific way. In other words, subjects were able to establish different RCs for the same position of the hand to compensate different initial loads. Once the appropriate RC was established, subjects were also able to maintain it thus allowing the neuromuscular system to generate automatic

responses to unloading depending on the difference between the actual and the virtual, centrally specified referent configuration of the arm.

The robustness of the RCs was demonstrated by the high R^2 values of the linear regressions fit to the invariant characteristic (IC) surfaces of each joint in all healthy subjects (Table 4). In most of the patients with stroke, construction of the RCs was also possible although the R^2 values of the ICs were lower compared to the healthy group. Four out of 12 patients had non-significant R^2 values and construction of invariant surfaces and their associated RCs was not reliable, suggesting that a high order control process – specification of the RC was impaired. Deficits in the specification of RCs may be responsible for disruptions in interjoint coordination (Levin 1996; Trombly 1993) and loss of intermuscular coordination (Dewald et al. 1995; Beer et al., 2000) in patients with arm paresis. This conclusion does not conflict with our finding that there was no correlation between R^2 values and the severity of arm motor deficits measured by clinical scales since our analysis focused on higher order control functions that are likely not captured by clinical impairment or functional scales.

Individuals with hemiparesis have difficulties compensating the influence of interactive torques on hand trajectories during voluntary changes in arm position (Beer et al. 2000). The interactive torques are those acting on one segment of the arm following the motion of the other arm segments. A deficit in compensating interactive torques could be responsible for the increased variability and hook-like shape of hand

trajectories in these patients (Fig. 16). Our analysis of RC configurations was made on the basis of the system's steady states. Since interactive torques are velocity- and acceleration-dependent quantities, they are reduced to zero at the end of the movement and thus are not a factor in determining the steady states. The deficit in the specification of the RCs in individuals with hemiparesis is thus independent of possible deficits reported in these subjects in the compensation of interactive torques.

Alternatively, the deficits in the specification of RCs in the participants with hemiparesis might be related to restrictions in their ability to produce movements in certain parts of joint workspace. For example, these deficits may be a manifestation of difficulties in controlling movements made outside of the pathological flexion or extension synergies (Brunnström 1970; Twitchell 1951). In the arm, the extensor synergy is characterized by the activation of a number of muscles of the arm and trunk leading to stereotypical movements involving shoulder adduction and internal rotation, elbow extension, wrist pronation and flexion. An opposite pattern of synergistic muscle activation is observed when attempts to move the arm evoke the flexor synergy. In our experiment, the movement in the PUSH direction could be considered as a movement made within the pathological extensor synergy (in this case, shoulder adduction combined with elbow extension), whereas the movement in the PULL direction was not similar to either synergy (shoulder abduction combined with elbow extension). However, the movement patterns used by participants with hemiparesis did not differ

from those of healthy subjects and our data do not suggest that individuals with stroke used a specific combination of joint rotations characterizing a particular synergy. Dewald et al. (1995) described abnormal coactivation between pairs of muscles of the elbow and shoulder during isometric contractions of the arm, measured by a multi-axial load cell, in 10 patients with hemiparesis of varying severity. Some of the flexor and extensor coactivation occurred in patterns consistent with pathological synergies. For our experiment, we specifically chose an initial position of the hand in an area of the workspace in which pathological movement synergies would not be evoked and the shoulder and elbow joints could function within an angular range in which movement control was not restricted. Our findings that patients did not produce movement patterns consistent with pathological flexor or extensor synergies do not however imply that abnormal EMG co-activation was not present.

The deficits in the specification of RCs may be related to the empirically observed limitations in the regulation of muscle activation thresholds that, in turn, restrict the range of shifts in the torque-angle characteristics, as revealed for elbow flexors and extensors in patients with hemiparesis. Levin et al. (2000) showed that for the elbow joint, even patients with severe motor impairment scores could control elbow flexion and extension movements within a reduced, compared to healthy subjects, angular range lying between the flexor and extensor activation threshold angles. This angular range was described as a 'reciprocal zone' within which movement control resembled that in healthy subjects producing

reciprocal activation of elbow flexors and extensors. Movements attempted to positions outside of this zone were accompanied by excessive co-activation. Extending these observations to the multi-joint system, Levin et al. (2002b) suggested that deficits in the regulation of muscle activation thresholds of adjacent joints would constrain the range of available RCs and thus restrict these joints to act together in specific areas of the workspace. The appearance of pathological movement synergies could also be a consequence of deficits in the specification of thresholds of muscle activation in biarticular muscles (which together with the thresholds of single-joint muscles determine the RC) and the influences of changes in the thresholds due to changes in the position of the adjacent joint. This hypothesis is currently under investigation for the double-joint arm system.

The present experimental analysis has been done in the theoretical framework of the λ model for motor control and the findings have been interpreted in this framework. However, this does not suggest that the data cannot be explained using other theoretical schemes. An essential point of our analysis, however, is that responses to unloading stimuli both in healthy subjects and patients can be fully explained without reference to internal inverse and feedforward models of behavior (see Desmurget and Grafton 2000; Kawato 1999 for recent reviews). Gribble and Ostry (2000) have also demonstrated that a number of phenomena often associated with predictive internal models, namely compensation for interaction torques during multi-joint movement and adaptation to motion-dependent force fields could, in principle, be accomplished using a simple scheme in

which non-force-based control signals are incrementally updated on the basis of positional errors determined by the deviation of the actual hand position from the target (see also Flash and Gurevich, 1997). Thus, viable explanations of motor behavior may involve neither the programming of muscle forces, nor the use of predictive or inverse simulations of the motor output.

6. References

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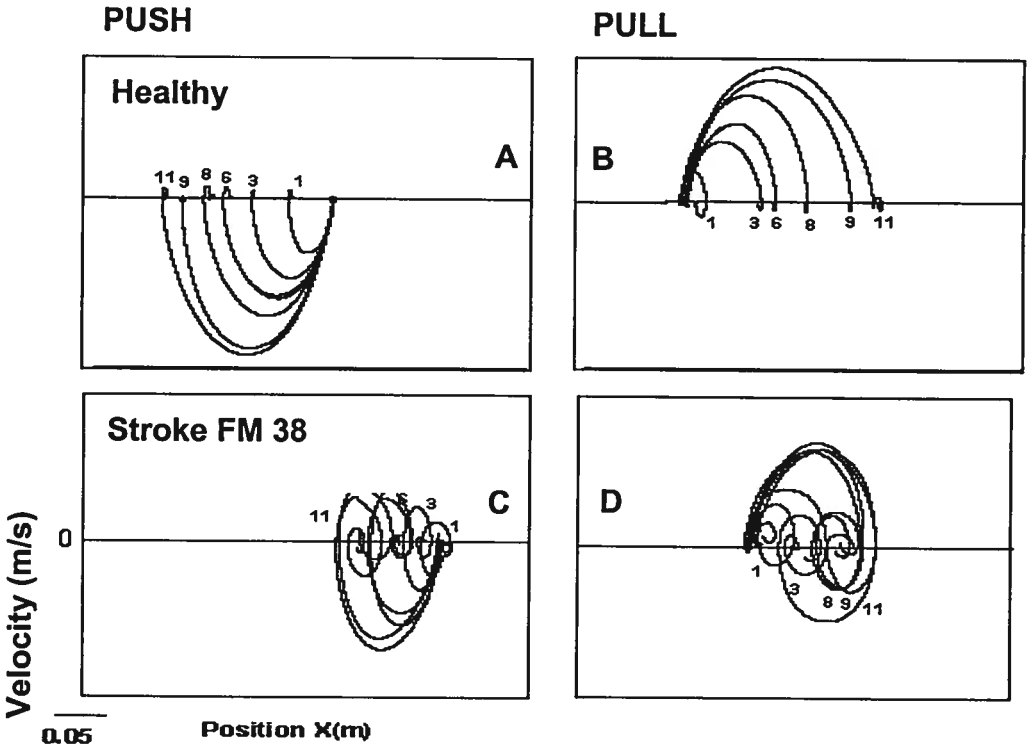
Chapter VI. Instability (The following section will be prepared as a separate article.)

1. Stability indices

The stability of the arm after unloading was analyzed for each subject using the logarithmic decrement of decay of the oscillation of the hand trajectory about the final position. In the healthy group, there was an initial overshoot followed by an undershoot, after which the hand remained steady in a new final position (Fig. 22, top panels). In contrast, unloading in the patient group was characterized by several terminal oscillations followed by a stable final hand position (Fig. 22, bottom panels).



-Fig. 23-

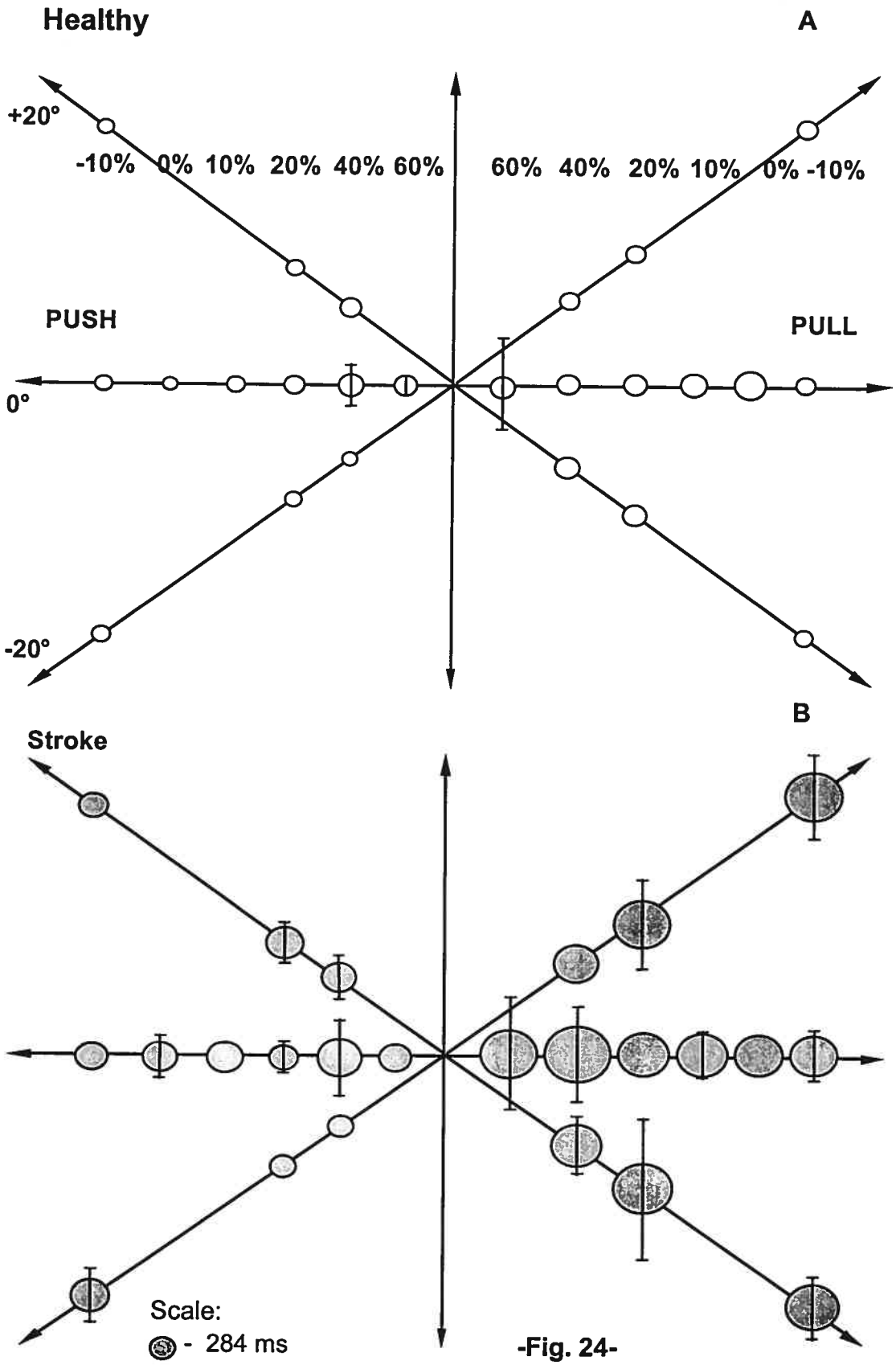
Phase diagrams for 6 different combinations of unloading for PUSH (panels B and D) and PULL (panels A and C) conditions for one healthy subject (panels A and B) and one patient with moderately severe hemiparesis (panels C and D) reflected by a Fugl-Meyer (FM) score of 38/66.

The instability indices varied between conditions. The patient group had significantly larger indices in all but 3 combinations for PUSH and 1 combination for PULL (Table 6). Overall, for the patient group, the mean index was greater for PUSH (369.8 ± 60.9) and PULL (593.0 ± 67.1) than for the healthy subjects (206.4 ± 34.3 and 252.0 ± 40.7 respectively). The instability indices for each combination and condition are summarized in Fig. 23. The figure shows the increase in instability in the indices of patients with stroke, which is more marked for the PULL compared to the PUSH direction (t-test, $p < 0.000$) regardless of the direction of amount of unloading.

-Table 6- The indices of instability for both groups. **Kruskal Wallis ANOVA**

PUSH	Healthy Mean	SD	Stroke Mean	SD	H(1, N=23)	*P*
60% 165°	252.0	121.2	359.7	142.1	5.268	0.022
40% 165°	275.8	257.1	494.7	506.5	3.235	0.072
20% 165°	223.8	91.7	321.7	200.6	2.035	0.154
10% 165°	200.4	71.8	402.5	187.8	6.785	0.009
0% 165°	161.6	46.5	382.9	279.2	9.235	0.002
-10% 165°	191.2	82.9	355.3	155.8	10.400	0.001
40% 145°	167.9	69.4	284.7	117.3	6.785	0.009
20% 145°	183.9	60.9	291.0	140.4	4.188	0.041
-10% 145°	205.6	55.7	425.4	353.9	7.446	0.006
40% 165°	234.8	124.9	381.9	291.4	2.404	0.121
20% 165°	199.2	66.2	417.0	268.5	4.712	0.030
-10% 165°	180.8	73.8	320.4	112.2	10.804	0.001

PULL						
60% 165°	273.0	571.3	630.4	734.5	2.216	0.137
40% 165°	248.8	128.1	732.7	617.3	7.446	0.007
20% 165°	256.6	115.6	567.5	235.5	11.635	0.001
10% 165°	287.5	135.3	563.0	309.7	8.862	0.003
0% 165°	347.2	182.2	534.1	215.3	7.112	0.008
-10% 165°	215.9	76.5	526.3	328.8	14.312	0.000
40% 145°	271.6	125.7	556.3	382.4	4.712	0.030
20% 145°	261.1	117.8	662.0	930.8	5.850	0.016
-10% 145°	205.2	82.5	588.4	414.0	9.615	0.002
40% 165°	214.0	122.0	493.3	222.7	9.615	0.002
20% 165°	213.5	94.1	622.5	588.9	12.496	0.000
-10% 165°	229.6	118.4	639.5	552.8	10.803	0.001



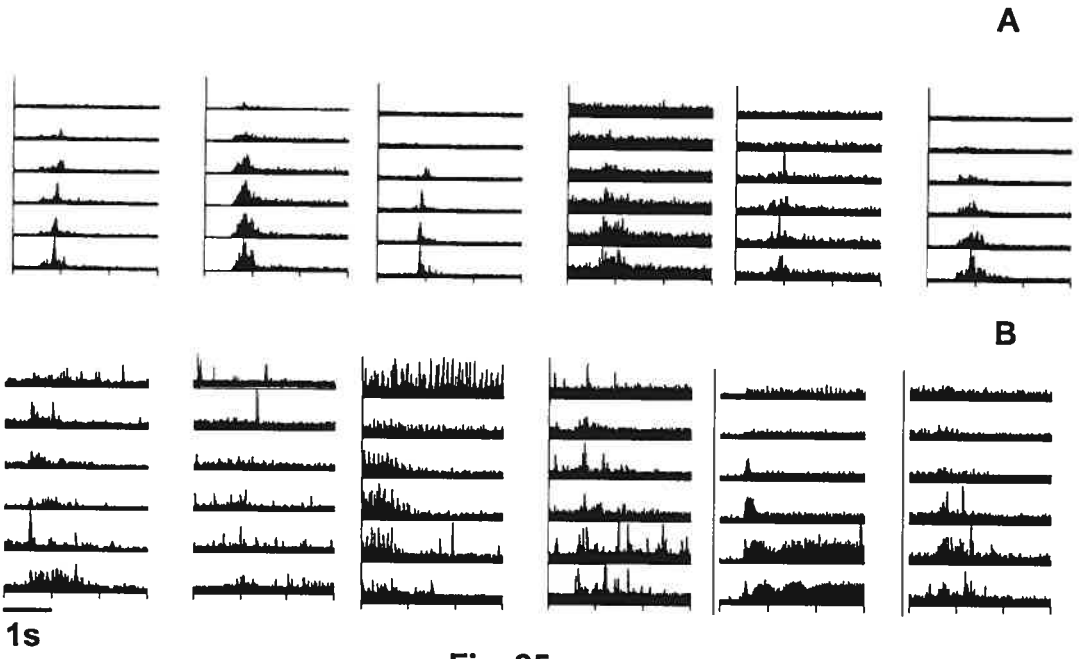
-Fig. 24-

Averaged (± SD) instability indices for healthy (panel A) and stroke (panel B) groups. The data are represented in space for convenience. The diagonals do not correspond to ± 20° from the initial direction.

2. EMG activity related to instability

2.1 Agonist and antagonist muscle groups

The root-mean squared muscle activity of each muscle was computed for each combination of unloading in the PUSH and PULL conditions. Representative data for PUSH 0° combinations are shown for one healthy subject and one patient with stroke in Fig. 24.



-Fig. 25-

Smoothed and rectifies EMG activity for PUSH in one healthy (panel A) and one stroke (panel B) subject. 6 different conditions of unloading are presented with zero deviation of the direction of the final force.

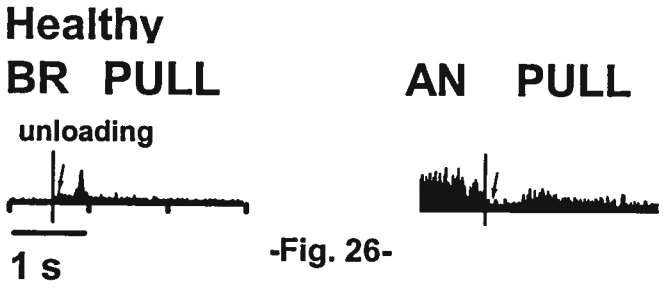
The roles of the two single-joint muscles (PM and DP) acting at the shoulder were easily classified as agonists or antagonists according to whether there was a silent period or a stretch reflex respectively in response to unloading. Thus, PM acted as agonist and DP was antagonist for PUSH and vice versa for PULL. The identification of the roles of the

two single-joint muscles around the elbow was more difficult because of the variability of the EMG activity. In the majority of cases, AN was antagonist and BR was an agonist for PUSH and vice versa for PULL. For the two double-joint muscles (BB and TB), a consistent main role was undetermined since both muscles were mostly active throughout the unloading. For convenience, we assigned BB the role of agonist and TB the role of antagonist for PUSH and vice versa for PULL conditions.

2.2 Coactivation ratios

In healthy subjects, the tonic EMG activity of all muscles changed systematically with the reduction in the external load (Fig. 24, top rows). The modulation of EMG activity was less systematic in the patients with stroke (Fig. 24, bottom rows).

We visually investigated these graphs in order to determine the agonist (AG) and antagonist (ANT) muscle groups. The roles were determined as follows: after the removal of the external load (approximately 0.5 s after initiation of the record) the activity of the muscles participating in the opposition of the load decreased abruptly until the hand stabilized in the final position. This movement phase continued in response to unloading usually for less then a second $\approx$ 0.9 s). In contrast, muscles not participating in opposing the load produced a phasic burst in the first 30 ms after unloading due to the mono-synaptic stretch-reflex (Fig. 27). These muscles were considered as antagonists.



-Fig. 26-

Specification of the muscle roles (Ag/Ant). Example of an increase in muscle activity in brachio radialis (BR) for the PULL condition and a decrease of the muscle activity in anconeus (AN) for the same condition in one healthy subject.

Mean coactivation ratios for each AG/ANT pair were computed by calculating the ratios after unloading for the combination where the direction of the final load was unchanged (0°). A correlation analysis was performed between the instability indices and the calculated coactivation ratios. We did not find any significant correlation between the instability indexes and coactivation values in healthy subjects. On the other hand a significant correlation ($r = 0.74$) between the instability indexes and ratios of PM/DP for PULL was found in the patient group. In addition significant correlations were found between the clinical impairment (FM scores) and PM/DP ratio for PULL (-0.56), sum of all three ratios for PULL (-0.73) and the sum of all three ratios for PUSH (0.64).

Chapter VII. Discussion

1. Basic findings: referent configuration of the double-joint arm

In the present study the postural and movement control of the double-joint arm of healthy subjects and patients with right hemiparesis due to unilateral stroke-related brain damage were investigated. The participants were asked to match their force produced by the right hand against an externally imposed load. The load was removed without notification, while the subjects were instructed not to intervene to this perturbation. As a consequence of unloading, the arm made a single, smooth, transition to a new point in space, at which point the hand stabilized in a new final position at equilibrium with a new final load. By following the "do not intervene" instruction, it was assumed that participants kept the parameters of the central command constant during the experiment. This was confirmed by the lack of inflection or reversal points in the velocity/position diagrams recorded in both groups of subjects (Fig. 2). Elbow and shoulder joint torques and angles for each level of unloading in three directions were used to determine the referent configuration (RC) of the arm (in terms of joint coordinates) determined by the central command. The specification of double-joint RCs is a fundamental element of the λ-model of the EP hypothesis for motor control. Our recordings of two RCs for the two different initial directions

(PUSH and PULL) demonstrated the ability of the CNS to modify the central command according to changes in the external conditions. The robustness of the RCs was demonstrated by the high R^2 values of the linear regressions fit to the invariant characteristic (IC) surfaces of each joint in all healthy subjects.

In most of the patients with stroke, construction of the RC was also possible although the R^2 values of the ICs were lower compared to the healthy group. Two out of 12 patients had non-significant R^2 values and construction of invariant surfaces and their associated RCs was not possible for each condition. This finding does not mean that these patients were not able to specify any RCs but may reflect their inability to produce appropriate RCs in certain parts of external space. Indeed, the restriction of the workspace in which RCs can be specified may be related to difficulties in controlling movements outside of pathological synergies. Brunnström (1970) described a synergy of flexion in the upper limb, characterized by abduction at the shoulder, flexion at the elbow, supination of the forearm and flexion at the wrist joint. Interesting results which could support the idea of the muscle synergies can be found in the work of Dewald et al. (1995) in which increases in the shoulder abduction torque were related to increases in strength of the elbow flexors (flexor muscle synergy).

An alternative interpretation is suggested in the λ - model. According to the model, movement control is accomplished by the

regulation of the muscle activation thresholds expressed in angular or muscle length coordinates (see the Introduction). Hence, impaired movement control following a stroke could be explained by deficits in the specification or regulation of this threshold. Levin and Feldman (1994), using stretch applied to spastic elbow flexors at different velocities, showed that some patients with hemiparesis were unable to shift their SR-thresholds outside of the biomechanical limits of the joint. This impairment in regulation may contribute to the deficit in the voluntary range (i.e. restricting the range of voluntary movement control, Levin et al., 2000).

Another control deficit reported in patients with hemiparesis is the inability to specify adequate coactivation in appropriate joint ranges under specific conditions. Thus, Levin and Dimov (1997) observed abnormal co-activation while unloading either elbow flexors or extensors in patients with hemiparesis which led to increases in the instability at this joint. Based on the assumptions of the λ-model, the authors proposed that the CNS of some stroke patients could be hampered in the specification of C-commands which results in inappropriate determination of muscle co-activation zones. Later the idea of alteration of the muscle coactivation zones was further elaborated by Levin et al. (2000). Angular zones in which reciprocal voluntary muscle activation could occur were drastically diminished in stroke patients, which is in accordance with the assumption of a decrease in the range of regulation of the SR-thresholds for agonist and antagonist muscles acting around a joint.

For both conditions, the torque produced by the elbow and shoulder and applied at the hand was a function of the difference between the referent and actual configuration. This means that the torque increased proportionally with the distance between RC and Q.

In this experiment, only one referent configuration per direction was constructed due to the lengthy recording procedure. Thus, we cannot draw any conclusions about whether stroke results in limitations in the range of regulation of RCs. For this purpose we would need to construct a series of RCs in a single session (see Limitations of the study - Section VIII).

2. Instability

Based on the finding that even the slightest perturbations, which were observed in some patients with very low initial loads, caused adequate motor responses, it is unlikely that the increased instability in the patients with stroke is due to lower initial torques. Arm instability in all subjects was expressed in terms of the inverse of the decrement of decay in the oscillations about the final hand position so that higher values reflected increased instability. These instability indices showed higher values for the PULL condition in comparison with PUSH (a feature observed in both groups). These agree with findings of greater instability in different parts of the workspace of the elbow joint (Levin et al., 2000).

In our experiment the movement in the PULL condition was characterized by horizontal shoulder abduction combined with elbow

flexion or extension. This condition required participants to move their hand away from their bodies or in the case of stroke patients to move out of the pathological synergy. This may trigger different pathological motor responses as instability may be greater when the arm moves out of the range (ipsilateral workspace) in which control is possible. However different levels of unloading did not trigger different motor responses since the patterns of movement in response to unloading were consistent within each subject and did not vary with the level of unloading.

Archambault et al. (2003) demonstrated the interdependence of the torques of the elbow and shoulder during double-joint movement. The torque in the elbow is dependent on the position of the shoulder and vice versa, which results from the changes in muscle activity. Double-joint muscles as well as single-joint muscles changed their activity according to changes in the angle of the neighbouring joint (e.g. anconeus muscle activity was modified with respect to the shoulder position) in healthy individuals. Earlier, Beer et al. (2000) used inverse dynamics modeling to demonstrate pathological changes in the control of elbow/shoulder interaction torque control in hemiparetic patients. The authors found that patients retained their ability to produce and voluntarily modulate joint-torques (which can also be explained by their ability to produce RCs, as described in the present project). They attributed the inaccuracy of the limb positioning not to weakness, spasticity or stereotypic movements (muscle synergies) but to disruption of the central command which

determines, in a feedforward manner, the interaction between joint torques that arise during mutli-joint movements. They hypothesized that disruptions in the central command occur due to the brain lesion or to disuse which followed from damage to the internal representation of the limb. Even though the idea of predetermination of muscle torques is not compatible with the theoretical basis of the λ-model the results from this study can be interpreted according to λ-model. The idea that the CNS predetermines zones of muscle co-activation for joint stability (Levin and Dimov, 1997) suggests that appropriate interactive torques are regulated within these zones. Thus the interpretation based on the λ-model suggests that patients cannot regulate appropriate coactivation within pre-determined angular zones.

Several studies (Archambault et al., 2003; Lestienne et al., 2000) observed the dependency of EMG signals on RC and Q configurations. The central command determining the RC configuration in space would influence the excitability of stretch-reflexes at different joint angles. Archambault et al. (2003) showed that the EMG activity is modulated according to the differences in coordinates of RC and Q positions as well as the coactivation command. The modulation of the muscle signals followed a pattern demonstrating that control of the muscles originated by the interaction of the arm with the external environment suggested that control signals do not directly program individual muscle activation. In our experiment, the sudden decrease in the external load triggered a mono-

synaptic reflex in the antagonist muscle followed by long a latency burst as well as a short-latency decrease in the agonist activity known as a silent period. The presence of reflex responses does not imply a change in the threshold levels of muscle activation but represent triggered responses mediated by the stimulus. Our data showed that the control of reciprocal muscle activation and coactivation which is thought to be the basis for stability during movements, is disrupted in stroke. We did not find any relationship between the instability indices and coactivation ratios in the healthy group suggesting that healthy subjects can regulate coactivation throughout a wide range. However, patients with greater impairment used more coactivation in the PULL conditions but less coactivation in the PUSH condition. This may indicate that following a stroke, regulation of coactivation is disrupted throughout the angular ranges and end-point positions investigated in this study. In a simpler single-joint systems, Levin and Dimov (1997) showed that the regulation of coactivation around the elbow was disrupted. Our data for the double-joint system suggests that coactivation is also disrupted and that the relationship between coactivation at two joints and end-point stability are not single valued but may be related in some complex way.

Overall, our data suggest that patients with more severe impairment used relatively less coactivation and had greater end-point instability, again suggesting a deficit in the regulation of coactivation.

3. Peripheral influences on the central command

The parameter λ has both central and peripheral (λ_{cv} and $\lambda_{feedback}$) components. Even though the central component can be specified independently of peripheral influences, the final limb position will be influenced by peripheral feedback as well by the modulation of reflex excitability. For example, in healthy subjects, functionally appropriate modulation of short latency leg muscle reflexes occurs during gait, while in stroke patients the reflex modulation is severely impaired (Faist et al. 1999). Changes in cutaneous withdrawal reflex responses in the upper extremity have also been observed in hemiparetic patients (Dewald et al., 1999). While the normal response consists of elbow flexion, shoulder extension and adduction, the impaired limb reacted with shoulder flexion and abduction. In addition the motor responses were characterized by later onset of muscle activity. These changes could be associated with alteration of corticospinal pathways or reorganization of the cortical neuronal elements. Another factor that could influence the specification of λ is for example an impairment of the perception of the external world (Small et al., 1994). In this case the feedforward mechanisms would be violated, and the arm position would not correspond adequately to the externally imposed load, which may result in a pathological motor response during unloading. Another factor could be alteration of the biomechanical properties of the muscles (Hufschmidt and Mauritz, 1985). Hufschmidt and Mauritz (1985) proposed that the intrinsic muscle stiffness

would increase due to the structural alteration of the muscle fibers, namely, an increase in the connective tissue with a decrease in the number of muscle fibers which leads to an increase in stiffness. Thus another factor influencing the response to unloading could be an increase in the stabilization time around the final position due to slower muscle response time resulting from changes in properties of intrinsic muscle fibres. Similarly the final position after unloading in the patients may be less stable because of alterations in the RC position caused by changes in the central processing of peripheral information. This may also contribute to the increase in the oscillations around the end-point in addition to a deficit in the regulation of the C command. All of these factors could contribute to deficits in the specification of the RC position of the limb, because of inadequate "calibration" in the CNS. In our experiment this inappropriate specification of the RC may require more reaction time for restoring of the equilibrium expressed by an increase of oscillations.

Chapter VIII. Limitations of the study

In our study the IC surfaces were presented as planes. However, the torque angle interaction for single-joint movement forms a curved line because of the non-linear characteristic of the muscles. This feature is valid also for the double-joint system. However, we obtained high R^2 values using linear fits but we cannot rule out that better fits (R^2) would have been obtained with 2^{nd} and 3^{rd} order models.

Although visibly, the RCs of the stroke group occupied a more restricted space in terms of elbow angular coordinates, the reproduction of only two central commands was not sufficient to demonstrate a significant difference in the range of regulation of RCs between healthy subjects and the patient group. In order to better map out the differences between intact and impaired motor control, one needs to measure RCs from several different initial conditions (in this case different initial forces with diverse initial positions). However, the large number of repetitions and the long length of the experiment make such an experiment in patients with stroke impractical.

Another limitation is that by calculating the inverse logarithmic of decay, we were not able to distinguish stroke subjects with increased level of oscillations from those with delay in their reflex response due to intrinsic muscular factors.

Chapter IX. Conclusion

Double-joint movement control and its disturbances following a stroke were investigated in the present study. As a theoretical framework we used the λ-model of the equilibrium point hypothesis. Significant differences in most of the kinematic and dynamic characteristics of movements between healthy and hemiparetic subjects were found while the specification of double-joint ICs was generally preserved in patients. Few studies have examined impaired motor control in double-joint systems. One of our major findings was increased instability of the arm as assessed by the unloading paradigm. This lack of postural stability can be regarded as a consequence of the inability of the CNS to produce an adequate response to external perturbations. The increased instability in patients is a response of the CNS acting in marginal zone of its capabilities. In this sense, the λ – model could be used as a tool to find the optimal range in which patients could reproduce adequate RCs, which would have impact on the rehabilitation of these patients.

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ANNEXES

ANNEX 1

INSTITUT DE RÉADAPTATION DE MONTRÉAL

affilié à l'Université de Montréal

CERTIFICAT D'ÉTHIQUE

Par la présente le comité d'éthique de la recherche de l'Institut de réadaptation de Montréal atteste qu'il a évalué le projet de recherche intitulé : **Le contrôle de la posture et de la stabilité du bras multi-articulaire, présenté par Madame Mindy F. Levin Ph.D, Monsieur Philippe Archambault, Msc, étudiant et Monsieur Pavel Mihaltchev, étudiant.**

Le comité d'éthique; composé de :

Dre Ayda Bachir, Université de Montréal;
Mme Marie-Ève Bouthillier, conseillère en éthique;
Madame Nancy Dubé, clinicienne IRM;
Monsieur Anatol G.Feldman, chercheur;
Mme Lisette Gagnon, présidente par intérim du Comité;
Dr Bernard Leduc, physiatre IRM;
Mme Mélanie Lefebvre, avocate, conseillère versée en droit;
Mme Anik Nolet, avocate, conseillère versée en droit;
Madame Monique Provost, A.P.Q.
Mme Bonnie Swaine, chercheure.

a jugé cette recherche acceptable sur le plan de l'éthique.

[redacted]

Lisette Gagnon,
Présidente du Comité d'éthique de la recherche

5-02-02
Date

LG/mm