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PREFACE

New technologies and recent findings in neuromotor control can provide useful tools to conceive new diagnosis procedures and to restore impaired motor functions. For this purpose the basic understanding of natural sensory-motor integration must be combined with the knowledge of neuromotor disorders and with the advanced technologies used in rehabilitation.

This is an interdisciplinary problem that requires the contributions of different researchers working in complementary areas (neurophysiology, neurology, rehabilitation, bioengineering and advanced technology).

This approach is typical of the "International Society of Electrophysiology and Kinesiology (ISEK) which is a multidisciplinary organization composed of scientists from the already mentioned fields working in basic research as well as clinical application.

At the IX meeting held in Florence attended more than 400 people including the most outstanding scientists in this field, who provided the main recent results of their research.

Therefore this book provides an interesting updating of research, developments and clinical applications going on the following topics: Neuromotor Control, Electromyography, Functional Electrical Stimulation (FES), Motor Unit Control, Neuromuscular Diseases, Rehabilitation, Muscle Fatigue, Kinesiology, Motion Analysis and Ergonomics.

I would like to express my gratitude to the Congress Board and to the ISEK Council for their contribution to the designing and organization of the meeting and special thanks to CE.S.P.RI, Carla Finocchiaro and Matilde Cordella for their enthusiasm and irreplaceable help.

Prof. Antonio Pedotti
Congress Chairman

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MOTOR CONTROL IN RELATION TO POSTURAL AND LOCOMOTOR CONTROL

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Biologically useful motor behaviors do not consist of isolated actions of single limbs, but are emergent properties of the total system that are not described by the sum of the properties of its parts (Brookhart, 1979). It is obvious that the results obtained from "reduced preparation" such as spinal and decerebrate animals cannot be directly applied to the full understanding of motor behaviors of intact animals. However, with the recent advancement of motor control concepts such as command hierarchies, which consist of high (strategy), middle (synergy) and low centers (reflex) for motor control (Brooks, 1986), and two modes of organization of the coordination between posture and movement (Massion, 1991), along with an accumulation of detailed information related to the neuronal framework descending from cortical, subcortical and cerebellar motor structures to the brain stem (Brodal, 1969; Kuypers, 1981), and putative neurotransmitters acting on individual cells in the brain stem (Garcia-Rill, 1986; Striade et al., 1990), it has become possible to get insights into the realistic operation of the central nervous system in relation to postural and movement control including their interaction and/or integration. In this chapter, the neural framework for synergistic activation of a number of skeletal muscles, which is the principal basis of the postural and locomotor control, will be reviewed and discussed.

Motor Hierarchy: Reflex, Synergy and Strategy.

Reflexes are automatic responses and are best defined as responses evoked with a great probability by particular stimuli originating within or outside of the body. Magnus (1924) assumed that several reflexes should add algebraically to produce the final postural attitude. However, the inference about a particular reflex could not be evaluated by removing one at a time. The spinal reflexes have been studied extensively aiming at elucidating "wiring diagram", "reflex center" and the neurons controlled by excitation and inhibition through the reflex arc. The term "center" is defined as an organized structure of subunits within a larger organization. Spinal centers are governed by hierarchic supraspinal ones. It has been shown that a large number of reflex afferents and supraspinal tracts employ common interneuronal circuits to control the outputs of the final executive organs. Synergy is related to the principles of the least interaction in a complicated multi-level system, which consists of an aggregation of subsystems having relative autonomy. Bernstein (1967) assumed the presence of some higher motor centers and the motor hierarchy. Each level of command relies on the capabilities of the one beneath it and equally relies on interactions with those above and below. For realization of movement, the higher motor centers have only to select and possibly tune the synergies with resultant reduction in the number of degrees of freedom of the executive organs. It has been suggested that movement training is the development of corresponding synergies which reduce the number of parameters requiring individual control. A new synergy is made on the framework of a small number of basic synergies and inherent neurophysiological mechanisms which lower the number of independent parameters of the controlled system. Strategy corresponds to motor plans or instructions. The highest level operates in the association cortex, and elaborates perceptions and overall motor plans. The decision to perform a task in a certain way is considered to be choices of strategy. Strategy is then converted into motor programs (tactics) located at the middle level. Descending pathways from the middle level, which consists of the sensorimotor cortex, the cerebellum and the brain stem, carry tactical instructions to the spinal cord, the lowest level, where they are coordinated and finally translated into properly timed commands for muscle force bearing on the active joints.

<u>Two modes of coordination between posture and movement.</u>

The postural adjustment associated with movement are called anticipatory because their onset occurs prior to the onset of the disturbance of posture and equilibrium resulting from the

movement (Massion, 1991). Massion (1991) proposed two modes of coordination between posture and movement. In the first "hierarchical" mode, the pathways controlling the movement performance give off collaterals acting on the postural network responsible for the anticipatory postural adjustments. The second mode is a "parallel" mode, where the postural adjustment and movement are controlled by parallel pathways. In this mode, the postural changes often occur shortly before the movement onset. The question which remains to be answered is at what level the coordination between posture and movement is organized. Massion (1991) suggested that premotor and medial frontal areas including the supplementary motor area (SMA) play a leading role in the anticipatory postural adjustments associated with hand movements. In addition, contribution of basal ganglia is mentioned because in the parkinsonian patients, both the postural reaction and coordination between posture and movement are impaired. These results indicate that the possibility of using the collateral pathway (gate) from the movement-related one and its gain to the control of posture depend on a supraspinal control from the SMA area and the basal ganglia. It is possible that the postural networks responsible for the anticipatory postural adjustments is located at quite a low level such as the brain stem and spinal cord, where the neuronal network involved in the postural reactions are located (Mori, 1987, 1989). It has been well established that the ventromedial system descending from the brain stem plays a crucial role in the control of posture and locomotion (Lawrence et al., 1968a,b; Kuypers, 1981).

<u>Descending cortical and subcortical pathways to the spinal cord and their functional implications to the control of posture and movement.</u>

The descending cortical and subcortical pathways to the spinal cord constitute the main routes by which the brain can govern the spinal motor mechanisms. These pathways correspond to those descending from the middle level of a command hierarchy (Brooks, 1981). The corticospinal fibers arise primarily from the pre-and postcentral gyrus. Subcortical pathways originate mainly from the brain stem. Lawrence et al. (1968a,b) divided the descending pathways into the dorsolateral system and the ventromedial system. The dorsolateral system includes the corticospinal tract and the rubrospinal tract. The ventromedial system includes the vestibulospinal, tectospinal and interstitiospinal tracts in addition to the reticulospinal tract. Cortical, subcortical and spinal inputs converge on the segmental interneurons and motoneurons. In addition to interneurons, the dorsolateral system projects terminal fibers to motoneurons innervating distal muscles, while the ventromedial system projects terminal fibers to the motoneurons innervating axial and proximal muscles. At the intermediate zone of the spinal gray, not only segmental interneurons but also propriospinal interneurons with short and long axons are located (Kuypers, 1981). The terminal distribution of the corticospinal fibers in the internuncial zone overlaps that of the dorsolateral and ventromedial fibers systems. It is obvious that integration of descending inputs takes place at the level of spinal interneurons and motoneurons, and it is a prerequisite for proper coordination of posture and movement. Lawrence et al. (1968a,b) suggested that the ventromedial system functions as the basic system by which the brain exerts control over movement. This control is especially concerned with maintenance of upright posture, integrated movements of body and limbs and with directing the course of progression. In contrast, the dorsolateral system provides the capacity for further fractionation of movements as exemplified by individual finger movements.

Higher level control of posture and movements obviously requires use-dependent refinement of neuronal subsets yielding coordinated synergistic muscular actions. A number of evidence has indicated that the reticulospinal fibers play an impotent role in a synergistic activation of a number of skeletal muscles. Furthermore, corticoreticular connections by corticospinal collaterals (Keizer et al., 1984 and 1989), vestibuloreticular and tectoreticular connections have been identified along with cerebelloreticular connections by way of the fastigio-reticular pathway (Brodal, 1969). Electrophysiologically, reticulospinal neurons appear to be subject to monosynaptic convergence of cortical, tectal, and cutaneous inputs and to be monosynaptically excited by vestibular impulses as well (Magni et al., 1964; Peterson, 1984). From these results, it can be conceived that the "interface" integrating posture and movements such as locomotion exists in the pontomedullary reticular formation. The pontomedullary reticular formation represents a

phylogenetically old part of the brain, and in fact, is regarded as a rostral extension of the spinal intermediate zone into the brain stem (Brodal, 1969).

The behavioral correlates of the medial reticular formation.

With regards to the two modes of coordination between posture and movement (Massion, 1991), the studies by Keizer et al. (1984, 1989) are worth mentioning. They demonstrated in cats that up to 30% of the corticospinal neurons in the medial and anterior parts of the precruciate motor area represent branching neurons which distribute collaterals to the contralateral medial reticular formation of the lower brain stem. They demonstrated in monkeys that the rostral part of the precentral cortical area and the SMA contains branching corticospinal neurons which distribute collaterals to the medial reticular formation of the lower brain stem. The corticobulbar fibers including the corticospinal collaterals were distributed to the nucleus reticularis gigantocellularis (NRGc), the nucleus reticularis pontis caudalis (NRPc) and the nucleus reticularis ventralis of the medulla oblongata. These nuclei contain many reticulospinal neurons. These findings suggested that the corticospinal collaterals along with the corticobulbar fibers establish indirect cortico-reticulo-spinal connections. The corticoreticular branching neurons were present mainly in these parts of the sensorimotor cortex which carry representation of back, neck and shoulder movements (Nieoullon et al., 1976), while virtually none were present in the areas, which carry the representation of wrist, forepaw and hindlimb movements.

Aforementioned results indicate that the corticospinal neurons with collaterals to the medial reticular formation are mainly involved in the control of neck, back, and shoulder movements. A single reticulospinal cell might function to facilitate muscles involved in head turning and forelimb muscles supporting the head in its new position. This would be in accordance with the earlier notion that the brainstem reticulospinal system mainly controls axial, body and integrated limb-body movements (Lawrence et al., 1968a,b). The control of the activity of neck and back muscles was proposed to take place along two routes: (1) by way of the direct corticospinal fibers to spinal interneurons and motoneurons, and (2) by way of indirect cortico-reticulo-spinal connections.

Because of the relatively large diameter of the reticulospinal fibers, the indirect cortico-reticulo-spinal activity was supposed to reach the spinal interneurons and motoneurons earlier than the direct corticospinal activity. The slower direct corticospinal connections may then superimpose a finer regulation upon this brain stem control. In certain cases, such a dual control can also be made by the same cortical neurons and they may function as cells of origin of the hierarchical mode and of the parallel mode related to the coordination between posture and movement (Massion, 1991).

Reticulospinal tract originating from the pontomedullary reticular formation are the major component of the ventromedial system involved in the control of axial and proximal muscles. A majority of reticulospinal neurons projecting to the cervical spinal ventral horn were branching neurons that also sent axons to the lumbar spinal cord. The widespread branching of individual reticulospinal neurons were thought of as imposing a type of "hard-wired coordination" within the descending reticulospinal system in that activation of a single reticulospinal neuron leads to direct synaptic input to neurons located at multiple levels along the spinal neuraxis, resulting in various patterns of motor synergies that underlie postural and orienting reflexes. From the high degree of collateralization of reticulospinal fibers and their characteristic termination in areas containing propriospinal neurons, Kuypers (1981) suggested that reticulospinal fibers were well suited for synergistic activation of a large number of muscular elements. Our study on the fine morphology of the reticulospinal fibers (Matsuyama et al., 1988), including the arborization of axon collaterals in the spinal gray matter and the location of terminals closely apposed to spinal neurons, provided morphological bases on "hard-wired coordination" of the reticulospinal systems. We have also shown that reticulospinal fibers play a crucial role in setting the excitability of the brain stem and the spinal cord, which is necessary for proper execution of motor behaviors (Mori, 1989; Takakusaki et al., 1989).

It is our current understanding that postural and locomotor control systems do not reside clustered in particular sites, instead, they are formed by subprograms that are distributed in the central nervous system (Garcia-Rill, 1986; Mori, 1987). It has been shown that postural and locomotor control systems share their subprograms at the levels of the brain stem and the spinal

cord (Mori, 1987, 1989). The neuronal circuitry related to each of the subprograms will be further refined by the acquisition and execution of the learned movements. Recent electrophysiological (Asanuma et al., 1991) and morphological evidence (Keller et al., 1992) supported the following hypothesis; experience-mediated changes in patterns of synaptic circuitry are involved in information storage related to motor learning and memory.

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FEEDFORWARD CONTROL OF POSTURE AND EQUILIBRIUM DURING MOVEMENT PERFORMANCE

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Introduction

Motor actions can be subdivided into two categories, depending on the function they serve. One function is to reach a goal, such as grasping a glass, which involves displacing the hand through a trajectory toward the target. The initial position is disrupted until the object is grasped and a new final position is built up.

The second function of motor actions is just the opposite. It consists of stabilizing a reference value against external or internal disturbances. The reference value to be stabilized can be either the position or the orientation of a segment such as the forearm, the head or the trunk, or that of the whole body. Another example of a reference value which needs to be stabilized is the center of gravity position in relation to the supporting surface : this stabilization is necessary to maintain equilibrium. Stabilizing all these reference values involves the use of error detecting sensors which induce appropriate corrections.

In most of our motor actions, functions of both categories are served at the same time. Raising a leg while standing is a good example of a motor action during which these two different functions are controlled simultaneously. The performance of the leg movement is preceded by a displacement of the center of gravity onto the supporting leg, which serves to maintain equilibrium. This results from an external rotation of the supporting leg around the ankle joint, which should theoretically lead to the trunk and head being inclined laterally. However, the head and/or trunk axis remains vertical due to a counterrotation of the head with respect to the trunk or the trunk with respect to the leg (Mouchnino et al., 1992).

Co-ordination between posture, equilibrium and movement is thus taking place, whereby the equilibrium continues to be efficiently controlled and the orientation of the position and that of the body segments maintained during movement performance.

In this paper it is proposed to analyze the following aspects of posturo-kinetic coordination : 1. Posture and equilibrium control. 2. Equilibrium control during movement. 3. Stabilization of the position and orientation of the body segments during movement.

1. Posture and equilibrium

Before beginning to analyse the goals of the anticipatory postural adjustments associated with voluntary movement, it might be appropriate to first outline the main differences between posture and equilibrium.

Body posture is mainly built up against the force of gravity. Each species has a genetically predetermined reference posture, namely standing.

The human standing posture is not a rigid block oscillating like an inverted pendulum around the ankle joint, although it can behave in this way (Gurfinkel, 1973; Nashner and McCollum, 1985). It consists rather of a set of superimposed modules (legs, trunk, head) build up from the ground to the head, each linked to the next one by a set of muscles having its own central and peripheral control. Specific defects in the postural maintenance of some modules have been described in the literature. Martin (1967) has reported that postencephalitic patients can show a forward bending of the head while resting, but adopt a normal erect position when asked to do so by performing a voluntary movement. This finding indicates that the automatic postural stabilization of the head can be dissociated from the voluntary control of the head muscles. The reference position of a module such as the head or trunk can vary depending on the context. The head for example is a module which involves three categories of sensors : the visual sensors, which collect information about the external world, the labyrinthine sensors, which detect the gravity vector, and the neck proprioceptors, which specify the head position with respect to the trunk. Depending on the context, the head position can be regulated with respect to either the external world (during gaze fixation), the gravity vector or the trunk position (Berthoz and Pozzo, 1988; Pozzo et al., 1989; Nashner et al., 1989; Amblard et al., 1990; Assaiante and Amblard, 1990). Postural control therefore involves multiple, parallel commands for regulating the position of each individual module depending on the task and on the environmental constraints.

In addition to the modular control, there exists an overall postural control, as shown by the decerebrate rigidity which affects all the antigravity muscles. The central organization of this whole-body postural control, which has recently been analyzed in the cat by Mori (1987; 1989), takes place at the brain stem level.

Equilibrium control in fact constitutes an additional postural constraint. The positions of the various body segments can vary within a wide range provided that under static conditions, the center of gravity projection onto the ground remains located inside the supporting surface, i.e. the area between the feet. This reference value is strictly regulated. Lacquaniti et al. (1984) have shown that in the standing cat, the center of gravity projection remains at the same place when the supporting surface is inclined forward or backward. The main reference value to be regulated in the cat is the leg axis, however (Lacquaniti et al., 1990). In man, several lines of experimental evidence suggest that the center of gravity projection onto the ground undergoes strict regulation during oscillations of the supporting platform (Gurfinkel et al., 1981; Dietz et al., 1989; Gollhofer et al., 1989) or during upper arm (Martin, 1967) or trunk (Babinski, 1899; Crenna et al., 1987) movements.

The following scheme (Fig. 1) indicates how the equilibrium is regulated (Massion, 1992). Four points need to be considered here.

1. The reference value to be regulated, which is the CG projection onto the ground, is centrally controlled.

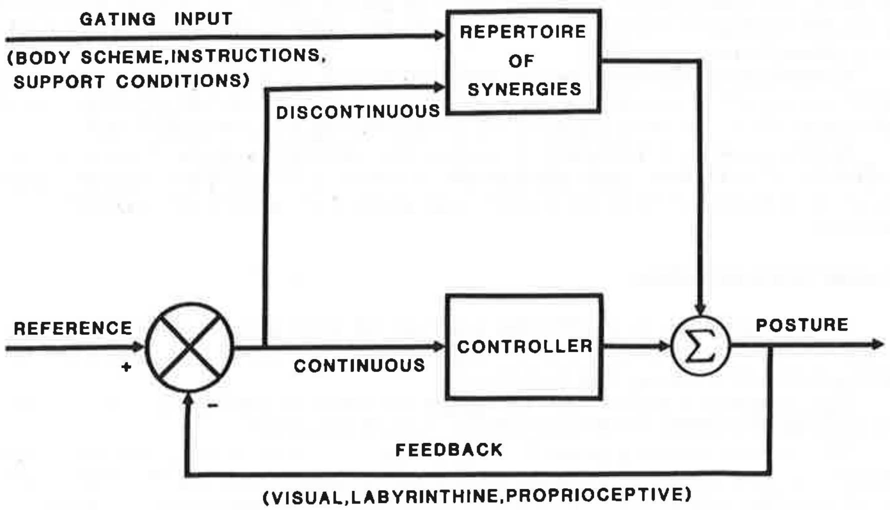

Fig. 1. Control of posture by continuous and discontinuous feedback. Note that three sources of input (visual, labyrinthine, and proprioceptive) contribute to the feedback regulation of whole body posture. The reference value for equilibrium control usually corresponds to the center of gravity position with respect to the ground, but it can change when explicit or implicit instructions are received, as for example when a subject is holding a glass full of liquid. The gated input preselects the appropriate set of synergies as a function of the body scheme, learning, and the support conditions.

2. Error detecting sensors, dealing mainly with labyrinthine, visual and proprioceptive information, indicate any mismatch between the required posture and the actual posture.

3. The postural body scheme, which according to Clement et al (1984) includes a representation of verticality, a representation of the body geometry mainly based on the inputs provided by the spindle afferents, and a representation of the dynamics, especially those of the support conditions.

4. The postural reactions which correct the mismatch between the set reference value and the actual posture. Two modes can be used here : a closed loop mode, which operates in the case of slow disturbances, and a fast open loop correction, which involves a whole repertoire of synergies.

Voluntary movements disrupt posture and equilibrium. The internal muscular forces required to perform a movement affect the body geometry, and hence the CG projection onto the ground. In addition, the internal forces involved in the movement are accompanied by reaction forces at the level of the body segments which execute the movement, and these reactions forces also disturb the body posture. The disturbance can be corrected using either of two modes, a reactive mode, which intervenes only after the disturbance onset, or an anticipatory mode whereby adaptive networks are triggered in a feedforward manner before the disturbance onset (Fig. 2).

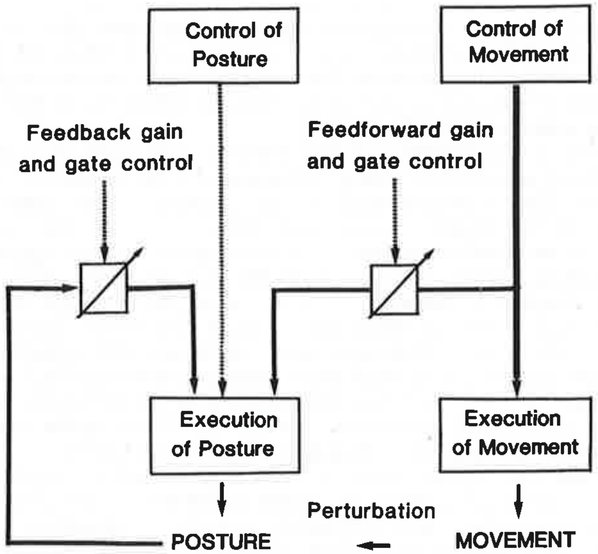

Fig. 2. Feedforward and feedback adjustment of posture. Diagram representing the two mechanisms involved in the compensation for a postural perturbation. In this diagram, the central control of posture is indicated by a striped line. Two phasic mechanisms are minimizing the postural disturbance : they operate through a feedback loop and a feedforward control. The feedforward control is acting through internal collaterals from the movement control pathways on an adptive network involved in postural control. Both mechanisms are under adaptive gate and gain control (from Dufossé et al., 1988).

Most voluntary movements are accompanied by so-called anticipatory postural adjustments, as first described by Belenkiy et al. (1967). Since this pioneer work, many types of anticipatory adjustments have been reported, and it has been assumed that they serve to minimize the postural or equilibrium disturbances resulting from movement performance (see Massion, 1992).

Two main categories of anticipatory postural adjustments do exist :those relating to equilibrium control and those related to the maintenance of the orientation of given body segments. Each of these categories seems to have a different central control. A third category of anticipatory adjustment should be mentioned : those which contribute directly to the movement performance, either by fixing the appropriate joints or by using the body inertia in order to increase the force of the movement.

2. Anticipatory postural adjustments subserving equilibrium control

One example of the anticipatory postural adjustments subserving equilibirum control is the upper trunk movements and the associated hip and knee displacements, which were first described by Babinski (1899) at the turn of the Century. These so-called axial synergies have been reinvestigated by Oddson and Thortenson (1986; 1987 a and b) and by Crenna et al. (1987) Pedotti et al. (1989).

During fast upper trunk movements, associated lower segment displacements can be observed. These associated displacements constitute a highly effective means of regulating the center of gravity position with respect to the ground. This was established by measuring the ground reaction forces, and by calculating the displacement of the CG projection onto the ground, which amounted to around one cm. In the absence of the associated hip and knee movements, it would have been more than 8 cm, which would have resulted in falling. Interestingly, the parameter which is regulated during the task is not the body geometry as suggested by Lacquaniti et al. (1990) in quadrupeds, because the body segments are displaced during the task. It seems to be rather the center of gravity projection onto the ground.

There exists a characteristic EMG pattern associated with fast upper trunk forward or backward movements. Early activation of a set of muscles in the trunk, leg and thigh occurs. The activated muscles are located in the back with backward movements and in the front with forward movements. In the case of backward movements, the set of muscles includes the prime mover, Erector spinae, thigh muscles such as the hamstring and leg muscles such as the gastrocnemius. After the initial burst, activation of the antagonistic muscles can be observed. As the thigh and leg muscles are activated at the same time or even earlier than the prime mover, they seem to contribute to the so-called feedforward anticipatory postural adjustments involved in equilibrium control during upper trunk movements. Ramos and Stark (1990) have been modelling the active and passive forces contributing to the kinematic changes and have observed that the early muscle activation actually contributes to the movement of lower segments in the opposite direction during upper trunk movements.

At this point, the opposite upper trunk and lower segment movements might be said to constitute a kinematic strategy organized under the control of the central nervous system in order to maintain equilibrium during movement performance. In order to test this interpretation, the kinematics of forward and backward upper trunk movements were investigated under microgravity conditions (Massion et al., 1990). Under these conditions, where equilibrium maintenance is no longer necessary, the associated lower segment displacements were expected to be absent. Surprisingly however, the same kinematic changes were observed under microgravity as under normal gravity. The subject was standing with his feet fixed to the floor of the space cabin. First, a change in the initial body position occurred as already described by Clement et al. (1984). The movement of the upper trunk and the hip and knee in opposite directions was still present. With forward movements, the opposite upper trunk and hip movements were the same as under normal gravity; with backward upper trunk movements there was a tendency to increase the knee flexion. These data were interpreted as indicating that a regulation of the center of mass position with respect to the feet occurred regardless of the gravity conditions and that an internal representation of the center of mass probably plays an important role in the organisation of movement.

The kinematic strategy for regulating the center of mass is therefore invariant under both normo-and microgravity conditions : the implementation of the strategy by the appropriate set of muscles does however adapt to the new environmental conditions. It was observed that during backward upper trunk movements, a change in the distal muscle pattern occurred : the early gastrocnemius activation was replaced by an early TA activation contributing to the forward movement of the knee. After return to normal gravity conditions, the previous pattern was not recovered immediately. The early TA activation was still present on the fifth day postflight and was replaced by the gastrocnemius activation observed prior to the flight only on the 8th day postflight. This finding indicates that the EMG pattern underlying the kinematic strategy is flexible. Three main hypotheses have been proposed concerning the muscle synergies involved in postural adjustments (Macpherson, 1991). 1. Fixed synergies which are genetically determined. 2. Flexible synergies which are build-up by training and adaptable to environmental changes. 3. Flexible synergies which are computed on-line depending on the environmental constraints. Our data on muscle synergies under microgravity conditions are in line with the second hypothesis, i.e. that muscle synergies are built up by training.

The last aspect of the anticipatory postural adjustments subserving equilibrium control is their central organization. Very little is known about this point. Some evidence was provided by Babinski (1899) and recently by Viallet et al. (1992) that the cerebellum, and more specifically its anterior lobe, may be involved in controlling this type of coordination.

3. Stabilization of position and orientation of given body segments

The position and/or orientation of given body segments such as head, trunk or limbs are stabilized during movement performance. This is due to the fact that position and orientation are utilized as an egocentric reference frame to calculate the position of the target in space and that of the movement trajectory. Hand reaching movements toward a trajectory involve a series of processes. The target position in space is first estimated in retinal coordinates, and then in head and trunk coordinates. The hand position is also evaluated with respect to the trunk in egocentric coordinates. As a result, the movement trajectory toward the target is calculated and the movement is performed (see Wise et al., 1990). If at any time during the process the head or trunk position are modified, the position of the target and the hand trajectory will be miscalculated and misreaching will occur. As the internal muscular forces driving the movement disturb the head and trunk posture, this should theoretically result in misreaching. The anticipatory postural adjustments stabilize the position and orientation of the segments serving as egocentric reference frame for the movement organization, so that the misreaching is prevented.

One example of feedforward stabilization of the position of particular segments is that of a human subject during a bimanual load lifting task (Hugon et al., 1982; Dufossé et al., 1985; Paulignan et al., 1989). Seated subjects keep their forearm loaded with a 1 kg weight in a horizontal position. When unloading is carried out by the subject himself, the forearm position is maintained although the disturbance caused by the unloading should have displaced the forearm position. This is due to a phasic decrease in the forearm flexor activity which occurs prior to the unloading and can be variably time locked with the activation of the biceps of the "voluntary arm". An anticipatory postural adjustment therefore takes place which very efficiently minimizes the disturbance of the forearm position due to the voluntary unloading movement. This type of co-ordination between posture and movement is an old habit acquired during childhood, which serves to stabilize the forearm position during manipulation of an object by the other hand.

Where are the anticipatory adjustments of posture aimed at stabilizing the orientation or position of body segments organized? Gurfinkel and Elner (1988), observe that a series of patients with an anticipatory adjustment deficit all had a lesion at the level of the premotor and supplementary motor areas. Using the bimanual load lifting task, it was first established that the anticipatory adjustments were still present in a patient will complete callosal section, indicating that the coordination between both sides occurred at a subcortical level. Secondly, in patients with unilateral lesion of the

SMA region, anticipatory adjustment deficits were observed (Viallet et al., 1992). Surprisingly, the deficit was present even when the lesion was contralateral to the only postural forearm. This indicates that the SMA region plays an important role in stabilizing the forearm position in this task, and surprisingly, the leading role in this coordination is played by the SMA contralateral not to the forearm performing the movement but to the forearm maintaining its position. In this bimanual load lifting task, the postural forearm is utilized as an egocentric reference frame for calculating the movement parameters, and higher order motor structures such as the SMA are involved in the gain and gate control of the appropriate learned networks responsible for the anticipatory postural adjustments maintaining the reference position during the movement. During movement execution, collaterals of the movement control pathways might activate the networks already gated on by the SMA area region in order to provide the feedforward postural adjustment of the postural forearm (Fig. 3).

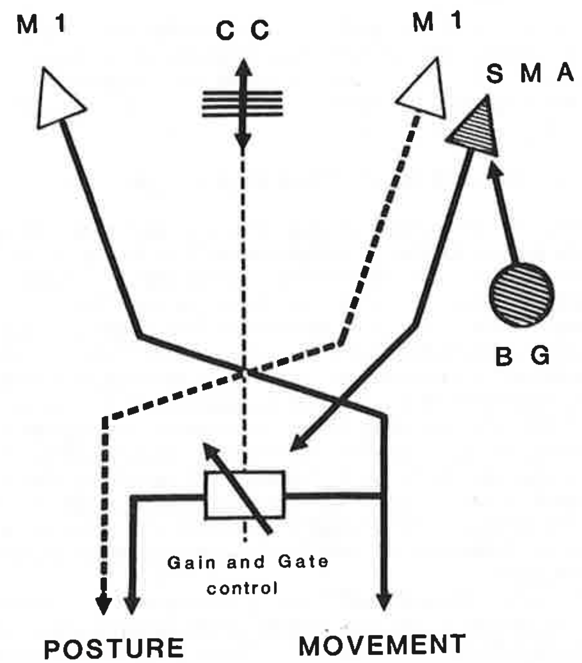

Fig. 3. Scheme proposed to account for the central organization of coordination between posture and movement in this bimanual task. M1 : primary motor cortex. SMA : supplementary motor area region. In this scheme, the motor cortex on one side (M1) controls the load lifting movement (continuous line) whereas the motor cortex on the other side controls the postural maintenance of the postural arm (dashed line). The coordination between the 2 controls is not performed through the corpus callosum (CC). The control pathway for movement sends collaterals at a subcortical level towards the postural arm, which are responsible for the anticipatory postural adjustment. The possibility of using this collateral pathway (gate) and its gain depend on a supraspinal control from the SMA area contralateral to the postural arm and also from the contralateral basal ganglia (BG).

The general conclusion which can be drawn from the above data is that one of the most important parts of the task of the CNS during motor performance is that of controlling the position or orientation of the body segments which serve as an egocentric

reference frame for calculating the movement parameters, and that high order cortical structures such as the SMA or the premotor area are involved in this task.

CONCLUSIONS

During motor performance, movement, posture and equilibrium are co-ordinated.

Two modes of co-ordinated control have been identified. 1. **The feedforward control of equilibrium** serves to regulate the CG (or center of mass) projection onto the ground. Kinematic strategies (movement in opposite directions of the moving segment and other body segments) are used whereby which the CG projection is regulated during movement performance. Muscles synergies, most of which are learned, are responsible for the implementation of the strategies. The cerebellum (anterior lobe) is involved in the control of the kinematic strategies. 2. **The stabilization of position/orientation of body segments serving as egocentric reference frame** depends on anticipatory postural adjustments which minimize the postural disturbances due to the movement. The SMA/premotor area are involved in controlling these reference values.

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CENTRAL PROGRAMMING OF "SIMPLE" VOLUNTARY MOVEMENTS

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Abstract: Discrete movements of a limb about a single joint are a widely studied experimental paradigm for the investigation of voluntary movement. Never-the-less, the literature on this apparently "simple" task does not reveal a consensus on how such movements are controlled. Rather, a diversity of control rules exist, one of which must be selected depending upon specific aspects of the task. For example, movements that vary in distance appear to be controlled according to different rules from those that vary only in speed. Since natural movements as well as experimentally contrived ones usually vary from each other in several "parameters", this appears to complicate the task of any central motor controller which is first, to select a control rule for a movement and then to apply it to the peripheral neuromuscular mechanisms.

In this paper, a computational procedure (program) will be defined that generates control signals for the motoneuron pools of agonist and antagonist muscles. This program can move a single-joint limb from one stationary position to another. The program accounts for moving different distances with different inertial loads and for the influence of different instructions concerning movement speed and accuracy. The model that generates these motor commands describes EMG patterns as well as force and kinematic trajectories which are consistent with much of the data found in the literature of these movements. The model is premised on the notion that kinematically defined tasks are accomplished by programming forces, based on only a few cognitively recognized kinematic and dynamic features of the task. Most of the features found in EMG and kinematic patterns can be considered consequences of the program's algorithmic procedures rather than specifically planned features of those movements.

EMG data will be presented that illustrate the behavior of the motor system performing different kinds of single-joint elbow movements in the horizontal plane. These will be compared to the predictions obtained from the motor program equations solved on a computer. A high degree of correlation will be demonstrated between modeled and actual EMGs under diverse conditions.

Introduction: One of the problems with which the voluntary motor control system deals may be posed as follows. It is asked to perform some kinematic task (such as move from position A to position B) with some dynamic load (such as the limb and its held objects) under some constraint (such as how accurately the final position must be attained and how long the movement may take). This is almost always an ill defined problem lacking a unique solution. This nonuniqueness can have two sources. One may be that the specified task can be exactly performed by many sets of muscle commands because of redundancy in the musculature or the joints. The second is that a specific solution is often not asked for, merely one (of many) that falls within a range of solutions. The nervous system must find a set of muscle commands that realize a specific solution to each such problem in a timely manner.

In spite of the wealth of acceptable trajectories, there is a considerable uniformity to the way even single-joint movements are performed. That suggests the CNS has a systematic way of finding such solutions that resolves the uncertainties of redundancy or vagueness. One example of such uniformities are illustrated by the right two series of movements in Figure 1. In each set, elbow flexions of 72° are performed in different movement times. These movements are "kinematically similar" (Gottlieb, Corcos, & Agarwal, 1992) for reasons that are evident from inspection of joint angle and its derivatives. Note however, that the myoelectrical patterns (EMGs) are not similar, nor are the trajectories of the inertial torque. In contrast, the left and middle series of flexions are kinematically quite different but "kinetically similar," with respect to their patterns of muscle activation and inertial torque trajectories. The right series of movements were performed with differing instructions to the subject regarding movement speed. The middle and left series had the subject instructed to move as quickly and accurately as he could with different weights attached to the limb in the middle series and different target distances used in the left series. It is the aim of this paper to discuss the control principles used to generate the muscle commands that give rise to these uniformities.

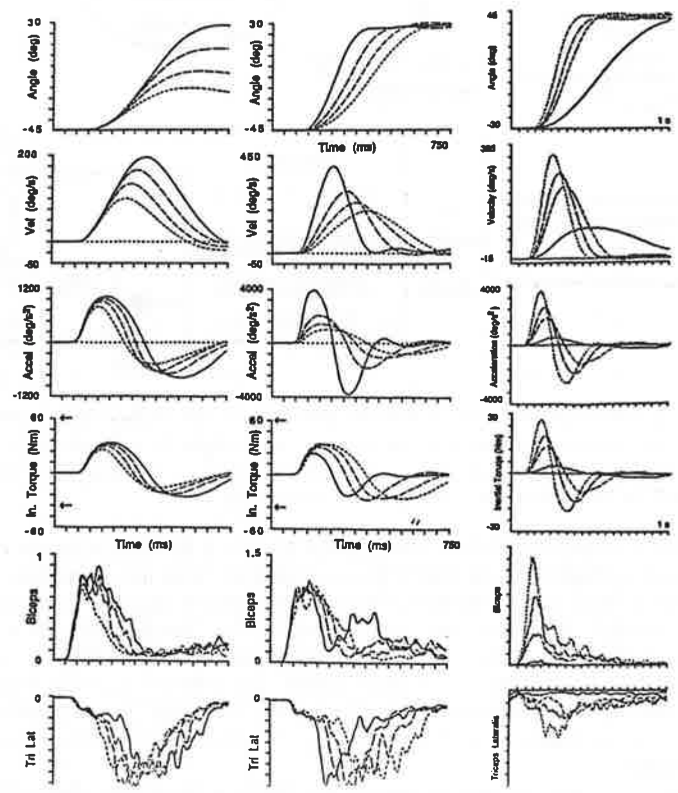

Figure 1: Kinematic and kinetic average trajectories of three series of movements. On the left are flexions of 18°-72°made as fast and accurately as possible. In the middle are movements of 72° with the same instruction as the first series but with four different inertial loads. On the right are shown 72° movements with the same inertial load but four different instructions regarding speed. Reproduced from (Gottlieb, et al., 1992).

Clearly, people can look at a physical scene, receive verbal instructions about a task they should perform, and produce a complex set of coordinated muscle commands that produces these trajectories. The sequence, from task definition to kinematic action can be summarized by figure 2.

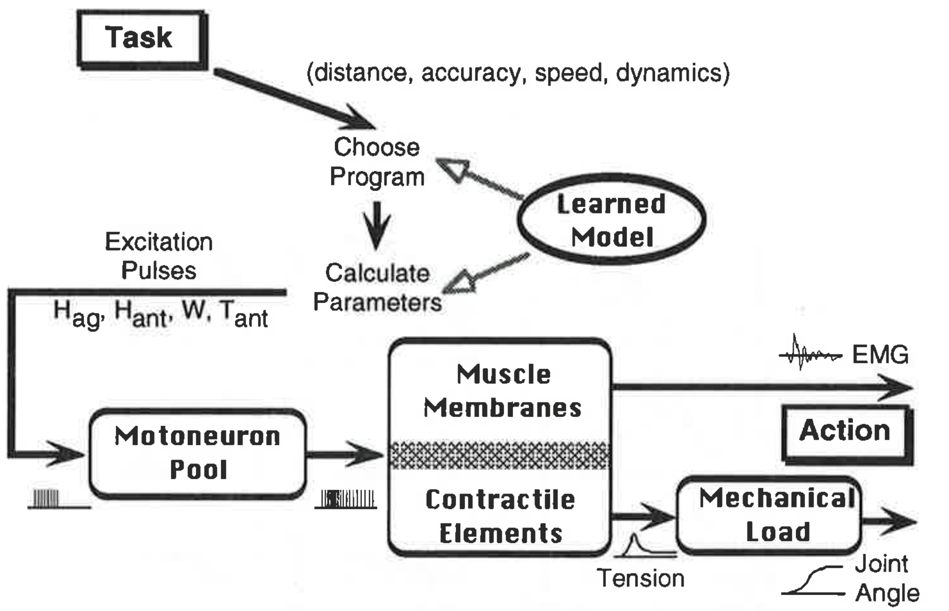

Figure 2 - A schematic sequence that leads from the definition of a task (in the external world) to excitation of motoneurons and muscles. The train of action potentials in the alpha motoneurons produces the observable phenomena, EMG, force and motion, that we characterize as action. This paper describes properties of a "Learned Model" or "Motor Program" contained in the above elipse.

When a subject observes the scene, certain features of the future task may be apparent. The target is clearly an important one since it gives information about movement distance and terminal accuracy among other things. Neither of these are completely specified however. Distance also requires knowledge of present limb location but we can assume that this is available from proprioception. Accuracy can be regarded as a function, almost always nonlinear, that assigns a cost to the deviation of final position from target center. For a button, that cost is probably zero for movements landing on the button and rises sharply everywhere else. For targets with a less tangible cost (such as visual targets on a CRT screen) the cost function is probably smoother and highly subject dependent.

The subject may also see the load, if any, that is to be moved. Regardless of vision, the subject must have some internal model of system dynamics that allows an estimate of the forces that will be needed to perform the movement. Vision can improve the quality of the model but is not a substitute for it. This, it should be noted, is an hypothesis we believe completely at odds with equilibrium-point models of movement which specify a central command that is completely load independent (Feldman, 1986; Bizzi, et. al, 1992). We shall not elaborate on those disagreements here.

Finally, the subject must have some criterion for specifying movement time. Asking for movements "as fast as possible" is not as unambiguous as it seems. The word "possible" carries a lot of hidden baggage. This allows the previously noted constraints of accuracy to intrude as Fitts (1954) demonstrated many years ago. It allows caution to intrude and protect limb and property from discomfort or injury. Beyond this, anyone who has asked for a "maximal effort" of any sort (even

from themselves) knows that to ask and to recieve are not the same thing. In our manipulandum, subjects asked to move 54°as fast as possible can improve their speed by 10-25% with prolonged (100-1500) practice (Gottlieb, Corcos, Jaric, & Agarwal, 1988).

If the subject is not performing a "minimum time with bounded control" movement, what criterion is used? This is a fundamentally important question. All control strategies that optimize some kinematic parameter such as jerk ($\int(d^3x/dt^3)^2dt$) (Hogan, 1984) or snap ($\int(d^4x/dt^4)^2dt$) (Wiegner, & Wierzbicka, 1992) have movement time (MT) as an explicit parameter that must be specified before the optimal solution *for that movement time* can be found. The reason for this is simply that purely kinematic or kinetic criteria have costs that are inversely related to MT. Longer is cheaper and forever is free. That is not a useful guide to present action.

It is also clear that under many conditions, neither movement time nor speed are tightly specified parameters. That is not always the case of course but the exact MT used to push the elevator button is not as important to the satisfactory performance of the task as the MT to raise my hand and catch a ball. Again we can consider the influence of instruction and find that it is less precise than it might seem. Telling subjects to move faster or slower than some previous movement is plainly not precise. Asking the subject to perform the movement in 250 ms is precise but also usually impossible for subjects to achieve. What they can do is change MT and if given feedback about the MT of the last movement, they can change the MT of the next movement to come closer to compliance.

The presumption made here is that neither speed nor time are the fundamental constructs of movement. Something closer to forces is used. In the following section, we will describe an algorithm, that is to say a completely defined computational procedure, that allows the construction of muscle commands from the kinds of information discussed above. Because tasks are not completely defined, this procedure can be considered as the expression of a "convention" that is used to select one specific movement from the large set of possible movements. We can hypothesize that this relatively simple algorithm is implemented by neural circuits of the CNS and constitutes a "motor program." We can also implement it on a digital computer and calculate features of the electromyogram which we can measure.

A Preliminary Task-Action Model of a Simple Motor Program We postulate four obligatory inputs to our model. The first is movement distance as described above. We assume that this is straight line distance. The second is terminal accuracy. For computational purposes, we will treat this as a simple number that characterizes equivalent target size (T_e). Larger numbers imply lower accuracy requirements. We will treat load dynamics as inertial, characterized by a moment of inertia, and a viscosity. Finally we will control movement speed by declaring a control parameter we will call "urgency." This number expresses in relative terms (from 0 to 1) how fast we wish to move. We will further restrict ourselves to relatively "fast" movements for which we know that the CNS generates brief, pulse-like contractions of agonist and antagonist muscles to accelerate the limb towards and arrest it near a desired target location.

These four input parameters are all specified by some combination of observational information from the environment (how far, how heavy, how accurate), instructional information from other individuals (how fast, how accurate), learned information about the physical behavior of the world (our intuitive understanding of Newtonian mechanics) and "arbitrary" decisions we may make to interpret this information and fill in any missing pieces. From these four inputs, two intermediate variables, extent and effort, are calculated according to equations 1 and 2.

The assumption that these movements will be controlled by force pulses allows us to generate rectangular control signals we call excitation pulses. These are signals from higher centers of the CNS that excite and contribute to the activation of the motoneuron pools. Five parameters are needed to specify two rectangular pulses for the agonist the and antagonist muscles respectively. These are the two widths and heights and the relative time between their onsets. We calculate them with equations 3-6. Pulse width (*W*) is specified only by extent. For lack of data sufficient to better verify the model, we will assume that the widths of the two pulses are identical so there are only four equations. Pulse heights depend linearly on effort while extent has a nonlinear effect. This nonlinear function makes the heights proportional to extent for "small" movements and independent of it for "large" ones. These four equations specify two excitation pulses illustrated in figure 3. Note that the equations specifying pulse height are not the same for the two muscles. The exponential factor added in equation 5 is the way we have implicitly incorporated the effects of muscle viscosity (ie force-velocity properties (Hill, 1938; Lestienne, 1979). The rationale for this formulation is that viscous forces, which assist in slowing movements at the target position, will have a greater net effect over longer distances. The justification for this requires a kinematic analysis that we cannot pursue here.

$$X = k_{DX}D + k_{JX}J \quad (1)$$

$$F = k_f UT_e \quad (2)$$

X is EXTENT — D is perceived target distance
F is EFFORT — J is estimated limb plus load inertia
U is urgency
T_e is effective target size
k_{DX}, k_{JX}, and k_f are constants

$$W = W_0 + c_0 X \quad (3)$$

$$H_{ag} = c_{ag}F\cdot\left(1-e^{-X/X_0}\right) \quad (4)$$

$$H_{ant} = c_{ant}F\cdot\left(1-e^{-X/X_0}\right)\cdot e^{-D/D_0} \quad (5)$$

$$T_{ant} = T_0 + c_T\frac{X}{F} \quad (6)$$

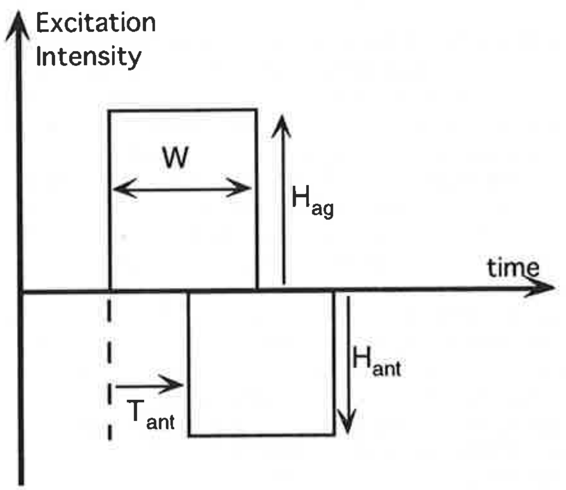

Figure 3: Excitation Pulses of agonist (upward) and antagonist (downward) are drawn to illustrate how the parameters completely specify them. All measurable signals such as EMG, torque and displacements are low-pass filtered consequences of these pulses.

Although equations 3-6 are the complete solution as far as the CNS is concerned, they describe completely unobservable quantities which makes them incapable of direct verification. Their is a direct relationship between the excitation pulses and both the EMG bursts and muscle contractile forces as can be inferred from figure 2. These relations have the form of low-pass filters

as we have discussed elsewhere (Gottlieb, Corcos, & Agarwal, 1989a, b). One way to test the validity of these equations is to examine their implications for the EMG bursts associated with the agonist and antagonist muscles. It is clear without extensive analysis that if the EMG signal is rectified and integrated, the areas of the EMG bursts should be proportional to the areas of the excitation pulses. The agonist and antagonist burst areas (Q_{ag} and Q_{ant}) are given by equations 7 and 8. It is also clear that if one examines the area for an interval of time shorter than W_0, then only the height of the pulse will be a factor. Equation 9 gives this for the first 30 ms of the agonist burst (Q_{30}). The only feature of the excitation pulses that is directly measurable is the latency of the antagonist burst relative to the agonist's onset (T_{ant}). This was already specified by equation 6.

$$Q_{ag} \propto WH_{ag} \propto (W_0 + c_0X)\cdot UT_e\left(1-e^{-X/X_0}\right) \quad (7)$$

$$Q_{ant} \propto (W_0 + c_0X)\cdot UT_e\left(1-e^{-X/X_0}\right)e^{-D/D_0} \quad (8)$$

$$Q_{30} \propto H_{ag} \propto k_f UT_e\left(1-e^{-X/X_0}\right) \quad (9)$$

In the following section, we will descibe the experimental validation of this model by electromyographic measurements made on normal human elbow flexions.

Experimental Model Validation A subject was asked to make elbow flexions as quickly and accurately as possible. The movements were made with a lightweight manipulandum in the horizontal plane (Gottlieb, et al., 1989a,b). The 6° wide target was located at one of seven positions ranging from 4.5° to 56° from the starting point (60° flexed from a straight extended arm). A digital computer recorded joint angle, angular acceleration and surface EMGs from biceps and lateral triceps muscles. The subject made seven sets of 15 movements at each distance with the distance changing after each set. The onsets of the agonist and antagonist bursts were identified visually on a computer monitor and the areas under them were calculated by the computer. The experiment was performed on eight subjects. Two of them demonstrated patterns of behavior that are not consistent with our model. We present results from one subject here that are highly representative of the other six. A complete description will be found in (Chen, 1982).

Figure 4 shows the average values of the parameters that were measured from the agonist and antagonist EMG records at each movement distance (open circles). The smooth curves are equations 6-9 with coefficients chosen by trial and error.

Discussion The model is contained in equations 1-6. These calculate four model parameters from four externally defined variables, two of which (U and T_e) are only qualitatively specified. Hence, one criticism of this model is that it is an arbitrary exercise in "curve fitting." Certainly there are other equations that might fit the data as well or better. It is worth considering therefore, what the constraints on the choices of these equations are.

The first point is that some of the features of these equations are not due to either physical or physiological constraints. For example, equations 1's kinetic equivalence of distance and inertia is based upon experimental observation (cf figure 1), not on any theoretical (from physics or physiology) principle. The constant term in equation 3 is also an experimentally based term and one of some consequence. Since W represents the duration of converging trains of action potentials, there is no obvious reason why it cannot fall smoothly to a minimum of 2-5 ms, the duration of a single action potential. The reason it does not is probably related to the fact that the twitch contraction of a single action potential lasts 50-100 ms (Burke, & Edgerton, 1975) so that muscle force cannot be smoothly brought to zero by modulation the duration of a train. It can be brought more smoothly to zero by varying the number of motor units recruited which corresponds to intensity modulation. The failure of force to go to zero as W goes to zero requires that H be the control parameter for small forces. This is consistent both with the data here and those of Hoffman & Strick, (1986; 1990) on wrist movements which had seemed to contradict our findings at the elbow. Their experiments were performed for small movements in which the nonlinear terms of equations 4-5 dominate. Ours were for larger movements for which those nonlinear terms became constant. The model describes both.

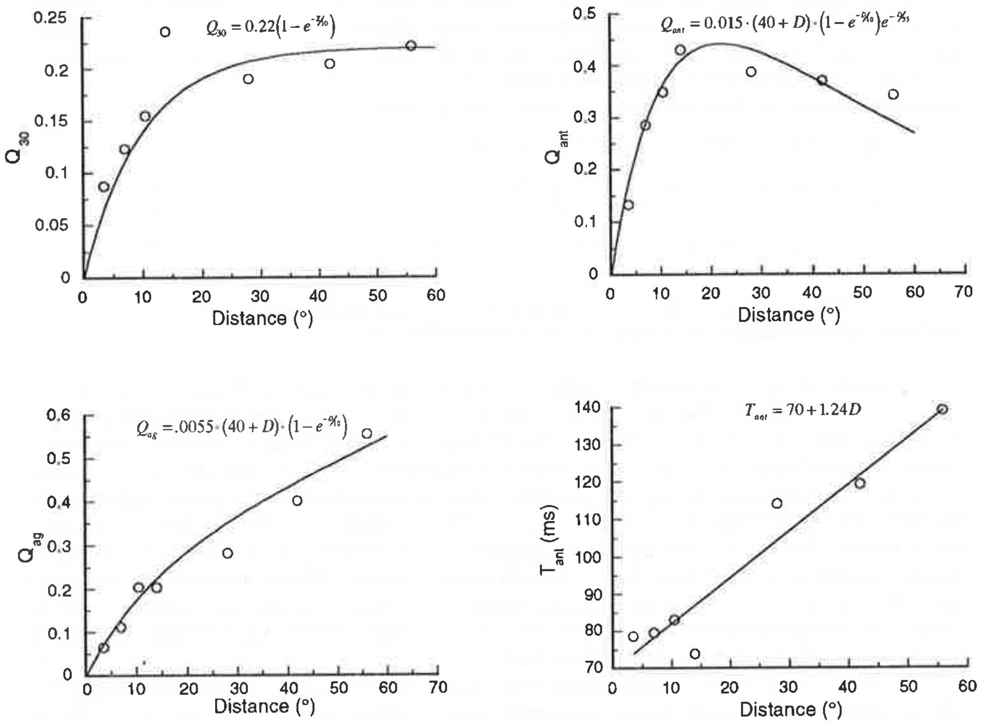

Figure 4. The rectified EMG was integrated over the agonist (Q_{ag}) and antagonist (Q_{ant}) bursts and over the first thirty ms of the agonist burst (Q_{30}). The onset time of the antagonist burst was determined by visual detection of the sharp rise in activity. All four variables are plotted as a function of movement distance. In each part of the figure, the smooth curve is a graph of the model equations shown.

The addition of the second exponential term to equation 5 is to accommodate the force-velocity properties of muscle. These properties provide a viscous-like behavior that is inescapable and allow reduced antagonist activation for longer distance movements while requiring stronger agonist activation (Gottlieb, Latash, Corcos, Liubinskas, & Agarwal, 1992). The model provides a single, uniform set of rules for a wide variety of single joint tasks and from these rules, both the Speed-Insensitive and Speed-Sensitive strategies that we have described previously (Gottlieb, Corcos & Agarwal, 1989a) are emergent properties.

The approach taken here is an alternative to control models based upon optimization schemes (eg Hogan,,1984 or Wiegner & Wierzbicka, 1992) and a supplement to models based upon equilibrium point notions (eg Feldman, 1986; Bizzi, et al. 1992). This model is good for launching movements but is very poor for accurately positioning the limb. Terminal accuracy is exquisitely sensitive to pulse parameters which, from observations of EMG signals, are highly variable. Accurate position control almost certainly exploits joint elastic properties as elaborated by the equilibrium point hypotheses. Those models do not easily account for controlling movement speed. They are compatible with the model here and each can account for different parts of the overall task..

The conclusions we can draw from this model is that for single-joint flexions at least, a simple set of rules can be used to compute muscle activation parameters from externally specified task parameters. This model incorporates a number of arbitrary features which represent *a priori* choices made by the nervous system to solve a potential problem of redundancy. This is similar to Bernstein's notion of synergy (Bernstein, 1967) at the level of a single joint. The parameters of this model can easily be learned by trial and error (Stein, Cody, & Capaday, 1988) to become part of our standard movement repertoire. When we are uncertain of some of the task parameters (for example we do not know how heavy the object to be moved is), we will either slow our movement to allow feedback to help us or we will risk error. Practice and repetition are associated with all skilled tasks so that the parameters of our internal control model can be calibrated.

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CONTEMPORARY CLINICAL EMG APPLICATIONS

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Historically clinical electroymyographers have emphasized the role of interpreting percutaneously recorded electromyographic signals to differentially diagnose diseases of myogenic or neurogenic origin or as a basis for determining the speed at which peripheral nerve elements conduct impulses. Other electromyographic signals have been evoked in response to cutaneous stimuli. For example, we can examine changes in levels of spinal cord motoneural pool excitability through recording H reflex responses to a variety of stimuli.

The development of sophisticated and timely microprocessing capabilities has permitted the accumulation, storage and retrieval of copious amounts of surface generated EMG activity within the context of kinesiological or neurophysiological experiments. Indeed, in a similar talk to this group in 1990, Rosenfalck [1] highlighted the emerging role of knowledge-based computer systems as active participants in the clinical decision making process, thus helping to pave the way for our collective thinking about clinical EMG in a far more dynamic manner. These contemporary additions to our clinics or laboratories have allowed us to refine our interpretations of how muscles work under a variety of circumstances. Moreover, quantification of this activity can be related to other important procedures, such as motion analyses and computer simulated movements. At the 1990 ISEK meeting, attention was given to the role of EMG as a vehicle for *both* treatment *and* quantification [2]. In fact the exorbitant growth in health care costs demands that our processing of the clinical EMG signal serve the dual roles of providing efficacious intervention and numerically documenting change; and, whenever possible, these roles occur simultaneously.

Among the marvelous functions played by our very young but popular publication, *Journal of Electromyography and Kinesiology*, is that of sounding board for new ideas that echo the emerging role of contemporary clinical EMG applications. For several years we have referred to the dynamic process of quantifying EMG data for the purpose of creating or altering therapeutic interventions, *applied kinesiology* [3]. Our journal has shown in magnificent fashion just how this powerful marriage can be successfully undertaken. For example the work in Sinkjaer and Arendt-Nielsen [4] first examined the sequence of activation and magnitude of response from muscles about the knee in normal subjects and patients with various degrees of joint laxity following anterior cruciate ligament repair. EMG evaluations of medial gastrocnemius revealed poor timing and activation during a normalized gait cycle. Feedback of this muscle's activity during simulated gait so that it could contribute to enhanced stiffness about the knee at heel strike and improved joint integrity.

Another example of applied kinsesiological investigation using EMG permits an exploration and testing of time honored neurophysiological principles as a basis for specific therapeutic interventions for neurological patients. Most neuromuscular retraining for stroke or head-injured patients presumes that hyperactive muscles must first be "inhibited" before a weak, antagonist muscle is recruited and that such a process should precede in a proximal to distal manner. Other investigators [5] have argued that simply teaching function irrespective of such principles is therapeutically prudent and just as effective. To test this notion Wolf and colleagues [6] have compared functional training by recruiting the triceps brachii with muscle feedback to elbow movement training without the benefit of visual information about triceps activity in two sets of eight chronic stroke patients. At no time did subjects receive any information about activity in the more active biceps. Within group comparisons showed significant increases in triceps EMG output during more functional measures from the feedback group and more cocontraction in the non-feedback group. Yet active and passive range of elbow movement improved significantly in both groups, thus revealing the possibility that functional changes may be due to biomechanical rather than neuromuscular changes about the joint. In principle applied kinesiological explorations contributed to the notion that subjects could be trained to increase movement without first specifically inhibiting activity in the spastic muscle.

Recently the architecture of multi-joint human muscles has been re-examined and evidence provided to suggest an anatomical basis for compartments [7]. Furthermore we presented preliminary data at the last ISEK meeting indicating that there may be quantitative differences in intramuscle activity across compartments during task specific behaviors [8]. More recently we have analyzed motor unit recruitment thresholds within compartments of human lateral gastrocnemius as a first approximation of physiological composition of muscle within compartments. These data indicated that motor units within the C head of lateral gastrocnemius may be more selectively active at higher forces than motor units from other insertion sites during ankle plantarflexion. Conversely motor units residing in B head seemed to require lower activation thresholds. If these data are supported through more rigorous study then one may imply that preferential motor unit types may either reside within individual compartments, or, more likely, are called upon based upon the context of the motor plan. If either possibility holds merit, then the implications for interpreting EMG data are profound. Clinicians and researchers may need to rethink such issues as electrode placements, interpretation of (normalized) EMG data, electrical stimulation parameters including placement and rate and duration of stimulation or even strategies to recruit components of muscles within functional retraining. Clearly there exist many new paths worthy of kinesiological EMG exploration.

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CLINICAL APPLICATIONS OF GAIT ANALYSIS

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Quantitative biomechanical gait analysis offers objective and quantifiable information that is potentially an extremely useful adjunct to routine treatment planning for a wide range of pathologies causing ambulatory dysfunction. Objective and quantitative evaluation are extremely relevant to today's current environment where there is a need for more objective outcome measures of treatment yet, gait analysis has been widely accepted as a clinical test and the value of the information generated during quantitative gait analysis is often questioned for its value, uniqueness and cost effectiveness. The purpose of this presentation is to examine some specific applications of gait analysis to various orthopedic treatment modalities.

The clinical relevance of gait testing will be examined with respect to certain criteria. These criteria include the following:

1) ***Uniqueness of Information***: The information generated from the gait tests must be unique and quantify measures that cannot be obtained through simpler, less expensive clinical examination.

2) ***The Potential to Improve Treatment***: The information generated from the gait analysis must have the potential to improve treatment.

3) ***Cost to Benefit Ratio***: It is important to demonstrate that information generated from gait analysis can increase the probability of improved treatment outcome and that the cost of gait analysis is warranted based on the potential benefit.

Specific examples of gait analysis applications will be used to illustrate some fundamental principles related to the use of gait analysis as a clinical tool.

Several examples of the clinical applications of gait analysis will be described. These include the use of gait analysis for the treatment planning of patients who are candidates for high tibial osteotomy for varus gonarthrosis of the knee joint. In this application, it will be demonstrated that dynamic loading at the knee joint (adduction moment) is a significant predictor of which patients will do well with a high tibial osteotomy. The application of gait analysis to the study of patients with anterior cruciate ligament (ACL) deficiency. Specifically, it will be shown that the majority of patients with anterior cruciate deficient knees walk with abnormal gait indicative of a tendency to reduce or avoid a net moment at the knee tending to require quadriceps muscle force. This population of ACL-deficient knees was also tested during more stressful activities such as running, running and cutting and running to a stop.

QUANTITATIVE ASSESSMENT OF GAIT DISTURBANCES IN NEUROLOGICAL PRACTICE: THE PATHOPHYSIOLOGICAL APPROACH

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Recently available technologies allow simultaneous monitoring of multiple gait parameters in freely-behaving, unconstrained subjecs. In order to apply the above methods for the analysis of gait disorders in the clinical practice, different strategies can be adopted. The most common approach involves collection of kinematic, dynamic and EMG recordings, and provides a quantitative description of the abnormal pattern of walking in terms of joint angles, joint torques and EMG phasing. Such an approach can be used for monitoring the evolution of the functional picture and for objective assessment of the effectiveness of treatments. Yet, it fails to provide easily interpretable insight to the mechanisms which possibly underly the disturbed motor control. We feel that an alternative strategy, primarily aimed at the extraction of pathophysiological information on relevant mechanisms responsible for the abnormal gait, should probably be more suitable for clinical purposes.

A pathophysiologically-oriented analysis of gait is of particular interest in the case of patients with neurological disorders. This is because motor impairments associated with neural damage are often the result of multiple mechanisms, both central and peripheral, which concur in varying proportions and may possibly change their relative contribution over time.

Despite previous studies concerning the pathophysiology of gait in neurological patients (e.g. Knutsson, 1979, Berger et al. 1982, Conrad et al. 1985, Crenna & Frigo 1985, see also Winter 1985) to our knowledge, no attempts have been made to dissect out the contribution of different factors, symultaneously involved in the disturbed locomotion of individual patients. In our gait laboratory, we recently developed a new approach aimed at addressing the above issue. The method (Crenna et al. 1992a) consists of a set of analytic routines, each devoted to the detection and quantization of a single pathophysiological factor which could potentially hinder locomotion in neurological disorders. To date we have identified four such factors: 1) defective recruitment of motor units (Paresis); 2) abnormal velocity-dependent EMG recruitment during muscle stretch (Spasticity) 3) non-selective activation of antagonistic muscles with loss of the normal reciprocal inhibitory pattern (Co-contraction); 4) changes in the mechanical properties of the muscle-tendon system (Non-neural Component).

The interference of the above factors into the locomotor function is tested across various rhythmic movements, covering different velocities and levels of difficulty (overground walking at natural speed, walking at maximal speed, stepping on the spot and alternate hopping on the spot). Investigation of locomotor tasks with various degrees of automatization and difficulty, which appear to require different settings of the segmental apparatus (e.g. different gating of peripheral afferents (Llawellyn et al. 1990)), is expected to test locomotor function over a wider range of central control situations. Moreover, the use of different movement speeds will produce different stretching velocities of the examined muscles, which in turn are expected to uncover abnormal velocity-dependent EMG responses.

For each set of rhythmic movements, multifactorial analysis (kinematics, dynamics and EMGs) is carried out. Three-dimensional lower limb kinematics are recorded simultaneously from both sides with a motion analyzer (ELITE) equipped with four cameras. Dynamic variables (ground reactions and joint torques) are measured by means of a force plate (Kistler) hidden in the floor, and EMG signals are collected by surface preamplified electrodes from pairs of antagonistic muscles of thigh and leg, bilaterally.

Efforts were made to process a minimal number of variables, relevant for the detection of the pathophysiological factors as defined above. At the same time, however, criteria for the assessment of the single factors have been established, requiring the convergence of kinematic, dynamic and EMG conditions. This "redundancy" is espected to increase the reliability of the information extracted. The main criteria applied to congenital hemiparesis, the pathology for which the method is currently being tested are summarized below.

The incidence of the Paretic Component is estimated from combined EMG and joint angle behavior. Criteria for detection during gait include significant reduction of the EMG output (50% or less) as compared to the homologous muscle on the contralateral non affected side, in the absence of activity on the antagonist. Congruent kinematic correlate must also be present in terms of abnormal excursion of the relevant joint, and items must be true for all the motor tasks examined.

Spastic Component is assessed on the basis of EMG and kinematic parameters. Criteria include synchronous EMG recruitment during stretching of the relevant muscle, and abnormal, positive correlation between EMG amplitude and velocity of muscle stretching (Crenna et al. 1992b). Lengthening velocity is computed as the first derivative of muscle length, the latter being obtained from joint angles by means of geometrical muscle models (Frigo and Pedotti, 1978).

One criterion for the detection of Co-contraction is a significant increase in the overlapping between EMGs of two antagonistic muscles. The degree of overlapping is quantified both temporally, as the percentage of the step cycle during which the antagonistic muscles are coactivated above a given threshold, and geometrically, as the area of rectified- integrated EMG over the same period, normalized to the mean locomotor activity of the two muscles (Rodano et al. 1992). Careful exclusion of artifactual co-contraction due to crosstalk is obviously a prerequisite for the detection of such a component. Decreased joit excursion concurrent with the co-activation phase indicates a mechanical effect in terms of joint stiffening.

Assessment of the Non-Neural Component necessitates a reduced joint excursion during passive mobilization (quantified by goniometry) in the absence of detectable articular pathologies. Criteria for the assessment during movement include: reduced excursion of the relevant joint angle; stepwise increase of torque during lengthening of the muscle-tendon considered; abnormal relationship between joint torque and joint angle , and absence of correlation between changes in the EMG output and the above biomechanical phenomena on the muscle examined.

For an easier clinical application, data processing will eventually produce a readable histogram-like presentation, the "Pathophysiological Profile", containing the relevant information, i.e. the relative contribution of the four pathophysiological factors, scaled in discrete levels. In our pilot study two levels of impairment are considered. For the Spastic Component, for instance, level one exists when criteria for detection are true only for maximal walking speed (and hopping), when the highest velocities of muscle stretching are espected to occur. Level two exists when the same items are true also for natural walking speed (and stepping), indicating a lower threshold for pathological muscle responses to stretch.

Preliminary analysis of Pathophysiological Profiles obtained in children with congenital hemiparesis and confined to distal muscles (pretibials and triceps surae) (Crenna et al. 1992a), revealed that, as espected, several factors can be present simultaneously in the same patient and each patient has gait with a distinct Profile. This indicates there is consistent inter-subject variability, even within the same clinical group. The Profile could be dominated by the Spastic, Co-contraction, or Non-Neural Component. Rather unexpectedly, the Paretic Component was rarely prominent in hemiplegic children, and most often attained its lowest scaling level.

Analysis of the Spastic Component revealed a marked increase in the amplitude and degree of synchronization of the EMG activity recorded during the stretching phase which takes place in triceps surae after ground contact, but no EMG recruitment during the lengthening (most often much faster) occuring in mid- swing. The abnormal velocity-dependence of the early-stance EMG burst is confirmed by the high slope and significance of the correlation between lenghtening velocity and EMG size. The same phenomenon could be observed in diplegic children, and adults with neural lesions resulting in a "spastic" gait (e.g. familial spastic paraparesis, cervical myelopathy). In part of these patients, deeper analysis of the transmission along the stretch reflex circuit was carried out, using a computerized method developed in our laboratory for the study of the H reflex in calf muscles during rhythmic lower limb movements (Crenna & Frigo, 1987). Results indicate that higher than normal excitability levels can be present in the stride phases characterized by pathological velocity-dependent EMG recruitment (Crenna & Frigo, 1991).

Increased Co-contraction was frequently observed in hemiparetic children on the affected side, but could also be revealed on the contralateral "healthy" side, especially at the highest walking speeds. This suggests that Co-contraction is not necessarely derived from a loss of selectivity in motoneuron recruitment, as a result of decreased reciprocal inhibition and/or disturbed centrally-generated locomotor program. Alternatively, this could be a protective strategy aimed at stiffening a certain joint, to compensate for postural instability, paresis, or reduced availability of periperal information. Analysis of rhythmic tasks requiring different postural or equilibrium demand (e.g. free walking vs hopping, or supported walking) could help distinguish the two conditions (see Richards et al. 1991).

In patients with cerebral palsy (CP) analysis of the degree of Co-contraction, together with phasing of the activity of representative muscles within the step cycle, provided also indirect information concerning disturbed organization of the locomotor program. In fact, features reminiscent of those described for the immature digitigrade gait could be observed in hemiparetic children (e.g. pre-contact EMG activation and absence of forward thrust burst on triceps surae; forefoot or footflat ground contact; ankle movement in phase with the knee movement; absence of typical push off peak in the vertical component of the ground reaction (Okamoto & Kamamoto, 1972, Forssberg, 1985)). Features of the immature digitigrade gait and of the early unsupported locomotion have been reported in CP patients by Leonard et al. (1991), who documented their persistence after the puberal age. These observations suggest the possible introduction of an additional pathophysiological component, i.e. the primitive, immature one.

When testing a new protocol, consistency of findings over repeated trials is obviously a major requirement. In this connection, we found that when the full recording session was performed twice within a ten day interval, the Pathophysiological Profile of a patient's gait was consistently reproducible. After longer periods, however, (one year) dramatic changes in single components (e.g. Non Neural Component) could be detected in non-stabilized patients, suggesting the possible fluctuation in the effectiveness of single mechanisms.

In conclusion, it is felt that dissecting out different neural and non-neural pathophysiological factors, quantifying their interference into the actual locomotor function and displaying this information in a format easily accessible to the clinical operators, might afford an additional tool for closer understanding of the hindering mechanisms, more rational planning of patient-tailored, therapeutic and rehabilitation procedures, and, possibly, for prediction about functional evolution.

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CLINICAL PERSPECTIVES OF THE EUROPEAN CALIES PROJECT

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To restore locomotion in paraplegic patients, FES appears to be the most realistic method. All tentatives made to improve local neural regeneration in spinal cord, as well as the implantation of foetal nerual cells, are still at research level. Therefore, stimulating different muscles for a coordinated task such as walking requires a lot of solutions to very complex neurophysiological problems. The use of a computer to make a substitution of motor brain control mechanism justifies the term of CAL (Computer-Aided Locomotion) and the implantation, within the body, of an electronic box delivering the appropriate sequence of muscle stimulation by the way of epimysial or neural electrodes is the most promising clinical way to use FES for a long period of time. Many problems have to be solved such as muscle fatigue, long term tolerance of electrodes or closed-loop system with goniometric and ground reaction force feedback. In some cases, orthotic braces have to be used in a hybrid solution. Nevertheless, we expect to minimize need for external equipment in order to make the fitting procedure easier and simpler. A European clinical network, grouping different rehabilitation centers in Europe is in the process of being installed. It will be the only way to guarantee the use of the same protocols before and after surgical implantation of the CALIES implant. The implantation according to the skin problems commonly observed in paraplegic patients has to be as less invasive as possible. An endoscopic surgical procedure to put the electrodes in place on muscles or nerves in quiet mechanical areas will be presented. The first implantation is expected to be carried out at the end of 1993. Only paraplegic patients injured at level between T4 and T11 could benefit from the CALIES program. Later on we expect to enlarge the clinical indications to some tetrapelgic patients with similar problems to reactivate sublesional muscles and to some hemiplegic patients, which represents however a completely different problem.

NEW TECHNOLOGIES FOR FUNCTIONAL NEUROMUSCULAR STIMULATION

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It is rapidly becoming possible to reanimate a paralyzed limb by applying well-controlled electrical stimulation to multiple nerves and muscles. However, fitting the technology to a particular patient with a particular disability now requires a large number of measurements, observations and decisions that are beyond the capabilities of most clinical rehabilitation environments. Fortunately, there are parallel technological advances in four key supporting areas - noninvasive morphometry, motion analysis, graphical user interfaces, and automated kinetic analysis - that may make this problem tractable, but only if they can be harnessed into complete, user-friendly clinical systems.

1 Introduction

Humans easily perform sensorimotor tasks that are far beyond the best robots, despite their reliance on slow, noisy, nonlinear actuators called muscles. The command and feedback signals and control schemes used by their nervous systems are only vaguely understood. Nevertheless, research aimed at restoring purposeful movement to the limbs of paralyzed patients by functional neuromuscular stimulation (FNS) is widespread and growing [for reviews, see 1, 2].

Much of the technological effort to date has concentrated on the interface between electronic devices and neuromuscular tissue, particularly the ability to apply well-controlled stimuli to multiple individual muscles in a limb without requiring extensive surgery or relying on mechanically vulnerable leads and connectors. One approach to this problem is the use of modular "microstimulators" that are small enough to be injected into individual muscles and which receive power and digitally-encoded command signals by RF telemetry from an externally worn coil [3]. While this interface is obviously the sine qua non of FNS, there is also growing appreciation of the importance of sensory feedback. In normal limbs, the signal traffic related to the control and responses of proprioceptive and cutaneous sensors is many times greater than that related to activation of extrafusal muscle fibers, even considering only the low threshold mechanoreceptors that are directly related to voluntary sensorimotor control. Fortunately, sensor technology is also developing rapidly and outward transmission of data by RF telemetry also appears to be feasible [4].

Once the inward and outward interfaces become reasonably sophisticated and readily available, the limiting factor in the clinical application of FNS will be the ability to design and fit a prosthetic system for the functional tasks of a patient with a particular pattern of sensorimotor impairment. The magnitude and importance of this task can, perhaps, be appreciated by a consideration of recent developments in cochlear prosthetics, the most successful application of multichannel functional electrical stimulation to date. Clinical success has been related closely to the ability to match the operating characteristics of the acoustic signal processor and stimulus controller to the level of sensorineural and cognitive function available in the individual patient [5]. This has required manufacturers to develop whole batteries of semi-automated psychoacoustic test materials and computerized systems whereby an audiologist can make use of expert knowledge to design and program the patient's wearable speech processor. The comparable requirements in FNS are even more daunting because of the much greater range of functional pathology, the plastic changes induced by both prolonged inactivity and responses to FNS, and the cumbersome technology required to quantify motor performance.

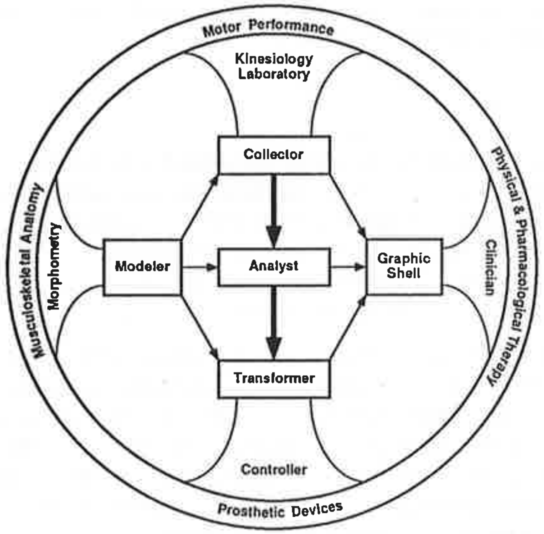

Fig. 1: Relationships between patient (outer circle), clinical interfaces (inside circle) and software modules in the computerized fitting system (square boxes).

Figure 1 shows a schematic representation of the problem. At the perimeter are the four clinical aspects that deal most directly with the patient:

1. *the patient's musculoskeletal anatomy, including estimated mechanical dynamics*
2. *the patient's current level of motor performance in daily living tasks*
3. *physical and pharmacological therapy available to modify the patient's condition*
4. *orthotic and neuroprosthetic devices available to assist the patient's function*

At the center is a structure that divides the computational requirements of a clinical diagnostic and fitting system into a set of interrelated modules. Between the computational system and the four clinical aspects are four corresponding types of technology:

1. *noninvasive morphometric methods such as magnetic resonance imaging (MRI)*
2. *motion analysis systems, including EMG, dynamometry and other kinesiological methods*
3. *visualization tools and interactive interfaces that provide graphical access to and analysis of complex data sets*
4. *programmable, wearable controllers for dynamic assistive devices such as FNS and active braces*

In the following sections, we highlight some of the current and planned work in the Canadian Network of Centers of Excellence that is directed toward the development of complete clinical systems for the functional recovery of patients with sensorimotor disabilities.

2 Computational Structure and Bookkeeping

The core of Figure 1 shows a set of interconnected modules for dealing with types and levels of data analysis. At the top is the data *Collection* step, which involves both hardware and software used to acquire visual images, marker locations, transducer and bioelectrical signals both in real-time and from archival media such as tape and film (see 3 below). The various data acquisition platforms need to be under the master control of a single platform that can keep track of the various types and formats of data as well as many different trials of different tasks collected during an extended course of rehabilitation.

In the *Analysis* step, the corresponding sections of the various data files must be brought together to produce multichannel records vs. time. Calibration factors must be preserved and used to produce intelligently scaled displays; specific types of records may require interpolation and/or filtering. The records must be easily scanned and edited to identify valid sequences and broken into repetitive cycles for averaging or comparison of different therapies.

In the *Transformation* step, the various types of observable data are used as inputs to functions that infer unobservable state variables according to biomechanical models.

For example, the trajectory of limb positions can be used to infer muscle lengths and velocities via models of gross musculoskeletal anatomy; these combined with EMG can be used to infer muscle force via models of muscle dynamics.

All of the above three steps can be greatly facilitated by access to a common *Model* of the musculoskeletal system of the patient. This model can be used to guide the selection of data in the *Collection* phase and to assure that the data files have proper identification and calibration. During *Analysis*, the *Model* data can be used by *Graphical* tools to create iconographic displays that relate kinesiological data to musculoskeletal anatomy in a way that is meaningful to clinicians (see below). Perhaps most importantly, it provides the basis for organizing the structure and the parametric data for the models employed in the *Transformation* process.

3 Integration of Kinesiological Data Sources

The typical kinesiology laboratory usually has several different platforms for real-time acquisition and archival storage of different kinds of data, including videotape, multichannel analog tape and high capacity digital files. Because of the rapid evolution of this technology, various clinical laboratories usually have different and changing mixtures of equipment. Unfortunately, there are no commonly accepted standards for timecodes, synchronization signals or file formats (although efforts are underway in Europe to standardize clinical procedures and coordinate frames for reporting kinesiological data). Motion analysis systems may be based on North American (NTSC, 60 fields/sec) video, European (PAL, 50 fields/sec) video, or nonstandard frame rates. When multichannel or wideband analog data are required concurrently, it may be desirable or even necessary to use a separate computer platform to digitize these data. Often only rudimentary techniques (e.g. light flashes, sound pips) are provided to indicate corresponding times that might have been signalled by an operator during the data acquisition or recording. Some data acquisition systems can only produce frame-rate synchronization pulses rather than respond to them. What is required is a universal timecode that can be locked to external synchronization pulses, recorded on all archival media, searched precisely for user-specified absolute times, and processed to create synchronized clock pulses at the various rates required by the different acquisition platforms.

We have selected the SMPTE timecode, which is in common use in NTSC video systems and which uses a Manchester-encoded carrier from which it is simple to extract a regular clock rate of 2.4 kHz. This clock (or multiples and submultiples of it) can be used to control the sampling rates of A/D convertors and variable frame-rate motion analysis systems. The SMPTE generator can be synchronized by an NTSC composite video signal, which in turn can be synchronized to a wide range of external clock rates via a phase-locked loop. Thus, even systems that can only generate sync pulses can be synchronized to individual image frames in simultaneous videotape recordings, to precise locations on analog tape tracks and to individual samples in digitized data files. We are in the process of bundling the various generators, readers and sync circuits in a single, microprocessor-based instrument that can be configured dynamically (via RS-

232 serial interface from any host computer) for a wide range of recording and analysis tasks in a typical kinesiology laboratory.

4 Graphic Visualization Tools for Kinesiological Data

Computers are increasingly being used to provide nontechnical users with access to sophisticated tools. However, this is still a love-hate relationship that succeeds only when the users are provided with an interface that relates naturally to the usual tools of their trades. In the case of medical and rehabilitation therapists, the natural way of viewing the body is based on anatomical structures (e.g. muscles, bones, joints, support points) rather than mechanical dynamics (e.g. torques, coordinate frames, degrees-of-freedom, reaction forces).

We have been developing an iconographic display in which successive phases of motion are depicted as a string of stick figures. At the user's request, the various types of data that have been *Collected* or inferred from model-based *Transformations* are depicted as quantitative icons attached to the corresponding anatomical points on the stick figures (see Figure 2). For example, the EMG recorded from a muscle may be shown as a symbol of the muscle whose thickness is related to EMG amplitude; net torque at a joint may be depicted as a wedge of a circle whose diameter and orientation at the joint indicate magnitude and polarity; the magnitude, direction and location of ground reaction forces may be summarized as an arrow of appropriate length and orientation pointing to the center of pressure. FNS applications are likely to generate an ever-expanding range of sensors, actuators and mechanical devices whose signals and states must be accomodated by easily identified icons in these displays.

The extension of this iconographic scheme to three dimensions is necessary and not trivial. It will be important to pick a basis set of 3D shapes that is sufficiently anatomically realistic so that they can be recognized by a clinician, yet sufficiently

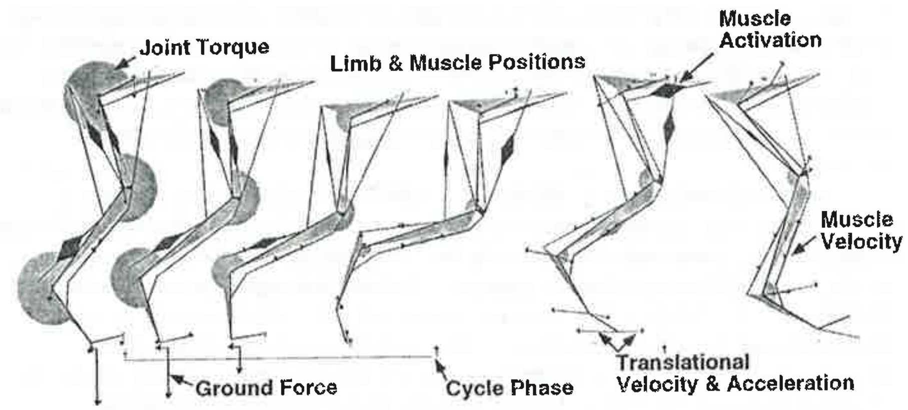

Fig. 2: Iconographic representation of stick figures depicting musculoskeletal mechanics at various phases of locomotion in the hindlimb of a cat (taken from a 2D color rendering on a Macintosh II computer using software developed by Chiang, Loeb and Levine; [10])

geometrically abstract so that complex figures can be drawn rapidly and without too much clutter obscuring the key features. The super-quadrix family may be suitable for this [6]. It will also be important to devise a convenient 3D pointing device, to specify aspect rotation, zooming and pointing to iconographic features. A number of force and position tracking schemes have been developed and are continuing to evolve for the analysis of 3D medical images, but they are still expensive and cumbersome.

5 Noninvasive Morphometry from MRI

Models of musculoskeletal mechanics make enormous demands on both the quantity and quality of morphometric data to describe the system. This is particularly the case for models to be used in inverse dynamic analysis of kinetics from kinematics and for forward simulations of trajectories from muscle activation. Both analyses are particularly important to understand pathological movements and to design corrective interventions such as FNS. Dynamic analyses of motor performance often rely on scaling of cadaveric data to the measured physiognomy of the subject, but this is likely to be unreliable when dealing with the sorts of pathology that accompany chronic neurological deficits.

Fortunately, it is now possible to obtain quite detailed and accurate cross-sectional images that reveal the dimensions and locations of individual muscles and tendons [7]. These can be used to quantify most of the key parameters for models of muscle function, including fascicle length and pinnation angle and physiological cross-sectional area [8].

6 Automated Kinetic Analysis

Both inverse dynamic analysis and forward simulation of trajectory require the equations of motion of the mechanical linkage under study. Even for simplified models being studied under highly controlled conditions, the generation of such equations is extremely tedious and provides many opportunities for error. Kinetic analysis will be indispensable for the application of expert knowledge to the fitting of FNS systems, but it must be readily available to clinicians with very limited knowledge of mechanics. Furthermore, it must be applied to complex and even variable linkages, such as patients equipped with crutches, mechanical braces, and surgical alterations of joints and tendons.

Commercial packages (e.g. SD-FAST, DYNAMAN, AUTOLEV) are now available and evolving rapidly to solve the equations of motion for arbitrary mechanical linkages. They have been used primarily for robots and satellite linkages, but are starting to be applied to biological systems. Much work is still required to make them compatible with biological structures and anatomical conventions, as well as the constraints and dynamics imposed by orthotic and prosthetic devices and their controllers. However, it is important to anticipate this requirement during the design of model structures for musculoskeletal anatomy and of coordinate frames for kinematic data.

Once the mechanical system is fully modeled (including both biological and prosthetic actuators and sensors), it may be possible to apply mathematical tools from

control engineering to the design of control systems that can be optimized for particular behavioral tasks [9].

7 Conclusions

The technologies outlined above represent promising solutions for most of the major requirements of a hypothetical clinical system for the fitting of FNS and hybrid orthotic-prosthetic devices. However, integrating them into systems suitable for clinical use and keeping up with each of these rapidly evolving technologies represents a major challenge. This challenge must be met before there is any chance of widespread application of FNS by the clinicians and therapists who must deal with the vast majority of neurologically impaired patients.

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FES POWERED ORTHOSIS (LSU-RGO) FOR PARAPLEGIC AMBULATION

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The Reciprocating Gait Orthosis (RGO) developed at Louisiana State University (LSU) in the 1970s, was first used in pediatric patients with severe musculosketeletal disabilities of the lower extremities (eg. spina bifida, muscular dystrophy, sacral agenesis, osteogenesis imperfecta, and limited cases of cerebral palsy). During the late 1970s and through the 1980s, successful applications were made to SCIs, mostly paraplegics, although some quadriplegics with residual upper extremity function also benefited. Since 1983, about 5,000 RGO units have been made and applied in the United States, Canada, Great Britain, The Netherlands, France, Israel, Australia and South Africa. At present, the RGO is fully covered by Medicare/Medicaid, private insurers, and by the health authorities of the various countries.

Although initial results with RGO in SCI patients were encouraging, the orthosis has several limitations, especially the relatively high cost of energy of ambulation compared with that of healthy subjects or wheelchair transportation. To reduce that energy cost, we developed (in 1983) a simple but effective and practical electrical muscle stimulation (MS) unit to power the RGO during walking.

Isolated upper-body work consumes much more metabolic energy per kilogram of body mass of active tissue, compared to combined work of the upper and lower body. This, added to the fact that the legs are paralyzed and the trunk and muscles are very active during locomotion with the RGO alone (without MS), results in excessive energy expenditure and stress on the arms. By using MS to power the RGO, we can produce strong enough contractions of the large thigh muscles to provide the swing and push-off functions while simultaneously creating combined upper- and lower-body work, thereby reducing the overall energy cost of locomotion. An additional advantage is the reduction in the work by and stress on the spinal and arm muscles produced by the swing/push-off in the absence of MS. (This is very important to the design concept, as will become obvious further on, when the system evaluation is considered.)

The LSU-RGO, therefore, maintains upright posture and balance, and electrical muscle stimulation of the rectus femoris of one leg simultaneously with the stimulation of the contralateral hamstrings will provide hip flexion and extension, respectively, for successful locomotion.

THE STIMULATION SYSTEM

As noted above, there are two objectives of muscle stimulation. The first is to initiate the swing of one leg simultaneously with contralateral push-off, thereby providing the power for locomotion while releasing the upper extremities and spinal muscles from that task. Second, because both the upper and lower extremities are active, stimulation reduces the energy expenditure per unit of body mass.

To stimulate the rectus femoris of one leg simultaneously with the hamstring of the contralateral leg, two channels are necessary. Furthermore, to initiate the next step, the cycle is reversed, which requires stimulation of the hamstrings of one leg with simultaneous activation of the contralateral rectus femoris. In all, four stimulation channels are required, with each pair active simultaneously.

The stimulation is accomplished with monophasic, charge-balanced pulses of 0.5 ms duration at a rate that varies from 18 to 26 pps, according to the individual patient. The objective of the rate adjustment is to generate a contraction of strength near 50 to 70% of the maximal tetanic force without inducing fatigue. Lower rates may accomplish this. Individual adjustments can be made to accommodate the muscle fiber composition of each patient and the changes resulting from the MS therapy administered to reverse muscle atrophy.

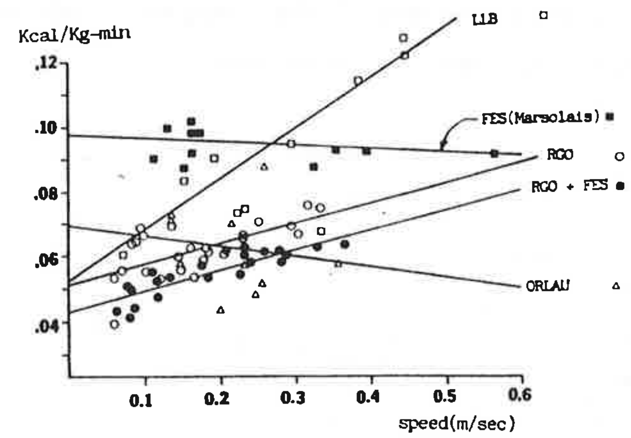
The pulsed current is applied to the patient via conventional carbon-impregnated rubber electrodes covered with Karaya solid gel.

We have designed a flexible copolymer electrode cuff, which is custom-made for each patient. The electrodes are placed in the cuff properly predistanced about the motor point. All electrode wires are passed between the outer shell and the internal foam cover and emerge from the cuff in a single cable with a plug connector. The connector can be inserted into the stimulator only in the correct way. The inside foam cover can be peeled off for washing or for replacing the Karaya gel; it also allows ventilation of the skin, prevents temperature buildup, and the absorption of sweat. Velcro straps fasten the cuffs snugly about the thigh.

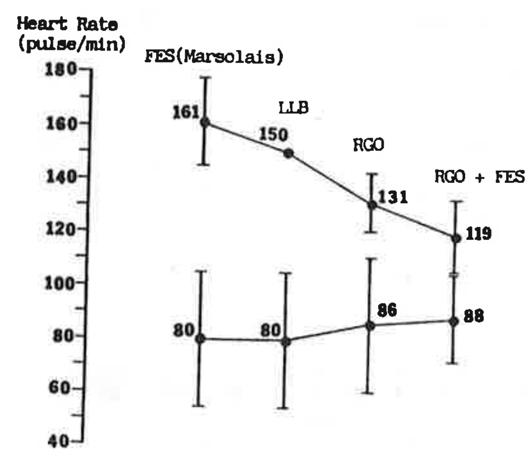
The stimulator is always in the "off" mode except when the patient decides to walk. By triggering a mini-switch mounted on each handlebar of the rolling walker, the patient activates the rectus femoris on the same side of the switch while stimulating the contralateral hamstrings. The trigger signal from the switch is transmitted to the belt-worn stimulator via a spiral cable from the walker.

EVALUATION

Six paraplegics were evaluated for energy consumption while walking in the RGO, RGO and MS, and compared to energy consumption of patients walking with the long leg brace and marsolais' muscle stimulation system. The results are given in the figure below.

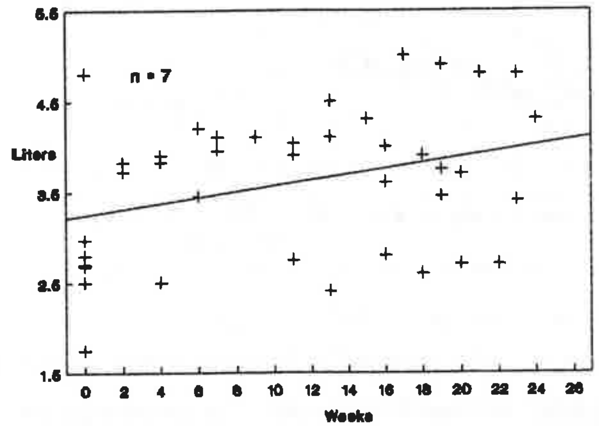


The same patients were also evaluated for heart rate before and after walking a 30 meter track using only RGO and then RGO and MS. The results were also compared to the heart rate of patients using the long leg brace and Marsolais' FES system as shown in the figure below:

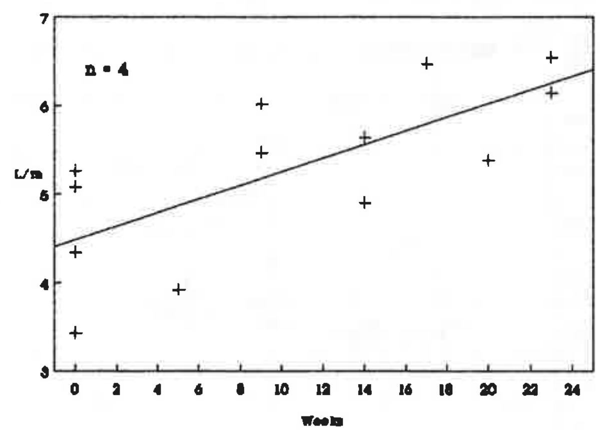


Additional long term evaluation of patient consisted on studying their cholesterol level, cardiac and pulmonary status as well as bone density before and after entering the RGO program. The results are shown in the figures below:

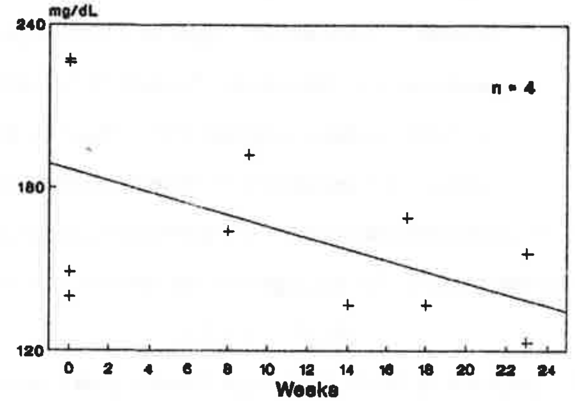
Vital Capacity



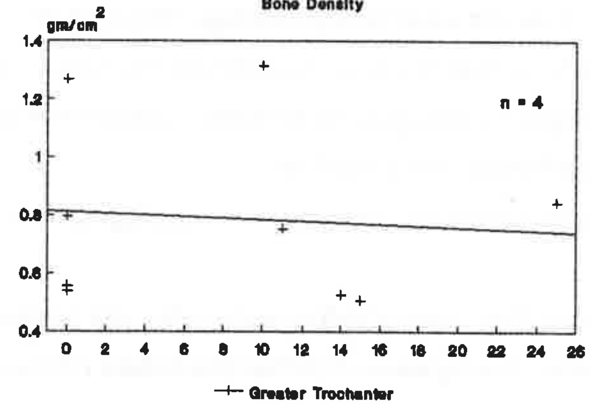
Cardiac Output



Total Cholesterol



Osteoporosis
Bone Density



It is now evident from the data above that:

A) The addition of electrical stimulation of the thigh muscles significantly reduces the energy consumption of paraplegics using the LSU-RGO for locomotion. In addition significant reduction in the heart rate after locomotion in the RGO and MS is the physical stress is reduced as consequence of using MS.

B) The locomotion of paraplegics in the RGO powered by muscle stimulation results in significant improvements in cholesterol content, cardiac/circulatory and pulmonary functions. No significant change was seen in bone mineral density.

Undoubtedly, providing paraplegics with the ability to walk freely has significant social, medical and psychological benefits to the patient and society.

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HYBRID FES ORTHOSES FOR PARPLEGIC LOCOMOTION

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INTRODUCTION

Approaches to the difficult problem of restoring locomotion after spinal injury include orthotics, functional neuromuscular stimulation (FNS), powered braces and hybrid systems.

The anterior floor reaction type ankle foot orthosis (FRO) [Saltiel 1969, Lehneis 1972, Yang 1986, Rizzoli 1990] makes use of the well known fact that a paralysed leg can be stabilised in extension without muscular action if the ground reaction force is directed anteriorly with respect to the anatomical knee joint axis. The FRO limits ankle dorsiflexion and thereby enables the base of the ground reaction vector to located in the metatarsal region of the foot. For reasons of safety, the application of the FRO should be restricted to patients with incomplete leasons who have sufficient voluntary control and have been adequately trained to avoid or recover from buckling at the knee. Knee buckling occurs during everyday activity whenever the ground reaction vector shifts behind the knee. If the patient cannot quickly recover then injury may occur as a result [Zeeb 1992]. With more extensive paralysis knee ankle foot orthoses (KAFO) are traditionally used. This approach provides very limited functional benefit and for individuals with complete lesions and are often rejected [Hussey 1973]. The gait is slow and the metabolic energy consumption unacceptably high [Gordon 1956] as are the force actions on the upper limbs [Crosbie 1990].

A significant innovation in hip joint control mechanism was independently developed by Scrutton in the UK [Scrutton 1971] and Motloch in Canada. The hip joints were mechanically linked to allow only reciprocal motion. A number of similar mechanisms have since been reported [Douglas 1983, Kemp 1989] and are now generically called reciprocating gait orthoses (RGO). Although these HKAFO systems have been successfully used with children the advantagous for adult users are less clear. For example, in a recent comparative study of the RGO and the Parawalker (HGO) HKAFO braces adult users often discontinued using the braces [Whittle 1989]. A significant number indicated encumbrance and interference with other ADL such as toileting to be a major factor. The RGO device can be donned and doffed from a bed or similar surface in 5 mins or more. Typically, the devices were worn for a few occasions per week for periods less than 45mins primarily for exercise. The devices are inconvenient to transport, cannot be rapidly donned and doffed from a wheelchair.

FES techniques have been recently reviewed in [Kralj & Bajd 1989]. Encouraging results have been obtained with paraplegics with incomplete lesions [Bajd 1985] using surface electrodes and impressive control of locomotion has been achieved paraplegics with complete lesions using multichannel implanted percutaneous electrodes [Marsolais & Kobetic 1988]. However problems remain that prevent

achieving reliable and control for everyday use for paraplegics with complete lesions. Some of these problems are a consequence of the high physical effort required [Marsolais 1988, Cliquet 1989], slow gait and rapidly induced muscle fatigue [Kralj & Bajd 1989], and the variability and habituation of flexion reflexes [Andrews 1989, Granat 1991]. At present, FES is considered useful for therapeutic exercise.

Powered exoskeletal walking aids have been reported, however, the encumbrance of power supplies and actuators limited clinical application. The concept of combining FES, external bracing and actuators was suggested in 1972 by Tomovic, Vukobratovic and Vodovnik [Tomovic 1972]. Some advantages have been demonstrated for hybrid systems including: reduced FES muscle fatigue; improved control of the pelvis and trunk especially for those with higher level spinal lesions; simplified control by limiting the degrees of freedom in joint motions; compensation for insufficient muscle force due to weakness or reflex habituation; a backup in case the FES component fails.

THE 2-PART HYBRID ORTHOSIS

The system comprises two parts, the first is a FRO/FES system worn to assist with transfers and limited range ambulation [Andrews 1988]. This part is unencumbering and is typically donned once and worn all day. The second part, a lightweight and compact hip/trunk component that is donned and doffed as required from the wheelchair and is intended to improve function [Andrews 1990].

Knee Stabilising Component The FRO can stabilise the paralysed leg without brace components encompassing or mechanically locking the knee joint. This allows free flexion during the swing phase that improves ground clearance and dynamic cosmesis. A sensor (force sensing resistors supplied by Interlink Electronics Inc USA) is used to detect incipient buckling of the knee and, in response, the computer control system immediately stimulates the knee extensors. Stimulation is removed after the knee is again stable in extension. Because the force vector is mainly oriented ahead of the knee during standing and the stance phase of walking, the activation of extensors can be reduced to levels that avoid FES induced muscle fatigue [Andrews 1988, Mayagoita-Hill 1990]. This can extends the endurance of upright activity such as walking if occasional rest breaks are taken, whilst standing, without collapse due to induced fatigue.

An improved design of FRO was fabricated from anteriorly moulded plastic as shown in figure 1. This is much stiffer in dorsiflexion, with thinner section material, than conventional posteriorly moulded designs. The FRO fits inside the patients normal shoe size and an elasticated calf band controls plantarflexion during the heel contact to foot flat phase of the gait cycle thereby avoiding the "rigid boot" feel of solid ankle designs. The FRO has been previously used by paraplegic individuals for FES assisted stand-sit manouvres [Andrews 1990] using phase plane switching control; prolonged standing [Mayagoita-Hill 1990] reciprocal [Andrews 1989] and swing through crutch aided ambulation.

Hip & Trunk Stabilising Component The first prototype incorporated modified Durr Fillauer RGO hip joints and a reciprocating mechanical linkage i.e. a flexible, push-pull, linear bearing (BowdenFlex 55 supplied by Bowden Controls Ltd UK) in place of the usual Bowden cables [Andrews 1990]. The linear bearing provided minimal backlash, very low friction with no cable stretching. Standing and walking was much more stable because bilateral hip flexion during the double support phases was inhibited. Two paraplegic users report having to use less upper limb effort during reciprocal gait than with the FRO/FES part alone. The hip/trunk brace could be donned and doffed in less than 30s from the wheelchair. When the brace was not in use could be folded and stowed conveniently on the wheelchair backrest or beneath the seat for ready access. Both users liked having an easily removable upper body component since it reduced interference with other activities. These initial laboratory trials also revealed problems with the hip/trunk brace hip joints and linkage including:

- The hip joint locking mechanism required the alternate use of both hands to disengage the linkage.
- The joints lock out only in one position and were difficult to engage if hip flexor/abdominal spasticity prevented full hip extension being immediately attained.
- The linkage required subsequent adjustments if the user was initially fitted with hip contractures that later stretched out. Conversly, adjustment may be required if contractures develop for some reason.
- In normal walking the relative motion of the unencumbered hip joints is complex. A typical trajectory is shown in figure 2. The linkage used constrained the hips to move in an approximately 1:1 extension to flexion ratio. This is unnatural way that may reduce efficient walking.
- It has been demonstrated that swing-through gait in which both legs are flexed and exdended together can be synthesised using FES to provide a potentially faster mode of walking [Granat 1991]. However, this is possible only if the linkage is disengaged with consequent poor stance phase control.
- It is impossible to lean forward to pick something up from a low table with the linkage engaged. The user must disengage the linkage and rely on his upper limbs for support.

To overcome these problems and provide for multi-mode gait control some form of "smart" hip joint mechanism was required that would provide dynamic control of hip motion during the different phases of locomotion. Such a joint and its power supply and drive train must be physically small yet be able to provide or at least restrain the high torques transmitted across the hip joint.

Hip Joint Motion Control Available batteries, electric motors and geartrains prohibit fully powered joint control of flexion and extension because of size and weight constraints. It was decided to investigate joint motion control restricted to braking. The wrapped spring clutch (WSC). The was selected for initial investigation mainly because of its inherently high torque to size/weight ratio and minimal power requirement.

The principle of operation of the WSC hip joint is illustrated schematically in figure 3. A drive spring is wrapped around two lightweight metal hubs on a common shaft that form the joint. One hub is fixed to the pelvic band and the other to the side steel of the thigh cuff. One end of the drive spring is fixed to the pelvic band and the other end is formed into a control tang that can be displaced

by an eccentric cam on the output shaft a small model servomotor. In the present arrangement the servomotor moves the control tang into one of two positions corresponding with "free" and "engaged" states. In the engaged state the joint acts like a toothless ratchet in which extension tends to unwrap the drive spring allowing free movement in that direction. However, if the motion is reversed i.e. flexed, then friction between the hub and the spring will immediately cause the drive spring to wrap down onto the hubs and prevent further flexion. In the free state the eccentric cam moves the control tang in the anticlockwise direction tending to unwrap the spring off the hubs thus freeing motion in both flexion and extension directions.

In the present arrangement the drive spring has dimension 54 mm diameter, 30mm length formed from 2.2mm square drawn wire and can hold up to 100Nm in the engaged state. The holding torque can be varied by fine control of cam displacement. If the applied torque exceeds the set torque the drive spring will slip on the hubs at the set torque. In the free state the frictional drag is less than 0.02Nm. In the engaged and free states the servomotor [Futaba model S148] consumes less than 50 mW at 7.2 volts in use. The weight of the WSC and servo components is 0.3Kg. The joint can switch states in less than 50ms.

Typically the required state is selected by a digital output from the control microcomputer. Typically the joint is "engaged" during stance phases and sit-stand manoeuvres and is "free" during the swing phases and stand to sit movements. Whilst sitting the wheelchair or standing for long periods the servomotor power can be switched off to conserve battery power. In order to stand up the user positions himself on the edge of the seat with both legs flexed and the body centre of gravity over the feet. Then with the hands on the support the joints are switched to engaged throughout the standup movement and whilst standing. Any difficulty in immediately attaining full upright hip extension is automatically accommodated since both WSC joints allow progressive extension with the user held in the maximum extension attained. The system has recently been used in the laboratory by one paraplegic with complete thoracic lesion (T1) for sit-stand transfer using surface electrode applied FES. No manual interaction with the joint mechanisms being required of the user and standing for periods in excess of one hour was possible. The hip/trunk component could be donned and doffed in less than 30 seconds.

DISCUSSION

Previous work with the FRO/FES part of the system suggests that the speed of reciprocal gait is slow, typically 0.1-0.5m/s and energy consuming and that swing-through gait involving bilateral FES flexion of the legs was faster (typically 0.6m/s) with similar levels of energy consumption [Heller 89, Heller 90, Granat 1991]. A limiting factor in swing through gait was the instability of the hips and trunk particularly during the initial stance phase. Laboratory tests are presently being conducted to determine the potential role of the present 2-part system with this gait pattern.

It is anticipated that some combination of manual control and sensor driven, rule based intention and event recognition will be used to control both the FES activation of muscles and switching of the WSC states. Determining the optimal balance of manual and automatic controls is a major challenge for further development of these cybernetic systems.

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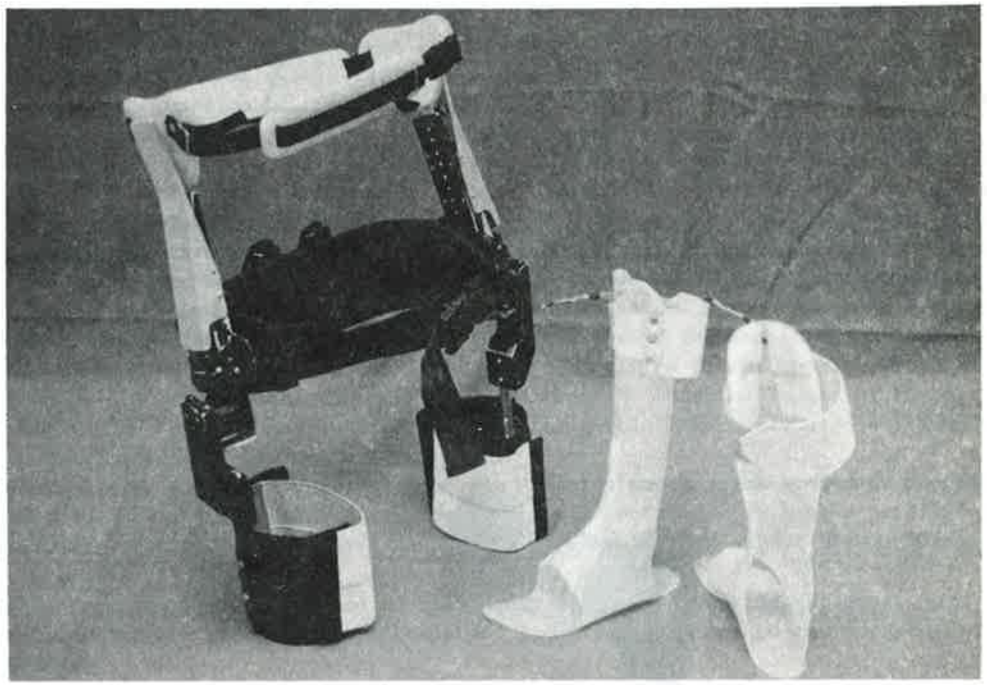

Figure 1 The prototype 2-part system showing the anterior floor reaction orthoses and the hip/trunk brace.

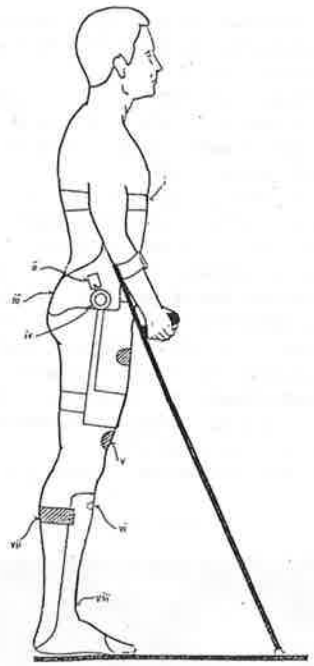

Figure 2. Schematic diagram of the two part hybrid FES HKAFO system showing i) chest band, ii) servo, iii) pelvic band, iv) WSC, v) electrode, vi) FSR pressure sensor, vii) elastic calf band, viii) anterior molded FRO.

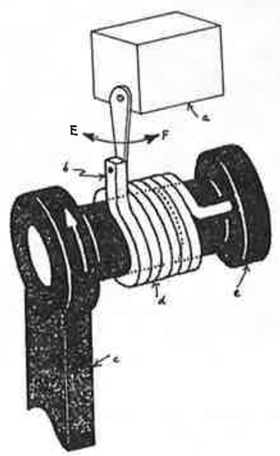

Figure 3. Schematic diagram of the right wrapped spring hip joint. Showing a) servo, b) control tang, c) joint hub connected to the thigh side steel, d) drive spring, e) joint hub connected to pelvic band.

SPECTRAL COMPRESSION OF THE EMG SIGNAL AS AN INDEX OF MUSCLE FATIGUE

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Among Health Scientists, it has become customary to describe and evaluate muscle fatigue in terms of the force that can be produced by the muscle.Of particular concern is the use of the failure point of a muscle to produce a desired force as the point in time when the muscle becomes fatigued. This index would detect fatigue only after it has occurred.Additional complications are introduced by the lack of direct access to the force produced by an individual muscle and the relative influence of the physiological and psychological components in determining the time at which the failure point occurs.Pratical considerations aside the notion of using force as an indicator of fatigue is problematic because even when the force output is maintained constant, time-dependent physiological and biochemical processes continuously alter the means by which muscles generate force casting concerns on the physiological meaning of the use of force for measurements of muscle fatigue.Consequently the use of force as a fatigue index is not well suited for applications in the clinical and ergonomics environments.
It is suggested that muscle fatigue is more correctly (and usefully) viewed as a continual function of contraction time.A technique that accomplishes this need is described and is referred to as the EMG spectral compression technique.This technique is based on the well-known fact that the frequency spectrum of the EMG signal is continuously compressed during a sustained contraction.The median and mean frequencies of the EMG signal are recommended as the preferred variable for tracking spectral compression. It will be shown that during a sustained contraction the median frequency is affected mostly by the PH in the muscle (which depends on the amount of net Lactate that is produced and removed)as well as some other unknown factor(s).As a corollary the decrease in the median frequency will be shown to be dependent on the fiber type composition of the muscle. Finally through the use of models it will be shown that the changing length of the depolarization zone, which propagates along the muscle fibers,may also contribute to the decrease of the median frequency.
It is argued that this objective and non-invasive non-painful technique provides superior means for assessing and monitoring muscle fatigue in humans performing voluntary contracitons as well as providing convenient means for studying some biomechanical modifications within the muscle without invading the muscle.Also this technique may be applied to individual muscles; a signifiacnt asset when considering the relative contribution of individual msucles to the torque at a joint.

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ASSESSMENT OF MUSCLE FATIGUE AS AN INSTRUMENT IN THE PREVENTION OF WORK-RELATED MUSCULOSKELETAL DISORDERS

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Musculoskeletal disorders is common in the general population and in the working population (1). Recent research has shown scientific evidence of work-relatedness for many disorders, especially in the upper limb including the shoulder and neck (1-4). Tendinitis in the shoulder has been reported as prevalent as 18 % among welders and 16% among plate workers over the age of 35 years (5). When contrasted to male office workers the estimated relative risks were 13 and 11 respectively. This implies that more than 90 % of the shoulder tendinitis in the welders and plate workers is preventable. The exposure that was attributed to the development of tendinitis in the shoulders among the welders and plate workers were static and repetitive work with the hands at or above shoulder level causing local load on the supraspinate muscle and fatigue. An increased risk of 11 for shoulder tendinitis was also reported among truck assemblers (6). The exposure for the assemblers was also claimed to consist of muscle fatigue (7, 8). Myofascial syndrome and tension neck syndrome are associated with work with video display units (1, 9). The prevalence of tension neck among data entry workers are in some studies higher than 50 % (1). The relative risks associated with keyboard work and tension neck syndrome are in between 3 and 4 which implies a preventable fraction of 67 to 75 % among the exposed workers. The etiologies of work-related myofascial syndrome and tension neck syndrome is still obscure. The presented hypothesis involves local muscle fatigue, selective motor unit fatigue, local ischemia and motor imbalance to mention a few (10-12) .

Prevention of work-related musculoskeletal disorders is not the only reason to improve the ergonomics of the work environment but also to improve health outcomes such as comfort is of importance. Comfort may influence the quantity and quality of production. Development of discomfort has been reported associated to the duration of VDT work and repetitive arm work (13, 14). Discomfort in the shoulder neck has been linked to development muscle fatigue measured by electromyography tremor (motor control), and perceived exertion (15-17).

Assessment of muscle fatigue is thus important in the prevention of not only work-related musculoskeletal disorders but also in work-related musculoskeletal discomfort.

Definition

My definition of muscle fatigue is a reversible impairment of motor function. The duration of this impairment can vary between microseconds to hours

maybe days. Pain or discomfort is associated with muscle fatigue as well as perceived impairment of motor performance. Muscle fatigue can be regarded both as an outcome to exposure but also as a dose of the same exposure. This is illustrated in the following model.

Model

An international scientific group has agreed upon a model for work-related musculoskeletal disorders (Figure 1) (2). The model confirms to a standardized terminology for occupational epidemiology (18). The exposure is external to man. Example of exposures are the requirements of work, shape of tools, weight lifted, speed, the geometry of the work place, the location of parts and tools etc. The exposure will cause a dose. The dose is internal in the individual. A dose could be the tension in the supraspinate muscle in a forward flexion of the arm to reach a tool. This tension may cause a response of an increase in intramuscular pressure. The increased intramuscular pressure may be regarded as a new dose causing the response of an impairment of blood flow. The dose response relationship is dependent and modified by the capacity. The impairment of blood flow may be dependent upon the blood pressure which is a capacity factor. The capacity can be muscle strength, blood pressure etc. and is also influenced by the response. The response may condition the tissue leading to improve health or cause mechanical, chemical or physiologic failure leading to disorder and pain. Muscle fatigue may not necessarily be hazardous to the muscle, but could be a normal physiological process in strength and endurance conditioning. Muscle fatigue can thus be both dose and response in the model.

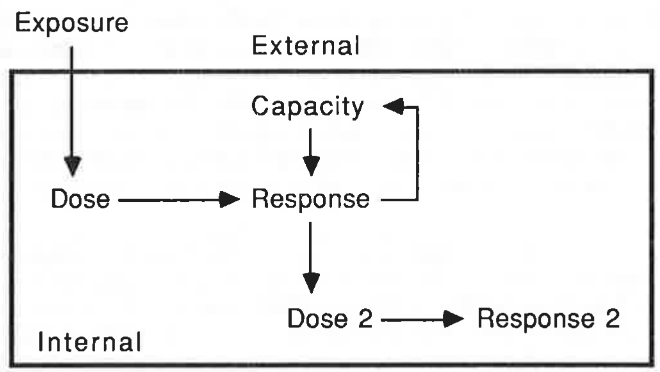

Figure 1. Model for exposure-dose-response relationships in the musculoskeletal system. In this model, muscle fatigue can be considered both a dose and a response.

Muscle fatigue as a response and a health outcome

Can muscle fatigue be used as a diagnostic aid in work-related musculoskeletal disorders?

Female workers in an electronic plant unable to work due to suspected shoulder neck myofascial syndrome had a greater susceptibility to electromyographic signs of muscle fatigue in the upper part of the trapezius and the supraspinatus muscle (19). This were shown by a short endurance time in their ordinary working posture which corresponded well to their physiologic muscle capacity measured as the time constant of electromyographic signs of muscle fatigue. Biopsies from patients with work-related musculoskeletal disorders have revealed disturbance located to the type 1 fibers (20).

In a later study Suurküla and Hägg (21) found more rapid development of electromyographic signs of fatigue among female assembling workers with more than a month of sick leave during the last year due to neck and or should disorders compared to all the others. Electromyographic fatigue development in the descending part of the trapezius muscle was significantly faster in a group of patients with ankylosing spondylitis during an endurance time test than in a group of referents (22). This indicated a change in muscle function consistent with previous reports of myopathy in patients with ankylosing spondylitis.

In a prospective study of low back pain Biering Sørensen (23) found that good isometric endurance of the back muscles prevented first time occurrence of low back pain in males. Muscle fatigue measured as back endurance may indicate back health status. Measurement of tremor has lately been suggested as a sensitive instrument for muscle fatigue (17, 24).

In a study of vibration exposed male works subjective reporting of tremor in hands and arms were common (25). When contrasting the vibration exposed workers to non-manual white collar workers the relative risk of reporting tremor in hands and arms varied between 7 and 46. The reporting of tremor in arm and hand may suggest an increased fatigability in the arm and hand which affects motor control (26). One inspiring hypothesis is that tremor measurement can be used for evaluation of work-related musculoskeletal disorders.

It is evident from what I have presented here that increased susceptibility to muscle fatigue may be used as a health outcome. It is still not clear what underlying pathology this health outcome reflect. Increased susceptibility to muscle fatigue as a health outcome can not be used as a single variable in diagnose setting or as a determinant of treatment. Furthermore, muscle tissue related pathology is only a fraction of the work-related disorders. For work-related tendon disorders or nerve disorders increased susceptibility to muscle fatigue is probably not an accurate health outcome.

Can muscle fatigue be used as an outcome to evaluate exposure?

In a study of shoulder posture and localized muscle fatigue Wiker et al (17) found that posturing hands above shoulder level significantly increased the risk for muscle fatigue and postural discomfort even in light weight manual

assembly work were postural exertions were small. Muscle fatigue was the health outcome to different postures. Furthermore they found that postural tremor were more sensitive metrics of muscle fatigue in the shoulder complex than EMG variables. In an automobile upholstery plant Habes (27) found EMG amplitude signs of muscle fatigue in back muscles which were more pronounced in a work task demanding more forward flexion. Muscle fatigue measured both by electromyography and by tremor measurements corresponded to the outcome discomfort. There are several studies that shown a relationship between electromyographic signs of muscle fatigue and perceived exertion and perceived discomfort (13, 15).

Muscle fatigue measured by EMG has been used as a health outcome in relation to different exposures, for example, work postures, work place design and work pace (13, 15). Great care should be taken in interpreting results of EMG muscle fatigue since besides sources of bias the finding of muscle fatigue may reflect a normal physiologic behavior of the muscle. Muscle fatigue measured as endurance, EMG amplitude and frequency changes, perceived discomfort could serve as a base when ergonomic guidelines and standards are developed (28).

Muscle fatigue as health outcome - conclusions

1. EMG muscle fatigue susceptibility may indicate myofascial syndrome.
2. Back muscular endurance as a measure of muscle fatigue may be a predictor of back pain incidence.
3. It is still not clear what underlying pathology increased susceptibility to muscle fatigue as a health outcome reflect.
4. Muscle fatigue may be an objective way of assessing health outcome in relation to different exposures.
5. Great care has to be taken when interpreting muscle fatigue as health outcome since the fatigue may be a normal physiologic event.

Muscle fatigue as a dose measure

Can muscle fatigue be used as a dose measure when evaluating work-related musculoskeletal disorders?

The dose is defined as the amount of substance that reaches susceptible targets in the body (18). Muscle fatigue may be used as an indicator of the dose and relate to health outcomes such as disorders or discomfort. There is yet no study that I am aware of which have shown a clear dose response relationship between muscle fatigue and a work-related musculoskeletal disorder. In some studies however, this relationship is indirectly indicated. Muscle fatigue in shoulder muscles measured by electromyography was found among assembly line workers (29). Different work tasks were examined and one task called "wet rubbing" caused the most pronounced EMG fatigue signs and was also regarded as the heaviest work task among the workers. Among construction workers in overhead building tasks, muscle fatigue in the supraspinatus and in the upper part of the trapezius muscle was shown.

Disorders in the shoulder and neck among carpenters and painters are clinical problems. Overhead welding was found associated with EMG signs of muscle fatigue (30). The welders had an excess risk of shoulder tendinitis. In truck assembling EMG signs of muscle fatigue was found during the work day in the supraspinatus muscle for four out of five assemblers (7). The assemblers were also at high risk for shoulder tendinitis and tension neck syndrome. When trying to use the development of electromyographic signs of fatigue in shoulder muscles during work as a predictor for incidence of neck and shoulder disorders a two year longitudinal study of female assembly workers did not show any information value (31).

Is measurement of muscle fatigue a valid dose measure for work-related musculoskeletal disorders?

Different measures of muscle fatigue may provide simple estimates of the dose for the muscle. It is likely that assessment of muscle fatigue is not sufficient to evaluate the strain on tendons, insertions, joints and nerves. Load duration, load variation and tendon and joint motion appear to be important dose measures in the evaluation of work-related tendon and nerve disorders (32).

Muscle fatigue as a dose measure - conclusions

1. Different measures of muscle fatigue may provide simple estimates of the dose for the muscle.

2. It is likely that assessment of muscle fatigue is not sufficient to evaluate the strain on tendons, insertions, joints and nerves.

Future trends and research needs

Although muscle fatigue can be measured accurately by EMG, the drawback with this method is the great amount of time usually required for analysis. Recent development with microprocessors have introduced small carry on units in which both sampling and analysis are done. It may be possible to use some of these units as biofeedback instrument to warn the user of muscle fatigue. This may be one contribution in the prevention of myofascial syndrome. It is doubtful whether this type of biofeedback can be used to prevent work-related tendon disorders or nerve entrapment disorders.

In the future there is a great need to relate muscle fatigue to different exposures in a systematically way. The data can together with discomfort evaluations serve as a basis for ergonomics standards settings. Much effort have been put into using EMG as the measure of muscle fatigue. There are indications that other measures such as tremor and motor coordination test may be more sensitive to muscle fatigue. When evaluating low level static contractions which are common in occupational settings, traditional EMG fatigue analysis may be insufficient (16). Fatigue pattern analysis evaluating fatigue in different parts of the muscle is a recent trend in EMG research which may contribute to the understanding of muscle fatigue (13). Biochemical measures e.g. calcium ions, enzymes etc. may be good estimators of muscle fatigue but further research is needed (33, 34). Since muscle fatigue affects motor coordination there might be a possibility to use

muscle fatigue evaluation as a determinant of risk for musculoskeletal accidents? Endurance performance as a risk factor to different work-related disorders needs further attention. In future research please define what you mean by muscle fatigue and whether you regard it as an outcome or dose

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MOTOR UNIT FIRING AND SERIAL REACTIONS: FUNCTION DURING CHANGES DURING REPETITIVE ACTIVITY

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Continuous motor performance in daily activity is provided by repetitive activity of the nervous system. Their limits in degree of force elicited and in time during which it can be maintained are generally indicated as maximal contraction and endurance (maximal time of performance). A more usual concept is that of fatigue that concerns with both biological and psychological events occurring while the performance is still in progress, but at a lower degree of efficiency. The neurological assessment of fatigue has shown so many difficulties and intricate questions that it has been proposed to give up his use in neurological sciences [1].

A study of fatigue implies three levels of investigation [2]: 1) work output which can be measured as overt activity (force) and as motor or neural function (firing of action potentials); 2) biological impairments at the tissue level; 3) fatigue proper i.e. the subjective residue of feelings of bodily discomfort and aversion to effort.

A basic definition of a fatigue index focuses on the decline in maximal force produced during a motor performance carried out under pressure for speed or effort, after a certain time (t_x):

$$\frac{F\,Mx\,t_0 - F\,Mx\,t_x}{F\,Mx\,t_0} \cdot 100$$

This represents the universally accepted criterion of classification of Motor Units types. However before such a decline takes place, particular functional changes occur which are worth considering for both physiologists and neurologists. An early effect that is observed during protracted motor performance is represented by a slowing of the muscular twitch. At the same time the firing frequency of the single M.U. decreases. This functional change has been interpreted as a M.U. wisdom effect rather than as expression of fatigue [3]. On the other hand a main stress has been laid on the slowing of conduction in muscle fibres that can be detected by means of discrete Fourier analysis of the EMG; according to Alfonsi et al, it should represent a reliable index of localized muscle fatigue [4]. However the real meaning of this change and its place among fatigue related processes is still controversial and the problem of central versus peripheral issue in local muscular fatigue remains unsolved [5].

Anyway consistent functional changes, beside decrease in MU firing frequency, occur during prolonged voluntary contraction before a decline of force and a refusal take place. Thus we have applied the 60'' test of Marsden [3] for recording the M.U. firing. Moreover since the main interest of the neurologist is concerned with the purposeful complex voluntary activity a further systematic interpretation has been carried out on

serial verbal choice reactions performed by the subjects in different conditions [5, 6]. Besides normal subjects a few patients with motor system impairment have been included in the study.

Methodology

1. Firing frequency of single M.U.s during 60'' voluntary maximal contraction.

Single M.U. action potentials were recorded by means of microelectrodes (Disa 13K87) and bipolar concentric electrodes (Disa 13K80) from the adductor pollicis and the fourth interosseus of the right hand of 10 normal subjects (32 to 65 years old) during a 60'' maximal voluntary contraction (MVC). In Fig.1 an example is given with simultaneous recording with microelectrode and bipolar concentric needle electrodes. Details of the technique - which was initially applied in the Service of EMG of Geneva University with Dr. Gérard Roth - were given in a previous paper using the Averaged Proportional Consecutive Internal Difference (APICD) [8]. In one case (male, 64 y.o.) M.U. action potentials were recorded from the tibialis anterior muscle. The same investigation has been carried out also in 4 patients with severe paralysis due to Motoneuron Disease, before and after 1 to 2 mg TRH-t im. injection [8].

2. Latency time and motor time in serial verbal choice reactions.

The standard research was represented by 12 serial reactions at 3 ± 1 s interval for the words /clan/-/cane/ and /tram/-/toro/ in random succession. Forty-five Italian speaking subjects, aged between 16 and 78 years, were investigated with this thecnique. In two cases the reactions were continued for 42 trials at a lower interval of 1 s. One patient (23 years, male), affected by Wilson disease with mild dysarthria in a phase of remission, under longlasting penicillamin treatment was also investigated. Details on the methodology used and the basic statistic evaluation are reported in a previous pubblication [6].

Fig. 1 Simultaneous recording of a single M..U. from Ist Interosseus dorsalis muscle of a normal subjects with microelectrode (upper trace) and bipolar concentric electrode (lower trace).

Results

1. Single M.U. firing. The variability in maximal firing frequency during the Marsden's test.

1.1 Normal subjects.

In the 60'' test the maximum firing frequency shows marked oscillations before a decline of mean frequency is observed. This was evident even when we measured the reciprocal of 10 ISIs (Fig. 2, 3 and 4). Taking into consideration these oscillations expressed by the standard deviation of the mean of 12 intervals (named jitter of the ISIs) and evaluating the progressive increase in the means along the 60'' MVC, we have found three main kinds of course of firing frequency during the test.

1.1.1 Tonic non-fatigable M.U.s (Fig.2).

Low threshold M.U.s could be recorded in all subjects (10 M.U.s). Mean firing frequency at onset was in the range of 23 to 27 Hz with a mean of 29 and it was rather constant until the 40th s. In this period mean ISI was 32 ms ± 7 (=21%). After the 40th s to the 60th s the mean firing frequency diminished but never reached a value as low as 20 Hz.; on the same time the ISI jitter decreased until 24 to 14%.

1.1.2 Fast non fatiguable M.U.s (Fig. 3).

Only three of such M.U.s were recorded in our subjects. Mean firing frequency at onset was in the range of 41 to 53 Hz with a mean of 45 Hz which persisted rather constant until the 60th s. The ISI mean value was 22 ms ± 8 (=28%).

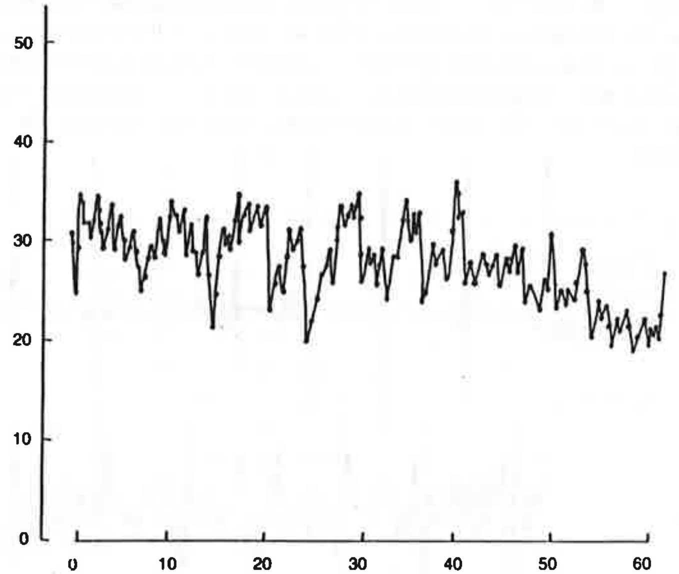

Fig. 2 Firing frequencies of one M.U. of tibialis anterior muscle of a normal subject (male, 64 y.o.) during 60' protracted MVC. In the ordinate: firing frequency per second (Hz) calculated as the reciprocal value of the ISI, each point corresponds to the means value of 12 discharges. The time of the trial in seconds is given in the ascissa.

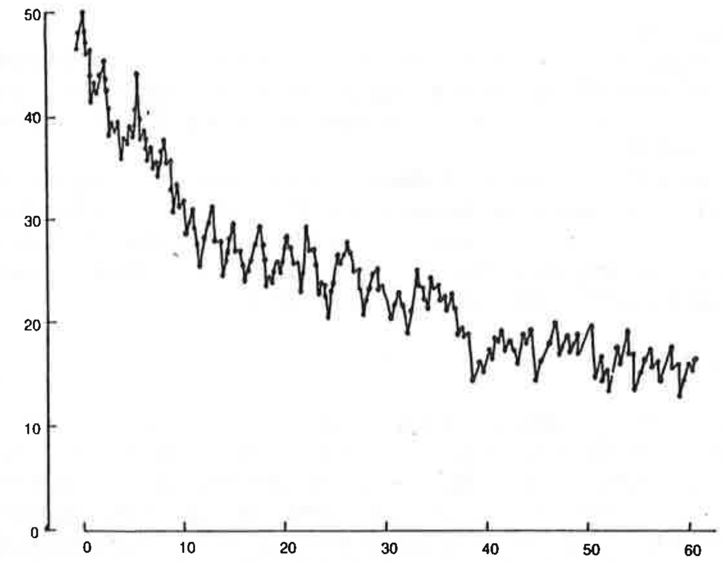

Fig. 3 Firing frequencies of one M.U. of Ist interosseus dorsalis muscle of a normal subject (male, 64 y.o.) during 60' protracted MVC. In the ordinate: firing frequency per second (Hz) calculated as the reciprocal value of the ISI, each point corresponds to the means value of 12 discharges. The time of the trial in seconds is given in the ascissa.

1.1.3 Fast fatiguable M.U.s (Fig. 4).

This kind of M.U.s could be recorded in 7 normals. Mean firing frequency at onset was in the range of 46 to 55 Hz with a mean of 48 Hz; after the first ten seconds it declined to a very low value (14 Hz). In this last period ISI jitter mean value was 86 ms ± 12 (=19%)

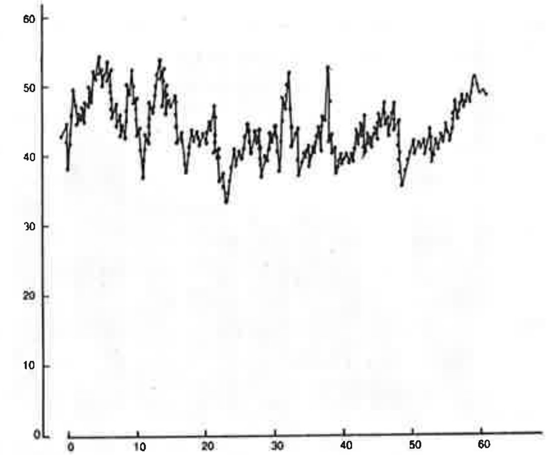

Fig. 4 Firing frequencies of one M.U. of Ist interosseus dorsalis muscle of a normal subject (male, 64 y.o.) during 60' protracted MVC. In the ordinate: firing frequency per second (Hz) calculated as the reciprocal value of the ISI, each point corresponds to the means value of 12 discharges. The time of the trial in seconds is given in the ascissa.

1.2 Motoneuron disease.

Two patients with motoneuronal systemic impairment were also examined. The former, a 54 years old male with a diagnosis of MND 6 years before, showed a maximal firing frequency of 27 Hz with an endurance of 25 s. The ISI mean value was 39 ms $\pm$ 6 (=15%).

The latter, a 45 years old female affected since 12 months by a very severe MND, showed a 11 Hz maximal firing frequency and 25 s endurance. The ISI mean value reached 95 ms $\pm$ 11. The im. injection of 2 mg TRH-t could increase both parameters: maximal firing frequency raised from 27 to 36 Hz and from 11 to 14 Hz with endurance increase of 85% and 130% respectively for the two patients.

2. Choice verbal reactions

2.1 Short series (12 repetitions with 3 s intervals).

No increase in RT (reaction time) nor in ACG duration (motor time, MT) were observed in the course of the 12 trials, but on the contrary a clear trend to continuous diminution in RT represented a constant finding. in Fig. 5 are represented the fluctuations for both EMG and ACG latency times: $\pm$ 50 ms variations for ACG and EMG RT mean values of /cane and /clan/ were observed in half the cases. No significatant changes were found in the duration of the ACG.

2.2 Long series (42 repetitions with 1 s interval).

If the subject was urged to speak a 18% increase in RT in the last ten reactions it was observed in 3 cases whereas no significant changes occurred in the ACG duration. No blocking was observed in these cases. On the contrary in the case with mild extrapyramidal symptoms (Fig. 6) a blocking occurred at the 17th trial with subsequent decrease in the RT below the mean value. A block was found to occur at the 25th trial but it was not followed by any significant decrease in RT.

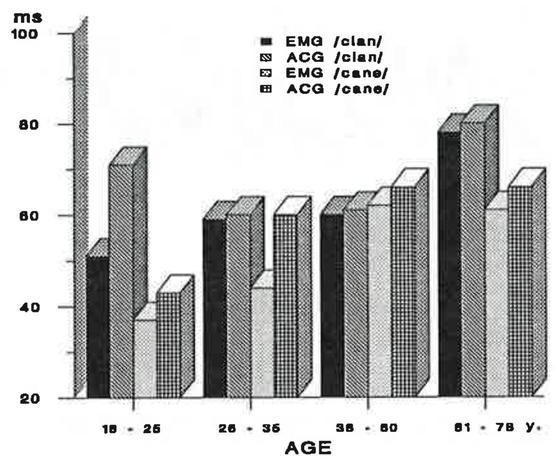

Fig. 5 Mean intraindividual ACG and EMG RT variability of choice verbal reaction for the words /clan/ - /cane/ in 34 healthy subjects divided in 4 age groups.

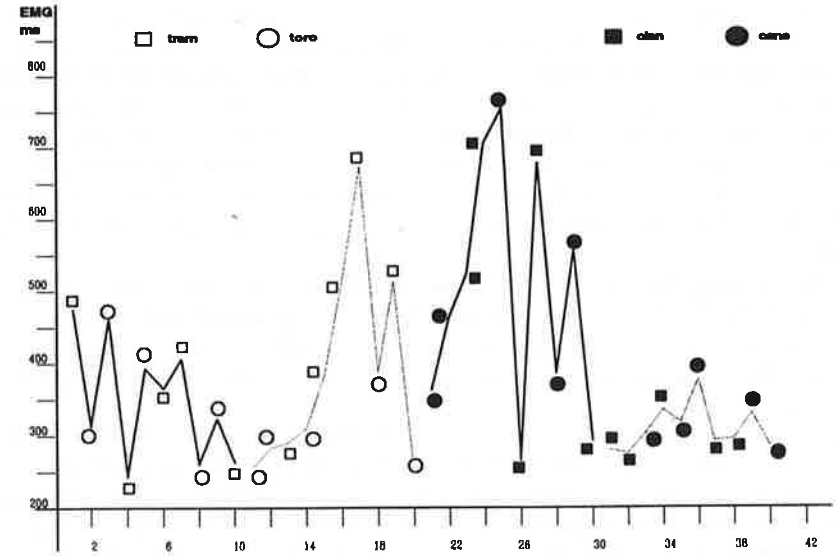

Fig. 6 EMG RT of 42 successive choice verbal reactions in a patient (28 y.o., male) affected by mild extrapyramidal symptoms.

Discussion

1. Single M.U. firing.

The Marsden's test represents a quite artificial motor performance: the pressure on maximal contraction leads to an unusual effect which can lead to extreme conditions with a failure in metabolic activity and hence to decrease in mucle fibre conduction velocity. This occurrence of muscular fatigue can exert a consistent effect not only on the muscle fibre conduction values but also on the M.U. firing frequency. The fluctuation in the firing frequency (ISI jitter) makes its appearance in the earliest phase of the innervation, when no leading effects of fatigue are present. It should therefore be considered as a still normal pattern resulting from a balance between opposite processes like decrease and respectively increased synaptic threshold [8].

The marked reduction in ISI jitter that can be observed in pathological conditions is in agreement with the results of Grimby [9] and Andreassen [10] which corresponds to a decrease in the level of high performance; the RT is abnormally long and in this reduced performance a stereotyped activity can develop. The effect of TRH-t shown in pathological decrease in M.U. firing frequency and endurance is in line with the gain effect of the TRH-ergic bulbospinal system that is activated during intense activity of the motor corticospinal system. The occurrence of double discharges induced by TRH-t [7] seems not related to fatigue phenomena but rather to a kind of temporal recruitment able to induce more efficient and rapid force production.

2. Serial verbal choice reaction.

A more natural motor performance is realized in the serial verbal reactions. When 3 ± 1 s intervals are left between two successive reactions no systematic changes in RT and ACG duration were observed. This constancy was found also for the oscillations or jitter in RT which could be related to oscillations in attention; avoiding psychological inferences they could be explained as the result of a balance between environmental request and refuse to continue. In sollicited and closer repetitions, the latency time may slightly increase whereas the ACG duration (which represents a motor time) remains unchanged, a dissociation that speaks in favour of functional changes occurring at a neural level.

The blocking events seem a rare phenomenon particularly occurring in very long activities far beyond that studied until now by us with the present methodology. In our patient with extrapyramidal impairment the blocking could be explained on the basis of deranged motor organization, at variance with the microsleep phenomena reported in normal subjects [11, 12].

In conclusion our findings on both neuronal (Marsden's test) and behavioral level (purposeful action) are in agreement with the concept of the neuron as a stochastic entity [13]. The origin of the fluctuations (ISI jitter and RT jitter) implies a dualistic neural organization. This in fact represents a common reality at all levels from the elementary one of depolarization versus repolarization in biophysics, to the more complex one of excitatory versus inhibitory processes in CNS functions.

The measurements of the localization of *muscular* fatigue maintains its value in some muscular diseases and for the assessment of artificial electrical stimulation. On the contrary in the neurological diseases the development of functional tests of fatigue should be concerned with the processes occurring at the *neuronal* and *behavioral* levels.

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MULTIPLE MOTOR REPRESENTATIONS IN THE AGRANULAR FRONTAL CORTEX OF PRIMATES

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Abstract. Classically, the agranular frontal cortex of the monkey and man has been considered to be formed by two motor areas: the primary motor area and the supplementary motor area. Evidence is presented that this subdivision is inadequate and that the agranular frontal cortex is constituted of at least seven different cortical areas. One of them (F1) corresponds to the primary motor cortex. Two areas are located on the mesial cortical surface (F3 or "SMA-proper" and F6 or "pre-SMA"), two form the superior sector of area 6 (F2 and F7), and two form the area 6 inferior sector (F4 and F5). It is suggested that these various areas play a different role in motor control and that some of them (e.g. F4 and F5) are mostly involved in transforming visual object properties into motor acts, some (e.g. F2) are involved in proprioceptive control and proprioceptive representation of arm movements and finally some (F6) in decisions about movement initiation.

The classical studies with electrical stimuli showed that in the agranular frontal cortex of the monkey there are two complete representations of body movements.[1,2]. The first is located on the dorsal cortical convexity just rostral to the central sulcus. It forms the so called "primary" motor cortex. The second is located mostly on the mesial surface of the hemispheres. It lies on the upper bank of the sulcus cinguli and the mesial brain surface dorsal to this fissure. This representation forms the so called "supplementary motor area".

Evidence accumulated in the last years showed that this simple parcellation of the frontal motor region is not adequate. This new evidence indicates that the agranular frontal cortex is formed by at least seven separate areas. The aim of the present article is to give a concise picture of the functional and anatomical organization of these areas. Particular emphasis will be given to the areas lying outside area 4.

Schematically, the agranular frontal cortex can be subdivided into 4 macroscopic regions: Precentral cortex (area 4), Inferior area 6, Superior area 6, and Mesial area 6. The three area 6 sectors are not homogeneous. Each of them can be subdivided into two parts. The multiplicity of subdivisions which results from this parcellation raises the problem of how to denominate them. The nomenclature we proposed [3], and I will follow here, is based on that used by Von Economo for the human brain [4]. The various motor areas are indicated with the letter F (frontal) and with Arabic numerals. The precentral cortex is referred to as F1, whereas the other areas have progressively higher numbers. A similar type of classification has been adopted for the visual cortex (V1, V2, etc.).

1. Area 4 (F1)

The cytoarchitectonic characteristics of area F1 are the following: a) Giant pyramidal cells are abundant. These cells tend to cluster together and to be organized in multiple cellular rows. b) There is a clear columnar organization. This organization extends from the superficial layers to the white matter. c) Cellular density decreases in the lower part of layer III. F1, as defined above, closely coincides with area FA of Von Bonin and Bailey. It extends, however, in rostral direction much less than area 4 of Brodmann [see 6].The validity of this anatomical parcellation has been recently confirmed by intracortical microstimulation studies [7,8].

2. Inferior Area 6

Inferior area 6 is formed by two areas: F4, located caudally, and F5, located rostrally. Both areas lie lateral to the spur of the arcuate sulcus. They are easily recognizable in material stained with cytochrome oxidase (CO) method. Their location corresponds to that of areas FBA and FBCm of Von Bonin and Bailey [5].

Intracortical microstimulation and single neuron studies showed that F4 is mostly related to arm proximal movements. Neck movements and facial movements are also represented there. In contrast, F5 is basically related to arm distal movements [9,10,11]. In both F4 and F5 mouth movements are represented in the lateralmost parts of the two areas.

F4 neurons respond to tactile and, very frequently, visual stimuli. They have 3-D visual receptive fields located in the animal's peripersonal space. Most of these fields are coded in body-centered coordinates [12]. Many F4 neurons fire during arm movements directed to a particular space sector independent of the initial movement starting point ("motor fields"). The visual receptive and the motor fields appear to be roughly congruent, but this point requires further studies.

F5 neurons become active during different types of distal arm movements (grasping, holding, tearing). Most of them are selective for specific types of grip such as precision grip, whole hand prehension, finger manipulation. A substantial amount of F5 neurons respond to visual stimuli. A first set of these neurons respond to object intrinsic properties in a way similar to that of manipulation parietal neurons described by Sakata and co-workers [13]. A second set fires when the monkey observes specific hand movements performed by the experimenter. Typically, the effective observed movements and the effective movements executed by the monkey are closely related [14].

These properties and the strict anatomical ties of inferior area 6 with inferior parietal lobule suggest that inferior area 6 plays an important role in transforming intrinsic and extrinsic object properties into the appropriate motor acts.

3. Superior Area 6

Similarly to inferior area 6 , superior area 6 is also formed by two areas: F2 and F7. F2 extends laterally until the spur of the arcuate sulcus and borders medially with F3 [6]. In antero-posterior extension it coincides with area 6aα of Vogt and Vogt [15]. F7 occupies the remaining, rostral part of superior area 6. It coincides with area 6aβ.

The following cytoarchitectural features characterize F2: a) A thin row of medium-size pyramids is well evident in the lowest part of layer III; b) Giant pyramidal cells are present but rare in layer Vb; c).There is a columnar cellular pattern similar to that of F1; d) Layer Va is rich in cells. The characterizing features of F7 are the following: a) A darkly stained, prominent layer V; b) A bipartite layer VI.

F2 and F7 differ in their cortico-cortical connections. F2 is strongly linked with the precentral cortex (F1) and has no connections with prefrontal cortex [16]. The opposite is true for F7. This area receives fibers from prefrontal areas but has no direct connections with F1. In the medial part of F7 there is the eye field described by Schlag and Schlag-Rey ["Supplementary eye field" 17,18].

In F2 many neurons fire when the monkey prepares to move ["set related" neurons; 19,20]. These neurons are intermixed with others whose activity is tied with movement execution. F2 appears to be somatotopically organized. "Leg neurons" are located medially, whilst "arm neurons" are located laterally [see 21]. In contrast to F2, F7 appears to be scarcely excitable with standard electrical stimulation [22]. The exception is its medial part from where neck movements and eye movements ("supplementary eye field") are represented

These properties and the strict link between superior area 6 and the superior parietal lobule suggest that F2 should play an important role in the proprioceptive control and proprioceptive representation of arm movements. At present there are too few data to speculate about a possible functional role of F7.

4. Mesial Area 6

The sector of area 6 lying on the mesial cortical surface has been classically described as the supplementary motor area (SMA). However, recent anatomical and functional data showed that the classical SMA is not a single entity, but results from two distinct areas: F3 or "SMA-proper" and F6 or "pre-SMA" [6,8,23,24].

The main cytoarchitectonic features of F3 and F6 are the following. F3: a) A columnar cell pattern similar to that of F1 and F2 is present, but limited to deep layers, b) The lower part of layer III is very rich in cells. It fuses with the adjacent equally dense Va, c) Giant pyramidal cells are occasionally present, but only in F3 caudal part. F6: a) Layer V is very dense and stands out with respect to the other layers, b) Layer Vb is absent, c) The superficial layers are homogeneous in terms of cell density.

Intracortical microstimulation of the mesial area 6 have shown that in F3 there is a complete, somatotopically organized, motor representation [8,25]. Hindlimb and forelimb fields form two oblique bands oriented in caudorostral and ventrodorsal directions. The hindlimb field is located caudally, whereas the forelimb field is located in the central part of the area. There is also a small orofacial field lying rostrally at the border with F6. The great majority of limb movements concerns the proximal joints. A clear separation between distal

and proximal movement representations was not found. With respect to F1 ("primary motor cortex"), F3 shows the following differential characteristics: a) A higher average excitability threshold; b) A greater percentage of movements involving two or more joints (complex movements); c) A much smaller distal movement representation.

Area F6 is electrically less excitable than F3. Furthermore, brisk phasic movements similar to those observed following excitation of primary motor cortex could be evoked from only 50% of the excitable sites. In the remaining sites the elicited movements were of "slow" type. They consisted in displacements of the forelimbs that mimicked postural adjustment or, even, natural movements of the animal. The somatotopic map in F6 is not complete. Essentially, it consists of an arm movement field.

As one can expect from stimulation experiments, the anatomical connections of F3 and F6 are markedly different [23,24]. F3 is strongly connected with F1 and with the posterior premotor areas (F2 and F4). It receives inputs from various sectors of superior parietal lobule and is directly connected with the spinal cord. In contrast, F6 is strongly connected with the prefrontal cortex and with the anterior premotor areas (F5 and F7). It lacks connections with F1 as well as with the spinal cord. It has surprisingly little connections with the postrolandic areas.

Investigations in which single neuron activity was recorded from the two areas confirmed their functional heterogeneity. The following differences were observed: a) Neurons responding to somatosensory stimuli are present in F3. They are rare in F6; b) The converse is true for visual neurons which are found almost exclusively in F6 c) Activity changes during the movement preparatory period are frequent in F6 but not in F3; d) Phasic, movement related activity is more frequent in F3 than in F6 and its onset in F3 is more frequently time-locked to the movement onset. [24, 25, 26]. Finally, in F6 there are neurons related to arm movements which show complex properties [27]. These neurons fire in large advance of the actual movements similarly to the set related neurons described by Wise et al.[19,20]. Their discharge, however, is influenced by the distance of the objects from the monkey and frequently shows rather complex excitation-inhibition patterns, depending upon whether the animal could or could not reach and grasp the presented objects [27].

It has been lengthy debated whether the SMA as defined by macroelectrode stimulation studies had "low-level" or "high-level" motor functions. [see 28]. The finding that the mesial area 6 is not a single entity allows one to clarify this issue. The properties of F3 reviewed above (e.g. cortico-spinal projections, link with motor cortex, high excitability, motor related activity) indicate that this area is responsible for the so called "low-level" motor properties of mesial area 6. It is likely that F3 plays an important role in the postural adjustment that accompany arm and other voluntary movements [see 29]. In contrast, F6 with its connections with prefrontal lobe and its highly complex neuronal properties is most likely the area responsible for the "high level" motor function postulated on the basis of slow potential [30] and blood flow changes [31] which occur before voluntary movements and during motor mental imagery.

5.Conclusion

The subdivision of the agranular frontal cortex into two main areas ("primary motor cortex", "supplementary motor cortex"), as still frequently used in neurology , is simplistic and far from the physiological reality. As in the sensory systems, where the periphery is represented over and over in the cerebral cortex, the movements of the body are controlled by a large number of distinct cortical areas. Although it may seem that seven motor areas is an exceedingly large number of cortical areas devoted to motor functions, there is little doubt that this number will increase in the near future. We already know, for example, that, in addition to the agranular motor areas described in this article, there are other motor areas related to body movements in anterior and, possibly, posterior cingulate cortex [6,8,32]. Our understanding of the roles of the different motor cortical areas in motor control is very limited. Experiments, however, like those reviewed above which correlate single neuron activity with monkey behavior, started to reveal some of the functional properties of the different areas. It is clear, for example, that the transformation of object-related visual information into reaching and grasping movements is mostly carried out in F4 and F5, whereas decisions about movements initiation are mediated primarily by mesial areas such as F6 and possibly the cingulate areas. These new physiological ideas about cortical motor organization should be a useful guideline for the study of human motor system with the new fascinating imaging and stimulation techniques.

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MUSCLE SPINDLE ACTIVITY DURING THE ELECTROMYOGRAPHIC SILENT PERIOD PRECEDING RAPID MUSCLE CONTRACTION

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When a rapid voluntary contraction is made upon a background of tonic muscular activation, a brief reduction in EMG activity occurs just before the phasic activation. This pre-movement occurrence is called the premotion silent period (PSP). Tonic activity before PSP is controlled by alpha-gamma coactivation. The present study was undertaken to determine whether muscle spindle activity would decrease during PSP concomitantly with a reduction in skeletomotor activity.

Five human subjects with informed consent lay prone on a bed with the ankle slightly plantar flexed and the foot secured to the steel plate. The subjects performed a rapid isometric plantar flexion after maintaining isometric plantar flexion of 10% maximum voluntary contraction for 2-3 sec. Muscle spindle afferent potentials were recorded using a tungsten microelectrode inserted percutaneously into motor fascicles of the tibial nerve at the popliteal fossa. Single unit afferent activities were identified as of muscle spindle origin by their typical responses to passive stretch and electrically induced twitch contraction of the receptor-bearing muscle. The EMG was recorded from the soleus, lateral gastrocnemius, and medial gastrocnemius using surface electrodes. The PSP in the receptor-bearing muscle EMG was detected. The interspike interval that overlapped with PSP was compared with interspike intervals during tonic isometric contraction.

Eight single units of muscle spindle afferents were recorded from tibial nerve fascicles innervating the soleus (five), lateral gastrocnemius (one), medial gastrocnemius (one), and soleus or lateral gastrocnemius (one). A weak voluntary contraction was sufficient to increase the discharge in six spindle afferents. Activity of two other afferents was slightly decreased during isometric contraction.

The units that exhibited an increased discharge during voluntary contraction showed absence of impulse during PSP or prolongation of the interspike interval overlapping with PSP. On the other hand, the activity of units that exhibited a decreased discharge during voluntary contraction did not decrease during PSP. Assuming that the increase in spindle afferent activity during weak voluntary contraction was due to beta or gamma fusimotor drive, deceleration of afferent activity may be associated with a decrease in the fusimotor drive during PSP. The descending commands related to initiation of PSP may induce inhibition of extrafusal and intrafusal fibers almost simultaneously.

CORTICAL MOTOR REPRESENTATION AREAS OF UPPER AND LOWER EXTREMITY MUSCLES IN HEALTHY SUBJECTS AND HEMIPARETIC PATIENTS

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Figure-8-shaped coils in magneto-electrical stimulation (MES) allow a more focal stimulation of human motor cortex than conventional coils, hence motor representation areas (MRA) can be studied in more detail (1). In patients after amputation motor evoked potential (MEP) amplitudes are increased and MRAs are enlarged for muscles ipsilateral and proximal to the stump (2,3), consistent with the hypothesis of reorganisation of motor pathways in children and adults. We addressed this question with regard to patients after unilateral cortical and subcortical ischaemic lesions and compared data with the unaffected side and with results obtained from normals.

Thenar representation areas were investigated in eight healthy subjects and five hemiparetic stroke patients. More detailed studies were performed in three normals and two hemiparetic patients, in whom a total of 20 (normals) or 8 (hemiparetic patients) muscle representation areas were mapped. The position of the magnetic coil was systematically varied over the scalp using a flexible cap with the 10-20 EEG system marked on it as a reference grid. Surface EMGs of upper and lower limb muscles were recorded in a number of sessions. Intensity of stimulation was 5% above motor threshold determined separately for each muscle. The orientation of the coil was held constant at 45 deg from the midline.

In healthy subjects the overall distribution conforms roughly to the known motor homunculus. The individual maps however show marked overlap and the exact location of the maps relative to each other varies. Furthermore there is no strict symmetry between right and left hemispheres regarding motor threshold and size of motor maps. The size of motor representation areas increases with higher stimulation intensities and with preinnervation of the muscles.

fig 1:

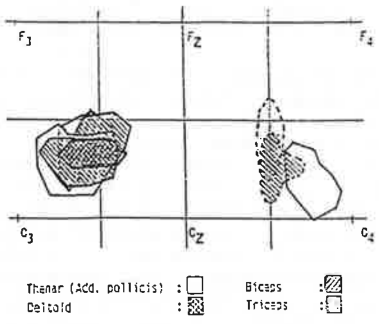
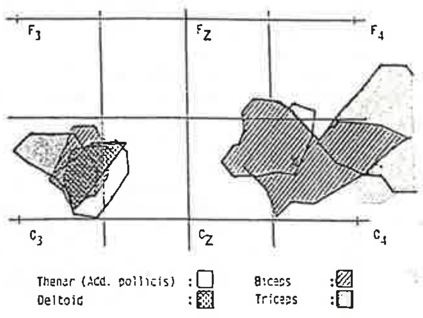


fig 2:



In hemiparetic patients different patterns of alteration of muscle representation areas can be observed: In a patient 4 months after a small right hemispheric subcortical lesion resulting in slight weakness of his left arm, motor thresholds were generally elevated on the affected side (fig.1). However the topographic arrangement of MRAs appeared to be normal in comparison to the non-affected side. In a patient 38 months after a large right hemispheric cortical and subcortical lesion with a plegic left arm (except for very weak voluntary activity of biceps and triceps which developed after intensive physiotherapy), a lateral shift and enlargement of MRAs of those two muscles could be shown (fig.2). Thresholds were not elevated compared to the unaffected side, latencies were not significantly prolonged and surface EMG showed no sign of preinnervation. Similar to results in patients after amputations, enlargement and topographical shift of muscle representation areas in stroke patients may indicate the presence of plasticity in adult human motor pathways.

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ORGANIZATION OF CEREBELLAR CONTROL OF FORELIMB MOVEMENTS VIA RUBROSPINAL AND CORTICOSPINAL TRACTS

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The concepts of the functional organization of the cerebellum have been greatly refined during the last twenty years. In each side of the cerebellar anterior lobe of the cat, ten sagittal zones have been identified. Each zone receives climbing fibre afferents from a circumscribed part of the inferior olive and in turn projects to a specific deep cerebellar or vestibular nucleus. In this way, a sagittal zone is functionally coupled to a specific set of spino-olivary pathways and controls specific descending motor systems.

The Y, C3 and C1 zones in the pars intermedia project to nucleus interpositus anterior, which controls the rubrospinal and corticospinal tracts through projections to the red nucleus and, via thalamus, to the sensorimotor cortex.

We have recently investigated the functional organization of the spino-olivary pathway and the olivocerebellar projection of this cerebellar control system. Climbing fibres to the forelimb areas of the Y, C3 and C1 zones received nociceptive and tactile cutaneous input via the postsynaptic dorsal column pathway. In the C3 zone, the cutaneous receptive fields of the climbing fibres could be divided into eight classes, with totally thirty subclasses, according to their spatial characteristics. Generally, the proximal borders of the receptive fields were located close to joints. The area from which maximal responses were evoked was often laterally or distally located on the forelimb. These characteristics suggested that the receptive fields were related to movements. Climbing fibres with receptive fields of the same subclass terminated within narrow, sagittally oriented cortical strips, thus forming microzones. Each receptive field class had its specific location in the C3 zone. It was hypothesized that a microzone receives climbing fibre mediated information about the activity in spinal reflex circuits acting on a single muscle, or a small group of synergistic muscles, in the forelimb. The microzone would, in turn, control the corresponding pool of motoneurones.

The latter part of this hypothesis is dependent on whether there are separate termination areas for microzones with different climbing fibre receptive fields in nucleus interpositus anterior. Therefore, in the present study, the organization of the cerebellar cortical projection to nucleus interpositus anterior was investigated. It was shown that peripheral stimulation evoked positive field potentials in the nucleus and that these potentials reflected the inhibitory synaptic action of climbing fibre activated Purkinje cells on nuclear neurones. The cutaneous receptive fields of the positive field potentials were mapped throughout the nucleus. These receptive fields had similar spatial characteristics to those described for single climbing fibres in the C3 zone. The results strongly suggest that microzones with climbing fibre receptive fields within a class terminate in a single area of the nucleus. Different classes had separate termination areas. In many cases, this was probably also true at the subclass level.

A comparison between the corticonuclear input to different sites in nucleus interpositus anterior and the movements elicited by electrical stimulation of these sites was also made. The results suggest that different microzones are associated with the control of specific movements in the forelimb. For example, stimulation at nuclear sites with a cutaneous receptive field on the ventral side of the paw, evoked dorsal flexion of the paw, whereas stimulation at sites with a receptive field on the dorsum of the paw evoked palmar flexion.

We thus propose that there is a modular organization of the cerebellar circuits controlling the rubrospinal and corticospinal tracts. Each module would consist of a group of microzones in the Y, C3 and C1 zones with identical climbing fibre input projecting to a specific set of nuclear neurones. The modules seem to control specific movements and the climbing fibre mediated information is probably related to the movement controlled.

FINE MOTOR CONTROL OF ABDUCTOR POLLICIS LONGUS AND TIMING OF WRITING IN HEALTHY ACTIVE SUBJECTS OF DIFFERENT AGES. AN EMG EVALUATIVE STUDY

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Electromyography and electrospeedometry of normal fine sensori-motor ability of a useful action may be an important approach to interpret the level of inhability usually seen in motoneuron diseases. This study aims to demonstrate different levels of fine motor control (FMC) ability achieved in healthy active subjects of different ages. The quality of this isolation (FMC with the aid of EMG biofeedback) and its variation are taken into account in order to find out to what extend they could be affected by the level of tests difficulty, types of contraction, speed regularity or by the age of the subjects. One group of young adults aged from 20 to 40 years old (n=13) and another of active senior citizens aged from 63 to 75 years old (n=13) participate. They are stabilized in a semi-reclined position at a multi-orthotic system to facilitate the suppression of body muscular activity and to allow only the right thumb-index voluntary motions. With indwelling fine wire electrodes connected to an EMG apparatus, myoelectrical responses of the Abductor pollicis longus (APL) muscle are analysed. The FMC of APL is tested while the thumb and index hold a pencil at an upward writing posture (S1) and are free to self write. It is also tested at the downward (S2) and at the upward (S3) postures. An electrogoniometric (EGM) device is used to record simultaneously the angular displacements during carpo-metacarpal abduction-extension and adduction-flexion of the thumb, required when writting a line downward and upward slowly (SD, SU) and rapidly (FD, FU). Observations lead to establish percentage-level of FMC achievements. So the interpreted levels of FMC are based on : A perfect isolation of a chosen SMU (100%); an inhibition of activity, a predominance of the chosen SMU, or transfer of SMU activity (80%); and an evoked light activity (60%).

Group / Test		S1	SD	S2	SU	S3	FD	FU	$\bar{X}$
Y	$\bar{X}$	95	80	83	85	71	70	73	81,2
	Sd	8	16	15	14	15	13	16	8,1
S	$\bar{X}$	92	75	81	78	73	72	74	80,1
	Sd	10	18	12	16	15	15	16	9,6

Table 1. Mean and Sd % of fine motor control success related to posture and writing motions tested in young adult (Y) and senior citizen (S) subjects.

Group / Test		SD	SU	FD	FU
Y	speed	0,00114	0,00158	0,0324	0,0414
S	speed	0,00183	0,00217	0,0254	0,0309

Table 2. Effect of FMC $\bar{X}$ % levels of success on the speed (m/s) of slow and rapid writing motions used by young adult (Y) and senior citizen (S) subjects.

Each test has its own level of success and degree of difficulty and, the perfect FMC in APL, can be achieved in any of the tests given. The means of success of FMC of these groups (Table 1), being Y=81,2 and S=80,1%, demonstrate the feasible degree of the testing procedure and the similarities of FMC levels of achievements in both groups. All subjects succeed to control inhibition at will in APL during maintained writing postures (S1, S2 and S3) and, analyses of the means and the frequency distribution of subjects related to levels of success of each static and dynamic test, show increase FMC difficulty in fast continuous motions (Table 1). Voluntary writing motion without assistance has a tendency to be faster during eccentric (FU) than in concentric (FD) contraction of the APL muscle in both groups (Table 2). In rapid motion under FMC, the lower speed performed by active senior citizens may point out a sigh of the locomotor system maturity (Table 2). Results, including the testing of FMC in old age in a useful activity, give further additions to results obtained in the only young adult group (1) and to classical EMG evaluations of FMC (2) base on the effect of primary muscular function.

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A METHOD FOR CORRELATING MUSCLE KINEMATICS AND EMG OUTPUT DURING HUMAN LOCOMOTION

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The levels of muscle contraction and the current muscle kinematics interact in determining the mechanical outcome of voluntary motor commands. Monitoring these parameters and correlating them during natural motor behaviours is expected to provide deeper insight into the strategies adopted by the CNS for muscle recruitment, either in normals or in patients with motor disorders. The present method was designed to quantitatively assess the relationship between muscle length (or velocity) and amplitude of EMG activity during unconstrained locomotor tasks.

Eleven muscles acting at the hip, knee and ankle joint were modelized according to modified versions of the algorithms proposed by Frigo and Pedotti (1987). Each muscle is represented by a line (linear segment or linear + circular segment) joining the origin to the insertion point. The length of the resulting segment is a function of anthropometric parameters and joint angles, both of which are used as inputs to the model. Anatomical parameters are measured in the individual subject as distances between relevant bony points. Joint angles are obtained by automatic motion analysis (ELITE system, sampling freq. 50 Hz). Output of the models are the instantaneous muscle lengths during locomotion, with reference to the resting values measured while standing. Muscle velocity can be also computed by first derivative.

EMGs are recorded by surface preamplified electrodes at 500 Hz, and bandpass filtered between 20 and 200 Hz. The mean value of the rectified EMG over 50 ms centered time windows, normalized to the maximum locomotor output, is taken as a measure of the instantaneous EMG amplitude. Down-sampling (500 to 50 Hz) is performed to match resolution with the kinematic signal.

Correlation between myoelectric and kinematic variables is obtained by plotting EMG amplitude vs muscle length (or velocity), for the overall step cycle or for specific time windows selected by the operator. Delays between kinematic and EMG values can be modified in order to find out the optimal correlation, which is finally described in terms of regression coefficient and slope.

Preliminary testing of the method in normal subjects is currently performed in our gait laboratory, as a prerequisite for the development of non invasive tests to be applied in patients with motor disturbances (e.g. spasticity).

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QUANTITATIVE EVALUATION OF NORMAL MUSCLE TONE

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Muscle tone can be clinically defined as the resistance felt to externally imposed movements. When a limb is passively moved from a rest position some factors occur to contrast such perturbation. These factors include the intrinsic passive properties of the tissues (intrinsic stiffness) and the neural components. The present study has been performed to recognize some worthwhile parameters to define an objective measure of muscle tone in normals because of the scant reliability of clinical evaluation scales.

Forty-eight normal subjects (24 males and 24 females, aged from 21 to 70) were evaluated: they were comfortably seated by the side of a servo-controlled torque motor: the forearm held in an adjustable support. A computerized equipment sent signals to the DC-torque motor connected to a tachometer, a position and torque transducers. EMG activity to imposed wrist movements in both flexor and extensor carpi radialis was simultaneously recorded. Subjects were asked to relax completely throughout the examination. Each subject was tested three times during a day. Each session was divided in 4 steps:1) research of the stretch reflex threshold (SRT), that is the minimum speed of wrist extension movement able to induce EMG reflex activity in wrist flexor muscles in at least 50% of displacements; 2) the measurement of the "intrinsic stiffness index" (ISI) at a very slow speed (10 deg/s); 3) the measurement of the total stiffness index (TSI), expressed by the resistance corresponding to both visco-elastic properties and neural factors with displacements at 500 deg/s; 4) the daily intraindividual ISI and TSI variability: for this purpose 8 out the 48 subjects were retested 8 times in one day. Statistical analysis was based on ANOVA, linear regression and Student's t test.

SRT. No stretch reflex was found in the 46% of the subjects: in the remaining, SRT ranged from 60 to 500 °/s; only in 5% SRT was ≤ 100°/s. No statistical difference was found neither for age nor for sex.

ISI and *TSI*. ISI was significantly higher in males ($p<0.0001$). No statistical differences in ISI and TSI was found among the three sessions. No statistical differences were found between TSI in subjects presenting stretch reflexes and TSI in subjects without stretch reflexes. In 29% of the subjects shortening reflexes could be detected, but their presence did not influence statistically TSI. The most stable parameter is surely ISI: in fact the variability at 10°/s is very low (SD: 0.1-0.7). On the contrary at 500°/s the intra and intertrials variability increases particularly in those subjects presenting several stretch reflexes (SD: 0.2-2.8).

Therefore the three parameters above described, SRT, ISI and TSI seem to well characterize the range of the so called "normal" muscle tone; they are fairly reliable and likely useful in the quantitative assessment of muscle tone, particularly in the light of some preliminary observations we dealt with a little sample of hemiplegics. *SRT* shows a wide range of variations in normals; on the contrary the spastic patients gather in a quite narrow range of speeds (from 30° to 100°/s), in agreement with other AA (Gottlieb et al, 1978; Powers et al, 1989). *ISI* shows very low intrasession and daily variability, that means good reproducibility.The only factor influencing this parameter is gender, probably because of the mass effect of muscles acting on the wrist and its correlation with arm volume (Wiegener et al, 1985). Besides ISI shows a marked increase in hemiplegics even in those without severe spasticity; this is in agreement with several literature data in favour of changing in muscle properties (Dietz et al, 1981). On the contrary *TSI* shows a high intrasession variability, particularly in those subjects where several stretch reflexes are evoked. Anyway the comparison of TSI in pathological and normal subjects shows a marked difference in the values so that it's very easy to discriminate between the two populations.

In conclusion it is quite difficult to consider the normal subject as a simple baseline from which to assess the responses of spastic joints because of the wide range of variety of different responses. The confirmation of a more stereotyped behaviour of spastic joints make us to further develop this procedure to a wider pathologic group expecially when the purpose is to check the effects of pharmacological or physical therapy.

STIMULATION LEVEL DEPENDENT LENGTH FORCE CHARACTERISTICS OF RAT GASTROCNEMIUS MUSCLE

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Length force characteristics of maximally and submaximally stimulated rat medial gastrocnemius muscle were studied at different levels of excitation. Geometry of the active muscle was recorded by photography. The geometry was assessed by measuring lengths of the muscle, proximal aponeurosis and distal fibre on projected images. Maximal stimulation was obtained by stimulating the nerve innervating the muscle with supramaximal current. Submaximal stimulation was obtained by using 600 Hz blocking stimuli in addition to the 100 Hz excitation stimuli. Using this century old technique (Wedenski, 1884), which was reintroduced by Solomonow and coworkers (1983 to present), motor units are derecruited in a sequence according to size.

Length force characteristics during progressively developing blocks showed muscle optimum lengths occurring at increasing lengths. A significant correlation ($r = -0.9$) was found between decrease of optimum force as a consequence of the block and the length shift for optimum length.

For the length range in which optimum lengths were found muscle geometry could not be distinguished on the basis of stimulation protocol used. However at short muscle lengths, differences of geometry were encountered: Fibre length was shorter for maximally excited muscle than for block excitation. This can be understood because elastic properties allow aponeurosis length to be increased as more force is exerted. It should be noted that at high muscle lengths considerable decreases of muscle force, obtained as a consequence of block stimulation, were not accompanied by substantial length changes of the aponeurosis. Unless local circumstances would prevent expression of local length changes of the aponeurosis in its total length, this would indicate that release of elastic energy from the aponeuroses would be limited during force decrease at these muscle lengths.

It is also concluded that any possible differences of fibre length and or shortening velocity could not explain the shifts of muscle optimum lengths of up to 14% .

Two major factors for explaining the stimulation level dependent length force characteristics were indicated but could not be distinguished unequivocally as yet:

A. The occurrence of a distribution of fibre optimum lengths with respect to muscle optimum length in such a way that fibres of large motor units reach their optimum at shorter muscle lengths and fibres of motor units of decreasing size reach their optimum length at progressively higher muscle lengths.

B. If the blocking technique used involved slowing of firing of motor units before derecruitment, a dependence of optimum length on firing frequency could have played a major role in the stimulation level dependent shifts of this length.

These findings have a functional significance which should be taken into account in muscle modelling as well as work involving Functional Electro Stimulation (FES).

COMPUTERIZED SYSTEM FOR SKIN SYMPATHETIC REFLEXES MEASUREMENT

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Skin sympathetic reflexes need to be recorded from more than one site over the hand and foot if the quantitative measure is to be of significance. For this purpose the instrument must to be sensitive enough to detect variations of a fractional of a percent of skin impedance.
Method: To support sympathetic investigation we have developed a system which combine the functionality of a battery of tests with the flexibility of a computer analysis environment based on a dual-channel self-balancing impedance reactometer. Measurement of skin sympathetic reflexes in 33 healthy subjects, aged under 50 yrs. was performed by noninvasive monitoring transient change in the skin impedance both in phase (resistive) and in quadrature (capacitive) on rest and after a stimulus. The procedure requires a placement of a pair of surface electrodes on the skin and application upon the subject an endogenous/ exogenous stimulus. 12 bit data resolution specially designed data acquisition card is used which converts the signals into digital form and transfers this to the computer IBM PC XT/AT data bus directly. The software developed for the system written in Pascal, displays the data on standard PC colour monitor in real time 1, 2 or 4 traces at a time.
Results: An autonomic response is an extremely complex state of the subject and can be characterized by many parameters which may have both temporal and spatial dependencies. The sympathetic response signal to be recorded is superimposed on a background signal which may be orders of magnitude larger but constant during the recording time. The insertion of the self-balancing circuit into the detection electronics, the background signal is compensated, the sympathetic signal is not affected and could be detected with the highest sensitivity. Transient skin sympathetic response was 2.18 +/- 0.31 %/s at hand, 1.41 +/- 0.42 %/s at foot and 0.27 +/- 0.14 %/s in other site of the body. The potential of the system is currently investigated for on line diabetic foot analysis.

EMG ACTIVITIES OF THE ABDUCTORES OF THE THUMB AND THE EXTENSOR CARPI ULNARIS

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The abductor pollicis longus muscle (APL) can be divided into a distal superficial and a proximal deep division (Oudenaarde,1991,a). Each division is innervated separately by branches of the radial nerve. In passive movements of the thumb, only the tendons of the superficial division show an excursion in all directions, regardless the position of the wrist (Oudenaarde,1991,b). During passive movements of the wrist both tendons show an simultaneous excursion to the same amount. These are small in pronation and in supination of the forearm but much larger in the mid-position between them. The same results are obtained for the extensor carpi ulnaris muscle (ECU) in movements of the carpal joint.

Aim of this study was to investigate: The importance of the APL and ECU as stabilizers of the carpal joint and CMC I; a task dependent functioning of the divisions of the APL; similarities and differences in EMG activities of the abductor pollicis brevis muscle (APB) and the deep division of the APL for movements of the thumb and the wrist.

In seven subjects, intra-muscular wire electrodes were used for the experiments. The thumb was brought in five positions (Terminology for Hand Surgery,1972): palmar abduction, opposition, radial abduction, adduction, reposition. Isometric as well as isotonic contractions were recorded with a constant force of 5 Newton. For movements of the carpal joint a constant force of 20 Newton was used in dorsopalmarflexion of the carpal joint, with the forearm successively in the position of supination, pronation and midposition In movements of the carpal joint the thumb was in a closely relaxed position to the indexfinger. The EMG signals of each subject were scaled in order to compare EMG activity for different subjects.

The muscles studied are active in all the positions of the thumb and the wrist. The divisions of the APL have a different task dependent activation. The deep division of the APL and the ECU show more EMG activity in extension of the wrist than in flexion. The APL and the ECU are more active in movements of the thumb, when the carpal joint is fixed. In situations when the wrist is free to move these muscles show less activity. The muscles show more EMG activity in isotonic than in isometric contractions.

The main conclusion of this study is that the deep division of the APL and the ECU are very important for stabilisation of the wrist. Specially the deep division of the APL act in movements of the thumb as a stabilizer of the wrist, for it has no insertion into the thumb. Perhaps the deep division of the APL supports the APB. This might be interesting in cases of peripheral nerves injuries.

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CUTANEOMUSCULAR REFLEXES IN DISTAL AND PROXIMAL ARM MUSCLES

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Low threshold cutaneous reflexes in man have been extensively studied in hand muscles giving non-painful electrical stimuli to the digital nerves or to the superficial radial nerve at the wrist.

Little is known about cutaneous reflexes in forearm and arm muscles.

We recorded muscular responses from voluntarily contracting biceps brachii, deltoideus, flexor carpi radialis, flexor pollicis longus, abductor pollicis brevis, abductor digiti minimi muscles with surface electrodes in five normal subjects. Median and ulnar nerves were stimulated at wrist at near-motor threshold intensity; electric pulses were delivered to the distal phalanges of one or more fingers (non-painful stimulus intensity of 2-3 times sensory thresold) to stimulate selectively tactile afferents. Raw and fully rectified EMG signals were acquired and averaged in the 100 or 200 msec following nerve stimulation.

We obtained short-latency and long-latency cutaneomuscular responses in distal and proximal upper limb muscles.

This shows that the distal cutaneous afferents could have a reflex action on multiple distal and proximal muscles in ipsilateral arm.

PRE-MOVEMENT DECREMENT OF MOTOR EVOKED POTENTIALS IN MAN DURING PREMOTION ELECTROMYOGRAPHIC SILENT PERIOD

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It has been observed that tonic EMG activities of agonist muscles become silent before an agonist burst of a rapid ballistic movement. This premovement event was called premotion silent period (PSP). At the same time, premovement facilitation of motor evoked potentials (MEPs) have been reported from transcranial stimulation (TCS) studies. The present study examines the effect of PSP on MEPs elicited preceding the movement.

Five healthy volunteers were tested after having obtained their informed consents. The subjects were trained to press a foot plate as rapidly as possible in response to a visual cue following 15% of maximum voluntary background contraction for 5 to 10 sec. Surface EMGs were recorded from soleus muscle and tibialis anterior muscle. To superimpose MEPs on PSP, the magnetic stimulus was delivered randomly at 50 to 250 ms after the visual cue.

Examples of MEPs obtained from the soleus are shown in Fig. 1. Upper trace shows a control MEPs recorded during the background voluntary contraction. Middle trace indicates a MEPs elicited during PSP. Lower trace shows a MEPs recorded from a trial without PSP. In 37% of all trials, MEPs were successfully superimposed on PSP. Pooled data of the five subjects, expressed as percentage of the control MEPs, are plotted against time interval between onset of MEPs and EMG burst (Fig.2). The MEPs off PSP (solid circles) was facilitated in 100 ms before the onset of EMG burst, as reported by previous studies.

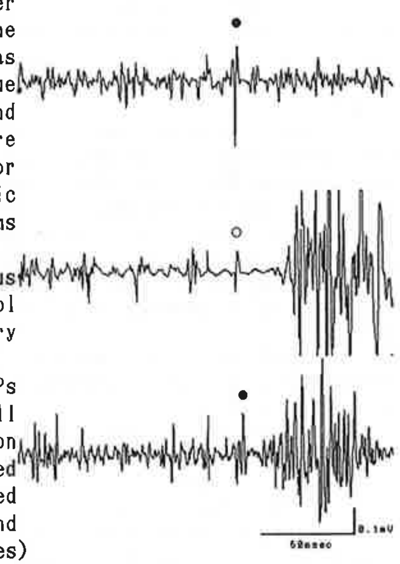

Fig. 1

In contrast, the MEPs on PSP (open circle) was not so facilitated as that off PSP. Mean amplitude of MEP on PSP was significantly smaller than that off PSP in 100 to 10 ms before the EMG burst. These results have shown that the inhibitory effect of PSP was present on the corticospinal activity preceding the voluntary movement.

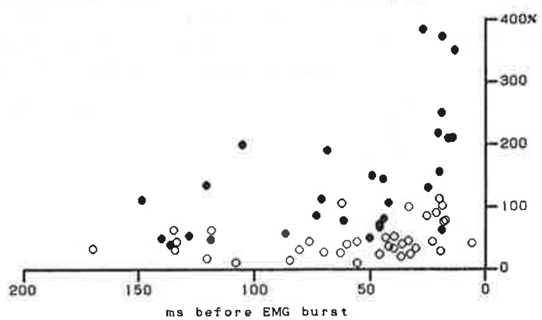

Fig. 2

In addition, EMG responses to magnetic TCS were recorded from the tibialis anterior together with the soleus. There was significant positive relationship between the amplitude of the antagonist response and that of the agonist. The results suggested that the inhibitory effect of PSP was not organized by the reciprocal inhibition system in the spinal cord.

VARIATION OF H REFLEX EXCITABILITY FOLLOWING VOLUNTARY MUSCLE CONTRACTION OF DIFFERENT DURATION

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A significant inhibition of H reflex at the Soleus muscle immediatly after a voluntary muscle contraction has been reported by numerous Authors.

This phenomenon has been ascribed to the presynaptic inhibition of homonim Ia afferent fiber(Schieppati,Enoka, Moore et Al.) and it has been refered to as Release Associated Inhibition phenomenon (R.A.I.).

The purpose of the present study was to determine if this phenomenon was influenced by the duration of voluntary contraction.

The H reflex at Soleus muscle was measured,at rest in six voluntary healthy subjects(age range 19 to 32 years) in prone position on the isokinetic dynamometer with their feet stabilized by straps at right angles.

Two sets of five measurements of H reflex were taken after a short(10") and long contraction in plantar flexion. All measurements were performed at 60% during the first ten seconds immediatly following release at 1", 2", 3", 6",10".

During the first 3" after the short period of contraction the inhibition of H reflex was statistically significant.

However the inhibition of H reflex was statisti= cally significant for ten sec. when the contraction pe= riod was longer.

The data of this study conferme the presence of inhibition after release, reported in previous literature and suggest a relashionship between the duration of voluntary contraction and the persistence of the H reflex inhibition after release. The mechanism of this phenomenon is unknown but it could be caused by an increase of presynaptic inhibition after long contraction or by a persistent inhibition after release from antagonist muscle.

SIMULTANEOUS RECORDING OF EMG AND 31P NMR SPECTROSCOPY

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Introduction

The physiological mechanisms behind muscle fatigue are not well understood. Important information has been obtained by studying electromyography (EMG) and nuclear magnetic resonance (NMR) spectroscopy, but only in few investigations the two methods have been used in parallel. The aim of this study was to develop a method by which EMG and NMR spectroscopy could be performed simultaneously.

Methodology

All measurements were performed in a whole body 1.5 Tesla NMR-scanner. The subject had the left foot strapped to a calf muscle non-magnetic ergometer (Fig. 1.). The EMG was recorded by bipolar surface electrodes placed at an inter-distance of 25 mm along the anterior tibial muscle axis. A surface coil was strapped to the muscle next to the EMG electrodes. Disturbances between the NMR and EMG were primarily avoided by attaching the electrodes and the leads firmly, by using shielded leads of equal lengths and by ensuring a high common mode rejection in the electrode-amplifier circuit. In addition the EMG amplifier was situated inside the shielded cabin surrounding the whole body scanner.

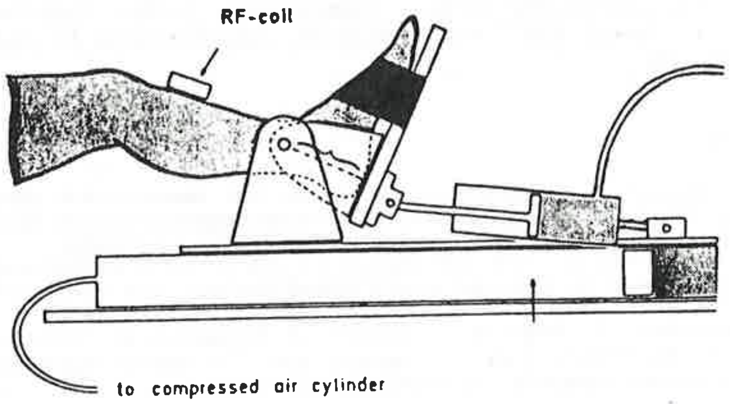

Fig. 1. Set up for simultaneous recording of EMG and NMR parameters. The ergometer was non-magnetic, it was individually adjusted, so that the axis of rotation of the ergometer pedal was aligned with the axis of rotation of the ankle joint. The RF coil was strapped to the muscle next to the EMG electrodes.

In 6 controls simultaneous measurements of EMG and ^{31}P NMR spectroscopy were performed in the scanner during isometric submaximal voluntary contraction at 25-50% MVC until exhaustion. The fatiguing contractions endured 5 to 13 min.

Concentrations of phosphocreatine (PCr), inorganic phosphate (P_i) and pH were analysed and correlated to changes in RMS and Median Frequency of EMG.

Results and discussion

The changes found in EMG and NMR parameters from the present experiments were similar to those reported from investigations, where the parameters were recorded separately. They also agree with findings by Miller et al (1987, 1988), who performed simultaneous measurements of EMG and NMR in the recovery period of fatiguing exercise in a small hand muscle.

In our study RMS of EMG remained almost constant until the normalized PCr concentration declined below 0.6-0.7 and pH declined below 6.75-6.85; then RMS increased by a factor of 2-3. Median Frequency decreased linearly with time and was linearly correlated to the decline in pH. The increase in EMG amplitude in a submaximal muscle contraction at constant force, may be explained by an inefficiency due to metabolic changes or to reduced exitation contraction coupling, but could also be caused by recruitment of high threshold motor units.

Conclusion

The technical step foreward was, that it was possible to record EMG and NMR simultaneously without interference, further experiments are necessary to reveal EMG and metabolic dependencies in normal and diseased muscle.

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The study was performed by *P. Vestergaard-Poulsen, *C. Thomsen, **T. Sinkjær, *M. Stubgaard, **A. Rosenfalck and *O. Henriksen.
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ROBUST ESTIMATOR FOR MEASURING MUSCLE CONDUCTION VELOCITY: PARAMETER DEFINITION

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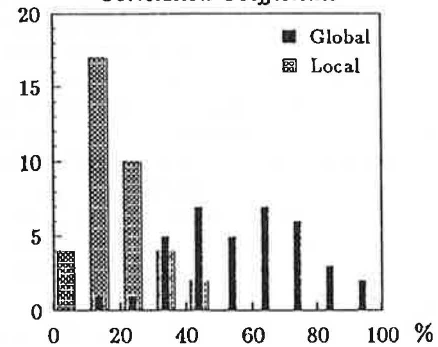
Several algorithms aimed to evaluating muscle fiber conduction velocity (CV) have been proposed in literature. Generally, CV is estimated by measuring the delay τ between myoelectric signals collected from different sites along the active muscle fibers. McGill and Dorfman (1984) proposed an algorithm for high resolution alignment of signals in the frequency domain. This algorithm uses the Newton's method to minimize the mean square error between the the Fourier transforms of the signals. The abscissa of the global minimum of the error function is an estimate of τ. When the interelectrode distance d is known, CV may be expressed as $CV = d/\tau$. In the last six years we used extensively this algorithm to evaluate CV of healthy subjects. Unfortunately, problems arose when applying this method to subjects affected by different forms of muscular dystrophy. In several cases the Newton's method failed to converge to the global minimum of the error function, thus causing a bad CV estimate. These estimation errors were always related to low values of the correlation coefficient (CC) between the aligned versions of the signals. To overcome this problem we modified the algorithm as follows: First CV is estimated by the Newton's method and the CC of the aligned signals is computed; If CC is below a given value α, we search again the minimum of the error function by means of a different method (Brent's algorithm), that assures the convergence to the global minimum; To increase the accuracy of the estimate the Newton's method is then used, starting the search from the abscissa of the minimum found by the Brent's method. The Brent's method assures that the difference between the real value of the minimum and its estimate is lower than a tolerance t fixed by the user. The aim of this work is: a) to describe the new algorithm and its performances, and b) to describe the methodology used to define the numerical values of α and t.

We analyzed 45 signals collected during voluntary contractions of the Tibialis Anterior muscle of patients affected by different forms of muscular dystrophy. Every contraction lasted 30 s and was divided into 30 epochs lasting 1 s each. In 8 cases out of 45 (18%) we found epochs in which the Newton's method failed to find the global minumum. For each contraction we found the highest CC value associated to the relative minimum closest to the global one, and the CC value associated with the global minimum. The results are depicted by the hystograms below.

By observing the histograms of the two populations it is evident that the correlation coefficients of local minima were always below 50%. Consequently, we decided to set $\alpha = 75\%$. The value of t was defined by analyzing the distribution of the differences between values of global and local minima. To correctly identify the global minimum the value of t must be smaller than the smallest diffference observed. Since the differences ranged from 1E-5 to 4E-3 ($\bar{\Delta}$ =2E-3), we decided to set t = 1E-6. By using these values for α and t, the new version of the algorithm could find the correct abscissa of the minimum of the error function in all those cases in which the original algorithm failed.

Correlation Coefficients



The financial support of Telethon - Italy to the project *Evaluation Protocols for muscular dystrophies: a multicentric approach* is gratefully acknowledged.

"WAVELET" ANALYSIS OF ELECTRICALLY EVOKED MYOELECTRIC SIGNALS

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During sustained electrically elicited muscle contractions the evoked response (M-wave) undergoes changes of amplitude, width and shape, that are associated to physiological modifications of the neuromuscular system. In particular, the M-wave expands in time and shows an increase in amplitude, during an initial part of the contraction, followed by a decrease during the last part of the contraction. Some modifications are attributed to a progressive decrease of muscle fiber conduction velocity, others are not completely understood. To provide greater insight into this issue a quantitative description of the time dependent features of the response is required.

One possible approach to the quantitative description of these changes is provided by the expansion in a series of orthonormal basis functions having limited support and shapes similar to that of the M-wave. One such series is based on the Gaussian function and on its derivatives. This series has been proposed by G. Sandri and C. Konstantopoulos (1), is referred to as the "Hermite-Rodriguez (HR) wavelet expansion" and has some interesting features that make it suitable for describing M-waves because of its limited time support, good approximation with 10-15 terms, possibility of optimization using the width of the Gaussian as a free parameter. An M-wave can be suitably expanded into (and interpolated with) a sum of properly weighted HR wavelets from order 0 to N and width parameter λ as indicated in the following equations:

$$f(t) \simeq \sum_{i=0}^{N} F_i^{\lambda} W_i^{\lambda}(t) = \sum_{i=0}^{N} F_i^{\lambda} \frac{H_i(t/\lambda)}{\sqrt{2^i i!}} \frac{1}{\sqrt{\pi}\lambda} e^{-t^2/\lambda^2} \quad where \quad F_i^{\lambda} = \int_a^b \frac{H_i(t/\lambda)}{\sqrt{2^i i!}} f(t) dt$$

where $H_i(t/\lambda)$ is a scaled version of the Hermite polynomial of order i and F_i^{λ} are the coefficients of the expansion. The identification of an "optimal" value λ_o, that allows convergence with a minimal number of terms, is of interest and is being investigated.

It can be shown that, given a function $f(x)$ and a function $g(x) = hf(kx)$, the respective HR wavelet expansions are related. In particular, if λ_g and λ_f are chosen so that it is $\lambda_g/\lambda_f = k$, then it is $G_i = hkF_i$ where F_i and G_i are the coefficients of the expansion of $f(x)$ and those of $g(x)$ for the two respective λ values. If $g(x) \simeq hf(kx)$, then it will be $G_i \simeq hkF_i$.

This property can be used to estimate the unknown values of h and k by finding the ratio λ_g/λ_f that maximizes the correlation coefficient C between the two vectors G and F. This ratio is an estimate of k. The regression coefficient R of the LMS linear fit between the G_i and F_i values is an estimate of hk. C_{max} or $1 - C_{max}$ may be considered as indices of shape change.

These relations hold promise for the estimation of k and h in the presence of noise or when the function $g(x)$ is a somewhat distorted version of $hf(kx)$. The two scale factors k and h could be estimated from the two wavelet expansions without using one of the functions as a reference for the other but, rather, by comparing the "main features" of one function with those of the other, where the "main features" are identified by the coefficients of the expansions and by the respective λ values. This possibility may provide a tool for separation and quantification of changes of amplitude scale, time scale and shape of the M-wave during a sustained contraction. This technique may also be useful for processing other types of evoked potentials, such as in EEG analysis.

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EMG OF SHOULDER MUSCLES IN DIFFERENT FORCE DIRECTIONS

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Introduction: Generally muscle function and muscle activation is evaluated considering position and leverarm with respect to a joint axis. The actual situation in the shoulder is more complex: first, muscles may also be important for stabilization of the glenohumeral joint. Secondly, muscle-function can depend on the abduction angle. Thirdly, the position and direction of the muscle force vector is often not clear because of a large attachment site (Van der Helm 1991).

Methods: EMG was recorded with use of two intramuscular wires (subscapularis, supraspinatus; Nemeth 1990) and ten surface electrodes. Five healthy subjects were sitting, 90° anteflexion of the humerus, the elbow splinted in 90° flexion, forearm vertical. The elbowsplint was attached to a force transducer (2D forceplane perpendicular to the humerus). Subjects were visually informed about the resultant force level and direction by a monitor. Force directions: 0° (strict elevation) and clockwise steps of 45° until 360°. Force level: 13.9 N in all directions. In each force direction, 1 s of EMG was averaged. A trial consisted of eight alternating force directions, every subject performed three trials. EMG was normalized to maximal EMG in every trial.

Results and discussion: Every muscle has its highest activity in a specific force direction. The figures represent two typical examples of activation patterns with respect to force direction. In this way the function of 12 shoulder muscles can be discriminated with use of EMG, avoiding technique-related problems such as influences by contraction velocity or muscle length. The reproducibility of EMG was good (mean intertrial coefficient of variation was 25.3%, range 18.2% to 36.3%). In a 3D biomechanical model of the shoulder (Van der Helm 1991), muscle forces in the above mentioned force directions will be calculated and compared with our EMG data.

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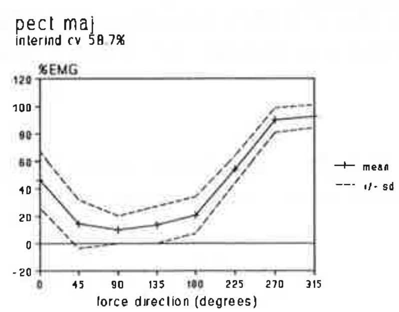

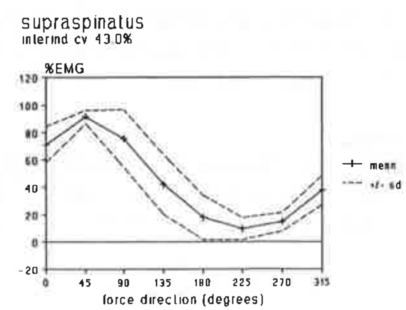

CROSSTALK MEASURES IN AGONIST MUSCLE GROUPS

IN SURFACE EMG

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In studies of human movement surface electromyography is the most common method of recording the activation patterns of all muscles except deep muscles. The convenience and non-invasive nature of surface electromyography is an obvious advantage. However, there are claims that surface electrodes (compared with indwelling) are prone to crosstalk between both agonist and antagonist muscles.

Crosstalk between antagonist muscles is very easy to assess; manual resistance tests demonstrate the presence or absence of crosstalk. For example, claims of crosstalk between the tibialis anterior (TA) and gastrocnemii (GAST) can be readily dismissed by manual tests which show TA to be silent when GAST are active and vice versa. Also, a review of raw surface EMG of these two muscles during gait demonstrates no crosstalk.

In agonist muscle groups, such as hamstrings and quadriceps the assessment of crosstalk becomes difficult because of common neural drive. This paper demonstrates a cross-correlation technique which quantifies the common signal in any two EMG channels. A total of 14 electrode pairs were placed in two rows of 7 each across the quadriceps at a spacing of 2.5 cm between electrode pairs. Based on 3 second isometric knee extensor contractions at 25%, 50% and 75% MVC at knee flexion angles of 30° and 90° the raw EMG cross-correlations in 3 subjects were calculated in order to quantify common crosstalk signal. Normalization of the cross-correlations and squaring their peak values yielded the following R^2 values averaged over all levels of contraction and muscle lengths: 2.5 cm spacing - .276; 5.0 cm spacing - .085; 7.5 cm spacing - .047. We interpret these results as predicting the worst case crosstalk between adjacent electrodes to be about 28%, dropping to less than 9% for 5 cm spacing and 5% for 7.5 cm spacing.

Based on basic volume conductor biophysics the theoretical pick-up range (95% of total signal) of surface and indwelling electrodes is about 1.7 cm from the conducting surface (Fuglevand, et al., 1992). A subsequent simulation analysed the myoelectric signal picked up from 8 electrode sites at 1 cm spacing around a large muscle of 5 cm in diameter. The cross-correlation between the signals from electrode pairs as a function of electrode spacing decreased in a similar fashion but more rapidly than that reported from the experimental results. This difference may be due to assumptions in the simulation model regarding motor unit population, recruitment and rate coding and thickness of fat tissue overlying the agonist muscles.

Both experimental results and theoretical predictions demonstrate that crosstalk in surface electromyography is not nearly as serious as claimed by some researchers. In fact, the onus on users of indwelling electrodes (because their indwelling electrodes are at a closer range to deep agonists) is to demonstrate that they are not subject to crosstalk from these deeper muscles.

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RELATIONSHIP BETWEEN M-WAVE SIGNAL AND MUSCLE CONTRACTIVE INTENSITIES

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The M wave is the manifestation of muscle compound action potential of a contracting muscle and this signal shape undergoes frequency and amplitude changes during muscle contraction of constant force intensities. Analysis of the surface myoelectric signals provide data relevent to altered motor unit action potential (MUAP) shapes and firing rates. Thus, this analysis may provide complementary information to the motor units activities during muximum voluntary contraction relative levels(%MVC).

The purpose of this study was to determine whether spectral and amplitude parameters of the M wave depend upon previous force intensities that allow for the order of the motor unit recruitment and firing rate, and whether changes of the muscle fiber action potential during electrically elicited contraction take place. Seven healthy adults(male) participated in this experiment, which was performed on the tibialis anterior muscle with the subjects sitting position. Supramaximal stimulation was applied, lasting 2.0s with a frequency of 5Hz at 20.0s interval to the appropriate motor point(end-plate) in order to elicit the M wave. Target for voluntary contraction intensities consisting of 40%, 60%, and 80%MVC were set on the scope. Each level of MVC was sustained for 180s(40%MVC), 100s(60%MVC), and 60s(80%MVC). Muscle fatigue was observed at the last values during the end contractions of relative level of MVC comparing with the initial ones. The Median Frequency(MDF), the Mean Frequency(MNF), the Average Rectified Value(ARV), the Root Mean Square(RMS), and the evoked Force(F) were calculated to evaluate the M wave modulation and motor unit recruitment during sustained contraction, quantitatively.

The modulation of the M wave shape was affected by force intensities of the muscle contraction and ARV, RMS, and Power density of the M wave increased in proportion to %MVC(refer to the right chart of Fig.1), while MDF, and MNF displayed slightly decrement on relative level of MVC with less significance. When dorsiflexion of the foot joint was restricted to higher intensity of MVC, the M wave enlarged in six of the seven subjects significantly. ARV, RMS, and Force increases were greater than MDF, and MNF those among contractive intensities of %MVC. The RMS enhancement were 111.2 + 24.6(40%MVC), 121.2 + 9.8(60%MVC), and 204.9 + 30.6(80%MVC),respectively. At the muscle fatigue, those enlargement of normalized ARV and RMS reached statistical significance on the last contraction value of 60%MVC(P 0.05).

M wave amplitude increase would suggest that much potentials of the motor unit were the result of fiber type II recruitment, higher firing rate and transmission in responses of the whole regions of the tibialis anterior relating to %MVC.

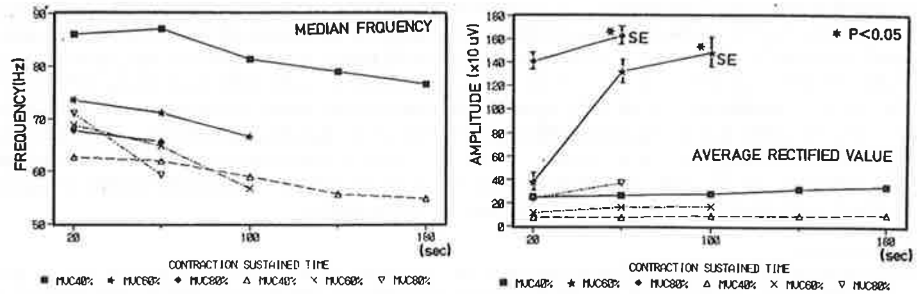

Fig.1 Time trend curves on typical data(MNF and RMS) of subj.MK-Solid line:M wave

THE INFLUENCE OF ANTHROPOMETRIC FACTORS ON THE EMG POWER SPECTRUM OF UPPER LIMB MUSCLES

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INTRODUCTION

The median frequency (MF) and mean power frequency (MPF) of the EMG power spectrum have been observed to present similar behaviours (increases) across increasing force levels for the anconeus (AN) but different ones for the triceps brachii (TB) [1]. Such varying behaviours between the MPF and MF of TB have tentatively been explained by a different low-pass filter effect of the skin on these two measures [1,2]. Moreover, dissimilar behaviours in both spectral statistics across increasing force levels have been observed between men and women, with women presenting less pronounced increases in MPF or MF [3]. The purpose of this study was to investigate the influence of several anthropometric factors, including skinfold thickness and sex, on the behaviour of both MPF and MF across increasing force levels.

METHODS

Twenty-nine subjects (16 women and 13 men) produced isometric ramp elbow flexions and extensions. The MF and MPF were obtained from EMG signals (TB, AN) corresponding to five different force levels [1,2]. The behaviour of both MF and MPF across increasing force levels was estimated by subtracting the value obtained at 20% MVC to that obtained at 80% MVC. For both spectral statistics, the result of this subtraction was used as the dependent variable in multiple linear regression models with stepwise procedure, where sex, skinfold thickness, body mass index (BMI = weight (Kg)/ height^2 (m^2) [4]) and circumference of the arm were used as the independent variables.

RESULTS

Table 1 shows, in order of importance, the variables that met the 0.15 significance level criteria for entry in the model, according to the stepwise procedure, for the two muscles and for both spectral statistics.

TABLE 1

Muscle	MPF/MF	Variables	ß-Coef.	partial R^2	p-value
TB	MPF	SKINFOLD	-1.2383	0.2463	0.0062
		SEX	11.7541	0.0786	0.0938
	MF	CIRCUMF	1.1941	0.1167	0.0697
AN	MPF	none	—	—	—
	MF	none	—	—	—

CONCLUSION

The present linear regression models show that the behaviour of the MPF across increasing force levels is significantly negatively related ($p<0.05$) to the thickness of the skin overlying the TB. The sex of subjects added only slightly to the model ($p>0.05$). Interestingly, the behaviour of the MF across increasing force levels does not appear to be influenced by the skinfold factor, which confirms our previous suggestion [1,2]. The arm circumference presented a weak relationship with the behaviour of the MF ($p>0.05$). For the AN, no variable was found to be significantly related to the behaviour of either statistic. The fact that the skinfold factor has no significant effect on the MPF of the AN, in contrast to the TB, can be explained by different characteristics of the skin layer, which is about 3 times thinner and 3 times less variable over AN than over TB [2]. It thus appears that the MF is more suitable when surface EMG evaluations are performed on muscles where skinfold thickness is not negligible.

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CHANGES IN PEAK TO PEAK VOLTAGES OF SINGLE FIBRE ACTION POTENTIALS AS A RESULT OF VOLUME CONDUCTION AND RECORDING DISTANCE

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Introduction. Quantitative aspects of single fibre action potentials (SFAPs) in skeletal muscle can be studied for one active fibre in animal experiments. Here we shall report about the dependence of the peak to peak voltage of SFAPs on the position of the recording electrode in practice and on inhomogeneous volume conduction properties in the tissue in simulations.
Methods. A method to determine the recording distance for wire electrodes has been described before with some preliminary data (Albers et al., 1989). Additional data using a comparable method was obtained. Also a multi-channel single fibre needle electrode was used at single fibre stimulation. The needle comprised 8 recording sites (distance between each other 130 micrometer; recording surfaces about equal to that of the wires); here the recording distance could be derived only relatively.
In simulations SFAPs were calculated with the model described by van Veen (1992), where inhomogeneities of the muscle were taken into account.
Results. The range of peak to peak voltages (Vpp) of SFAPs was comparable for wire and needle electrodes, up to a few mV. In wire recordings Vpp was not clearly related to the recording distance. The relatively high Vpp at large distance of the active fibre and the variable change of Vpp with recording distance were in accordance with the multi-channel needle recordings, where the recording distances were related to the chosen electrode configuration. In both conditions it occasionally occurred that Vpp increased with increasing distance. Simulations with connective tissue layers represented in the model only showed relative and never absolute increase of Vpp.
Discussion. Our wire recordings cannot be compared with data of other research groups. The experimental procedure is complex and it cannot be stated that the recording distances are fully reliable. However the general trend in the plot of Vpp against the recording distance needs not to be doubted. Moreover the results of the wire and the multi-channel recordings are in agreement. These considerations point to a failure of the volume conduction theory used in the simulations.

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SPATIO-TEMPORAL CHARACTERISTICS OF MULTICHANNEL ELCTROMYOGRAPHY

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ABSTRACT

This paper discusses the spatio-temporal properties of multichannel electromyographic (EMG) signals and investigates muscle force estimates obtained by multichannel myoprocessor. The multichannel design is adopted to include the spatial information and thus enhances the sensitivity of the myoprocessor. In our experiment, fourteen pairs of electrodes with differential preamplifiers are applied to four major muscles about knee movement. Multichannel EMG signals and torque generated are measured during isometrical extension and flexion contraction in burst and ramp movements. To investigate the relationship between EMG signals recorded at different sites, the coherence function measuring the linear relationship in frequency between two EMG signals is utilized. To remove the spatial coupling effect between multichannel EMG signals, the spatial prewhitening processing is used to eliminate this effect.

In this research, the power and shape of EMG spectra are found to be useful for determining the activation level and signature of a specific movement. It can be seen that the EMG signals measured from two adjacent electrodes have higher coherence values indicating better linear relationship in frequency. The frequency relationship of EMG signals measured at different electrode sites seems to be in a Gaussian distribution form. The results of muscle force estimation indicate that the more the electrodes are used, the less the estimate error is obtained. In the study of using model obtained in ramp force estimation to estimate constant force, the superiority of multichannel myoprocessor is observed. In addition, the result shows that the estimation error at lower contraction levels is less than that of higher contraction levels.

The results of this research show the advantage of multichannel EMG processing. The extension of this study could be applied to functional neuromuscular stimulation and active prosthesis control. This study would be helpful in determining the applicability of multichannel surface EMG signals as a diagnostic and assessment tool for rehabilitation and neurology studies.

EMG STUDY OF THE ANTERIOR, SUPERIOR AND POSTERIOR AURICULARIS MUSCLES IN MAN

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INTRODUCTION

The extrinsic musculature of the ear has practically not been studied electromyographically. In the literature, we found only the work of Serra et al. (1985), who studied the posterior auricular muscle in lesions of the facial nerve. These muscles are considered, by many text books, to be atrophied and without action, while other authors consider that each of these muscles move the ear in the direction of its fibres, thus the anterior would dislocate forward, the superior, upward and the posterior backward. By virtue of the scarcity of literature and the controversiality in the text books we decided to study these three muscles electromyographically.

MATERIAL AND METHODS

30 young volunteers were, studied by means of a TECA TE-4 electromyograph. electrodes as was utilized by SUMITSUJI et alii (1965) were used, which consist of a steel thread of 80 um diameter, introduced manually through the skin reaching the muscle. This operation in faclitated by the lack of fascia of the mimic musculature. The muscles were studied in the following movements: moving the auricula upward, backward and forward, smiling naturally, and yawning. Isolated movements of the auricula upward, forward and backward were not observed.

RESULTS AND DISCUSSION

In 16 volunteers, it was observed just small dislocations of the auricula upward and backward; in these cases the three muscles acted as a group. In smiling and the yawning, they were concomittantly active, with strong action potentials in all 30 volunteers.

This results shows that muscles are not atrophied and without action. They act in conjunction in dislocation of the external ear, and the result of these three forces of the auricular muscles, each one applied in one direction, caused the auricula to be dislocated upwards and backwards. And the auricularis act in facil expression as smiling an the yawning.

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ACTION POTENTIAL VELOCITY MEASUREMENTS IN THE UPPER TRAPEZIUS MUSCLE WITH SURFACE EMG - A PILOT STUDY

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Most motor unit action potential velocity (MUAPV) measurements with surface EMG described in the literature have been performed in m. biceps. Occasional investigations have been performed in m. triceps, m. tibialis anterior and m. vastus lateralis. The trapezius muscle is of special interest in ergonomics since it responds to long time low level static exposure with chronic myalgia in a way which is hardly seen anywhere else in the human body. Previous investigations by the same author, applying zero crossing technique, indicate relations between fatigue response and muscle disorder. These relations might presumably be further elucidated by MUAPV measurements.

A measurement procedure based on the double differential technique (DDT) was set up. An electrode with four parallel silver bars, 5 mm long and with 10 mm interspace was oriented along the muscle fibres in the upper distal part of the trapezius muscle. A polar correlation algorithm was implemented on a PC to estimate the delay τ between the two DDT channels. Each estimate was based on 1 s of data. The average MUAPV was calculated as d/τ where d is the electrode distance.

Measurements were performed on the trapezius muscle in 6 males and 6 females while holding out the arm horizontally in the sagital plane. The electrode was initially located in the most distal position giving appropriate EMG signals (Pos. 1). Up to three trials were made with minor electrode adjustments in between to obtain a MUAPV estimate with a cross-correlation $R \geq 0.80$. After moving the electrode 5 mm proximally this procedure was repeated. In this way totally 5 positions were tested.

The results in Table 1 clearly demonstrate that the distal positions are most suitable for reliable estimates. An innervation zone is likely located in the middle part of upper trapezius which implies action potentials travelling in both directions in this area. Further more, reliable results are easier obtained on males than on females. This is probably explained by larger muscle dimensions in men but also by less subcutaneous fat. The failure in M6 was caused neither by subcutaneous fat, nor by small dimensions. A possible interpretation is that some individuals have a more diffuse and widespread innervation zone. The absolute values are in good agreement with results from other investigators in other muscle groups. Females tend to have higher estimates. This may indicate greater type I fibres in the female trapezius. However, methodological ambiguities should be further investigated.

Trials with single differential technique were fruitless. Double differential technique seems to be essential in the trapezius muscle.

Table 1. Conduction velocity in 5 position of the upper trapezius. M=Male, F=Female

	Conduction velocity m/s									
Ind.	Pos. 1		Pos. 2		Pos. 3		Pos. 4		Pos. 5	
nr.	F	M	F	M	F	M	F	M	F	M
1	5.78	4.17	-	3.73	-	4.13	-	-	-	-
2	-	5.02	5.78	4.92	-	-	-	-	-	-
3	4.25	3.39	4.48	4.17	5.83	-	8.69	-	-	-
4	6.08	4.69	-	5.45	-	5.94	-	-	-	-
5	-	5.86	-	6.19	-	-	-	-	-	-
6	-	-	-	-	-	-	-	-	-	-
Mean	5.37	4.62	5.13	4.89	5.83	5.03	8.69	-	-	-

A SYSTEMATIC APPROACH TO THE DESIGN OF 2-D ARRAY ELECTRODES: SPATIAL CHARACTERISTICS OF SURFACE MYOELECTRIC SIGNAL

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In the last ten years several researchers reported applications of mono- and bi-dimensional array electrodes to surface myoelectric signal processing. The most common utilization of these multielectrodes is represented by the location of the innervation zones beneath the probe. The position of innervation zones is generally obtained by observing the time relationships between single- or double-differential signals detected in different points along the active muscle fibers. It follows that the choice of the distance between consecutive detection surfaces (grid spacing) is not critical, and it is primarily related to the desired spatial resolution. However, it is suggested that the study of the potential distribution on the skin above an active muscle could allow researchers to gain a deeper understanding of the electrical phenomena taking place inside the muscle tissue, and, possibly, to obtain some information about muscle structure (i.e., muscle fiber direction, tendon insertion, average length of muscle fibers ...). In order to achieve these goals, we must be able to represent without appreciable distortion the potential distribution on the skin, due to muscle activation. To this extent, the proper choice of the grid spacing is crucial. In fact, if the detection surfaces were too distant, the observed signals would be affected by relevant *spatial aliasing*, whilst too small interelectrode distances would cause *spatial oversampling* without increasing the information content of the sampled signals.

The aim of this work is to present some design rules to allow experimenters to choose the proper interelectrode distance by taking into account the relevant physiological and anatomical characteristics of the muscle to be studied. These rules are obtained basically by applying the Nyquist's criterion to spatial sampling. The issue of combining mono- or bi-dimensional samples to obtain different spatial filters is not dealt with in this work.

The potential distribution due to muscle activation on the skin was simulated by means of a two-stage model. The first stage consists of a finite difference model of a muscle fiber based on the Hodgkin-Huxley equations. This model takes into account some electrical characteristics of the muscle membrane and of the interstitial and intracellular fluids, and simulates the evolution in time of the potential distribution along the finite length muscle fiber. The second stage takes as input the potential distribution computed by the first stage at a given time instant along the muscle fiber and computes the potential distribution in a cyclindrical finite volume-conductor that surrounds the muscle fiber itself. The knowledge of the potential distribution caused by a particular muscle fiber on the lateral surface of the volume-conductor allows us to study the distribution of the power of the signal among its spatial frequencies. The design rules to be used in determining the grid spacing are then obtained by studying the mono- or bi-dimensional cumulative power distribution of the signal.

ACKNOWLEDGMENTS
This work was supported by the Italian C.N.R. (grant # 91.02901.CT07).

EFFECT OF ELECTRODE LOCATION ON SURFACE MYOELECTRIC SIGNAL VARIABLES: A SIMULATION STUDY

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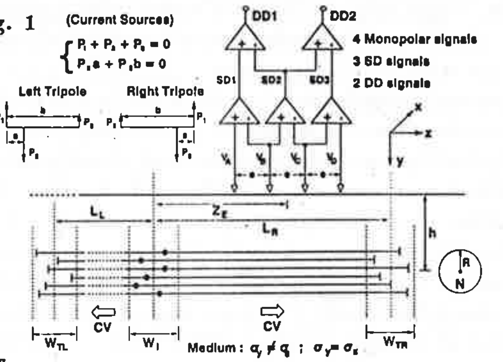
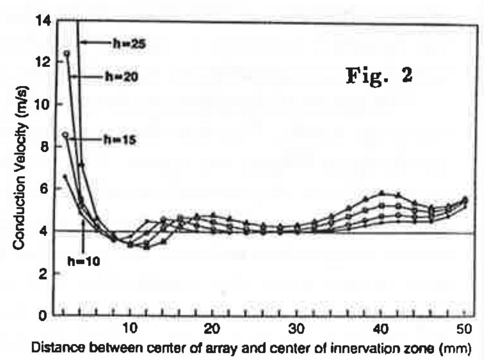
A current tripole model has been used by many authors (1,2,3) to simulate an action potential travelling along a muscle fiber. Two tripoles are assumed to be generated at the neuromuscular junction (NMJ) and to travel toward each tendon end in opposite directions. At the instant of excitation, for each direction, the three monopoles (whose sum is zero) are superimposed at the NMJ, then the first moves away from it at a velocity v for a distance a, the second pole then separates and moves for a distance b-a while the first moves to b and finally the third pole moves. From this instant the entire tripole travels toward a fiber end until the first monopole reaches it and stops. The second and third monopole travel toward it, the tripole narrows, until the second monopole and then the third reach the first and cancel out.

This model has been widely used (1,2,3), although it simulates the generation and extinction of the action potential in a rather arbitrary way. A motor unit is simulated as N parallel fibers scattered with uniform distribution within a circle of radius R. The center of the circle is at a depth h, with respect to the surface, into an infinite homogeneous semispace having electrical conductivities $\sigma_y = \sigma_x \neq \sigma_z$ (Fig. 1). The individual NMJs of the N fibers and the fiber ends are uniformly distributed within regions of width W_i, W_{TR} and W_{TL} respectively. The distance between the centers of the NMJ and termination zones is L_R and L_L on the right and left side respectively. M motor units may be simulated, with either synchronous or asynchronous firing, each with a different set of parameters. An array of four point electrodes with interelectrode distance e is placed on the surface at an angle θ with the direction z of the fibers. The distance between the center of the innervation zone and the center of the array is Z_E. Four monopolar (M), three single differential (SD) and two double differential (DD) voltages are detected as indicated in Fig. 1. Spectral and amplitude variables are computed from the SD2 voltage, conduction velocity is computed from the two DD voltages. The model allows the simulation of a variety of different conditions and the testing of algorithms for myoelectric signal processing. Fig. 2 shows the estimates of v based on the two DD signals for different locations of the electrodes along the right half of a single motor unit with N=100, R=5mm, L_R=L_L=50mm, $W_i = W_{TR} = W_{TL}$=5mm, v=CV=4m/s, e=10mm, θ=0 and h=10,15,20,25 mm.

Fig. 1





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CONTINUOUS PERCUTANEOUS MEASUREMENT BY LASER DOPPLER FLOWMETRY (LDF) OF SKELETAL MUSCLE PERFUSION IN RELATION TO VARYING ISOMETRIC WORK DETERMINED ELECTROMYOGRAPHICALLY (EMG)

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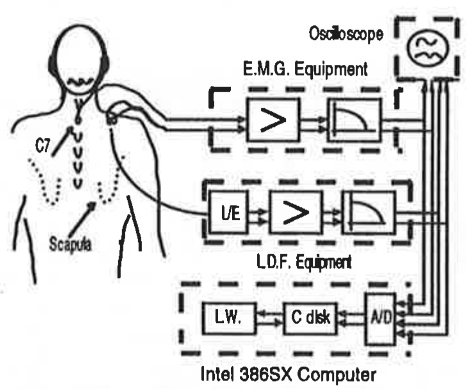
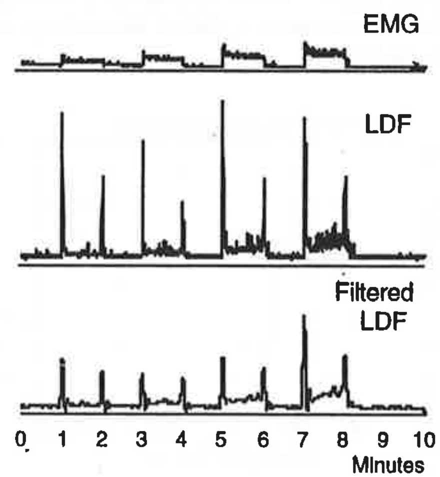
Swedish Work Environment Fund, project no. 87-0788.

Skeletal muscle perfusion was measured during muscle work, fatigue and recovery. Single-fibre LDF and surface EMG of the trapezius muscle were performed continuously during two 10-minute series alternating one-minute periods of static contractions and rest, followed by a 10-minute recording of fatigue and recovery. Stepwise increased contraction force was induced by keeping the straight arms elevated at 30, 60, 90 and 135 degrees, respectively, in the second series with 1 (women) or 2 kg (men) hand load. The arms with hand load were then kept at 45 degrees until fatigue. On-line computer (Intel 386 SX) processing of the LDF and EMG signals made possible interpretation of the relationship between the perfusion and the activity of the muscle. Root mean square (RMS)-converted LDF- and EMG-signals were normalized to reduce the variations which were of comparable magnitude. Frequency spectral analyses of EMG showed lowest variability for median frequency (MDF) and frequency range 10-1,000 Hz, as compared to mean power frequency (MPF). Disturbances as well as the filtering effects were studied by analyses of the LDF power spectral density obtained at low (rest), and high (contraction) muscle perfusion. Autoregulation was indicated by rhythmical variations with 5-6 cycles/minute. Biological variables causing temporal and spatial variations in muscle perfusion appeared to be main factors of a relatively large intraindividual variability. Increased muscle work resulting in an 80% increased EMG activity (amplitude) and fatigue (EMG frequency shift) was followed by an 80% increased LDF during the first two minutes of recovery. Muscle perfusion was increased also during sustained high-force contraction.



Experimental set-up. >=amplifier, [filter symbol] = filter, L/E=laser electronic signal converter, L.W.=lab-window software.



MS-converted LDF and EMG signals. The LDF signals are shown before and after processing. Initial disturbances are caused by changed arm position.

INNERVATION DENSITY ESTIMATE WITH NONSTATIONARY DATA

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The innervation process for a skeletal muscle motor unit can be modelled as a stochastic renewal process. As such, the probability density function, f(x), for the interspike interval random variable, x, is sufficient to completely describe the innervation process. Implicit in this statement is that the process be stationary, or otherwise the density function must be replaced by a time dependent function, f(x, t). From a measurement point of view, the estimation of f(x, t) requires an ensemble of sample functions from the process over which an average can be taken for a given value of t. This is, in most applications, impractical and most measurements are made on a single sample function with averaging over time replacing ensemble averaging. In this case, the experimenter will assume, and attempt to assure that, over some finite data observation interval, the innervation process is self-stationary. The assumption is questionable, and the difficulties of assuring self-stationarity many. This paper investigates the effects to be expected from innervation process nonstationarities on the time-average estimate, $\hat{f}(x)$, of an underlying density f(x).

The theoretical and simulation results show that if the mean of the underlying innervation process is nonstationary, the estimate $\hat{f}(x)$ will be broadened with respect to f(x). If the mean and standard deviation of the process are nonstationary, $\hat{f}(x)$ will show both a broadening and a skewness with respect to f(x). Measurements of f(x) for the biceps brachia are made under stationary and nonstationary conditions. Data is obtained from single motor units using fine-wire intramuscular electrodes. The degree of nonstationarity is assessed qualitatively by run tests, and quantitatively by the variance of the mean estimate relative to expected sampling variance. Figure 1 shows $\hat{f}(x)$ for a) a stationary data set and b) a nonstationary data set. The $\hat{f}(x)$ for the nonstationary data is skewed and broadened relative to that from the stationary data set, as predicted.

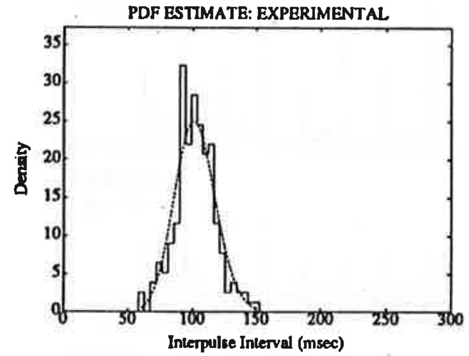

Figure 1 (a)

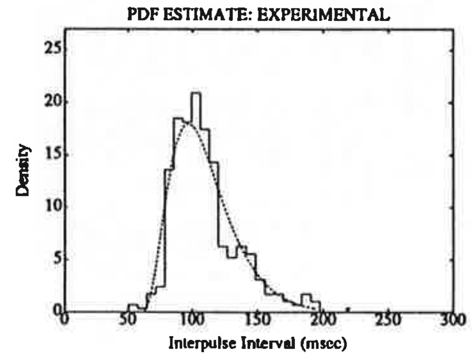

Figure 1 (b)

ASSOCIATION BETWEEN ANATOMICAL AND ELECTROPHYSIOLOGICAL PROPERTIES OF THE TIBIALIS ANTERIOR MUSCLE

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The tibialis anterior muscle is suited for non-invasive muscle fibre conduction velocity (MFCV) determination by using electrode arrays along the muscle (Masuda and Sadoyama, 1987; Merletti et al., 1990, own experience). Two peculiarities found with this special electrophysiological technique will be discussed here in association with anatomical features of the muscle.

Electrophysiology

The first aspect is the region in which MFCV measurements are possible. The distal part of the muscle is well suited for MFCV determination, whereas more proximally the muscle shows a rather abrupt change-over to an unstructured signal behaviour in which reliable MFCV measurements are virtually impossible (Broman et al., 1985; Masuda and Sadoyama, 1987; own experience). The second finding is related to supramaximal motor-point stimulation of the muscle. Electric stimulation allows determination of a muscle fibre conduction velocity in two ways. A velocity (U1) is determined by the travelling time between the two compound muscle action potentials (CMAPs), recorded from two electrode pairs, with a 12 mm separation in fibre direction. A second time interval can be measured between the moment of stimulation and the moment of arrival of the CMAP at the electrodes. This interval determines a mean velocity (U2) between the site of stimulation (a motor-point) and the site of recording. The peculiarity is that this velocity is much higher (U2=16-20 m/s) than the local MFCV (U1=4-6 m/s).

Anatomy

An anatomical prerequisite for a successful non-invasive determination of the MFCV from the surface EMG is a parallel organization of muscle fibres in a not too short section of the muscle. In a cadaver study in six different cases it was confirmed that such a section is only present as a rather superficial structure of a few millimetres thickness distally in the muscle. This layer, situated over the distal tendon, is up to 10 cm long and 2 cm wide and ends gradually in the tendon at the distal side. In proximal direction this parallel fibre section is interrupted by a pinnate part of the muscle with short fibres. Below the distal tendon the muscle is also of a pinnate type. These observations can explain the electrophysiological findings. The tibialis anterior anatomy will be presented in detail for one of the cases studied.

The striking difference between the two conduction velocities U1 and U2 can be explained by assuming that the motor point is situated somewhere along the deep peroneal nerve or along one of its main branches. It was confirmed anatomically that the motor-point is likely to be situated where one of the peroneal nerve branches enters the tibialis anterior. It must be assumed that this branch proceeds in distal direction. With a mean nerve conduction velocity of 40-50 m/s the consequence of this assumption is that the pathway from motor-point to recording site (about 20 cm) is for about 75% a nervous pathway. This is in disagreement with the common concept that a motor point is anatomically not too far from the neuromuscular transmission area. For confirmation a further anatomical study will be performed to follow the pathway of the nerve branch within the muscle.

The SEMG peculiarities analyzed seem to have a straightforward anatomical explanation. In general it is expected that confusing findings in muscle electrophysiology can be explained by specific studies in cadavers. Such studies give insight in specific details and also in the anatomical variation which is hard to obtain in another way.

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MOTOR FLUCTUATIONS IN PARKINSON'S DISEASE EVALUATED BY MEANS OF ACTOMETERS

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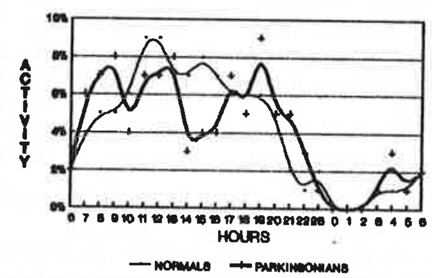
Motor fluctuations are usually reported by patients with Parkinson's disease, mostly after long lasting L-DOPA therapy, but the assessment of temporal pattern of motor performances is difficult on a clinical base only. Therefore activity monitors might be useful to evaluate circadian fluctuations (1,2).
In this study we used a solid state activity recorder, developed by ANTEL Snc (Teramo - Italy), to analyze the temporal pattern of motor activity in patients with Parkinson's disease and in normal subjects.
Eleven patients, 3 women and 8 men, mean age 69.6 ± 17, with Parkinson's disease lasting at least 3 years, in steady state L-DOPA therapy, were evaluated during an hospital admission. Parkinson-like disorders were excluded by means of clinical and neuroradiological examinations. Patients with a Hoehn-Yahr score not in the range 2 to 4 and patients with relevant on-off phenomena requiring more than three doses of therapy per day were also excluded. Only 3 of the included patients reported motor fluctuations in their medical history.
The daily motor activity of this group was compared with that of 17 normal elderly subjects, 6 women and 11 men, with a mean age of 64.4 ± 12: 6 patients had been admitted for headache, 5 for transient ischemic attacks, 3 for asymptomatic carotid stenosis, 4 for a history of non-complicated head trauma, 1 for late epilepsy. Patients with abnormal neurological, neuropsychological or neuroradiological examinations were excluded.
The recording apparatus was a computerized microdevice with a piezoceramic sensor which was worn on the right leg during a full 24 hours period by the subjects freely moving inside the clinic.
An ad hoc software package produced a total score of motor activity and the percent distribution of activity on a hourly temporal base. Statistical analysis was performed using the Student's T test.
As expected, results showed a significant decrease ($P<0.05$) of the total activity score in patients with Parkinson's disease. The temporal distribution of activity (fig.1) in normal subjects showed a progressive increase of percentages until noon and then a progressive decrease during afternoon. On the contrary patients with Parkinson's disease showed three different peaks of activity during the day: between 8 a.m. and 10 a.m., between 11 a.m. and 2 p.m. and between 7 p.m. and 10 p.m. The relative activity was significantly higher in parkinsonians between 8 a.m. and 10 a.m. ($P<0.05$) and between 7 p.m. and 11 p.m. ($p<0.01$) but significantly lower between 11 a.m. and 1 p.m. ($p<0.05$) and between 3 p.m. and 5 p.m. ($P<0.01$).

fig.1: **DAILY MOTOR ACTIVITY**



In conclusion our study show that actometers are useful for the analysis of temporal patterns of motor fluctuations.

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SPEED RELATED CHANGES OF GAIT IN PATIENTS WITH PARKINSON'S DISEASE

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Instrumented tecniques are used to assess gait analysis but often these tecniques are pure laboratory research tools.(1)
An evaluation of spatial and temporal phases of gait was performed on 10 patients with Parkinson' s Disease (PD) with Hoehn -Yahr scores ranging from 1 to 4 and 20 healthy sex and age matched subjects.
We utilized 3 pairs of microswitches placed under the heel and the head of the first and fifth metatarsal bones.
The data were recorded by a portable device (size 21 x 15 x 7.5 cm; weight 250 gr) which contains a microcomputer and were sent by an RSR 232 to a 386/25 PC for storage and statistical analysis.(2)
The subject were asked to walk for an 8 m. course at different speeds (normal, slow, very slow and normal, fast and very fast).
This task was performed on the day of admission and on the following day without any change in therapy.
The data about spatial and temporal phases of gait are shown in table 1.
PD patients showed a significant reduction of stride length and heel strike ($p<0.001$) and double support time ($p<0.05$) while cadence is similar to normal subjects.
Adjusting mean values for walking speed, the stride lenght was still decreased in patients with PD, while cadence was significantly higher ($p<0.01$).

Tab. 1 GAIT PARAMETERS AND THEIR CORRELATION WITH WALKING SPEED

	Unadjusted values			Adjusted values	
	Controls	PD	slope	Controls	PD
Velocity (m/s)	0.83	0.79	-	-	-
Stride length(m)	0.96	0.90 a	0.76 a	0.95	0.91 a
Cadence (cyc/m)	48.3	48.2	23.03 a	47.8	48.8 b
Heel strike(%)	18.35	15.39 a	-5.51 a	18.5	15.7 a
Flat foot contact%	15.68	16.00	-17.61 a	16.1	16.8
Fore foot contact%	23.66	24.35	16.46 a	23.3	23.6
Swing%	42.36	42.43	6.04 a	42.2	42.1
Double support%	15.04	13.92 c	-10.40 a	15.3	14.5 c
Single support%	84.62	85.08	9.91a	84.4	84.6

*Regression slope of walking velocity of gait parameters.
a=$p<0.001$; b=$p<0.01$; c=0.05

Our device, using tasks at different speed, reveals some typical gait patterns in patients with PD and it may be used to follow the progression of disease and its changes with therapy.

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SPECTRAL ANALYSIS OF MOVEMENT IN PATIENTS WITH NEUROLOGICAL DISEASES

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INTRODUCTION

A physical model for the evaluation of human voluntary movement assumes that it is performed with a previously established program and controlled by means of a sensory feedback with several levels of integration with different resonance frequency.
Based on this model, the frequency spectrum obtained by the Fourier Transform of the movement function will show components related to different controlling loops.
The aim of the study was to evaluate with spectral analysis the upper right movement during a simple task in order to detect specific modifications in different neurological diseases.

MATERIAL AND METHODS

Sixtyeight patients with neurological disease (30 with Multiple Sclerosis (MS), 15 with Parkinson disease (PD) , 15 with cerebrovascular right hemiparesis (HE) 8 with cerebellar signs (CE) and 80 controls (CTRL) were studied.
All MS patients have neurological signs in the righ upper arm: 8 pyramidal signs, 8 cerebellar signs and 14 pyramidal-cerebellar signs.
Patients were asked to perform a Gibson spiral test on a drawing board with a stilus connected through a serial interface to a personal computer.
The sampling frequency was 100 Hz. The signal was filtered by using a Kaiser window with a cutting frequency of 20 hz and than analyzed by Fast Fourier Transform (FFT).
Twenty spectral bands (0-20) were analyzed in the evaluation of both absolute and relative spectrum power.
Data collected from patients and controls were analyzed with a multivariate analysis of variance (MANOVA) adjusting for age of patients and angolar velocity.

RESULTS

Our data showed significant differences between controls and patients with neurological diseases.
Specific patterns were found for different clinical syndromes.
In addition, our findings showed difference between spectrum of pyramidal and cerebellar MS patients when compared to those of other patients with similar neurological syndrome.

COMMENT

Our data suggest that Gibson's spiral test with power spectral analysis can usefull in the assessment of different clinical picture associated to neurological diseases in clinical and research practice.

TOTAL MOTION ANALYSIS: THE KEY TO INTEGRATED MOTOR CONTROL

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Abstract. In multisegment voluntary human movements it is essential to understand the multivariate nature of the movement task, the competition and collaboration between muscle groups and the confounding problems associated with the considerable redundancy in the musculo-skeletal system. Using level walking as an example the functional sub-tasks that must be successfully accomplished are defined and the role of each motor pattern is related to those tasks. Two examples are presented to demonstrate these issues, one from normal walking, the other from the assessment of a complex pathological gait problem.

1. Introduction

Most total body movements such as walking, lifting and throwing involve the coordinaton of a large number of muscle groups acting across many joints and controlling many segments. Researchers and clinicians are faced with a multivariate movement task with both collaboration and competition between muscle groups. The converging nature of the neuromuscular system adds considerable redundancy which confounds the interpretation but also results in considerable adaptability to the system. Using normal level gait as an example three functional sub-tasks are defined. Each are necessary (but not sufficient) to achieve the total movement we call walking: maintenance of balance and posture in plane of progression and the frontal plane, defence against collapse against gravity and forward propulsion including safe trajectory of the lower limb (minimum toe clearance and a gentle heel contact). All the motor patterns that we can quantify must be related to those three sub-tasks. In this way we can readily identify complementary and competing patterns. The term motor pattern includes EMG profiles, joint moments-of-force and joint mechanical power patterns.

2. Balance, Posture and Support-Sub-Tasks

The first two sub-tasks are orthogonal in nature and they involve two well identified synergies that defend against a fall forwards, sideways and vertically. They are regulatory in nature and are superimposed on motor patterns associated with forward propulsion.

Control of the dynamic balance of the head, arms and trunk (H.A.T.) in the plane of progression is an active control that regulates H.A.T. as an inverted pendulum system to keep it erect within a few degrees over the gait cycle [1,2] and the A/P horizontal accelerations decreasing as we progress upwards from the pelvis to the thorax and head. From an inverted pendulum model of the H.A.T. during locomotion (Figure 1) it is evident that the major perturbation to unbalance H.A.T. is the horizontal acceleration of the hip joint of the support limb. This acceleration creates an unbalancing couple, $M_u = ma_x d$ which must be resisted by an extensor/flexor motor pattern by the hip

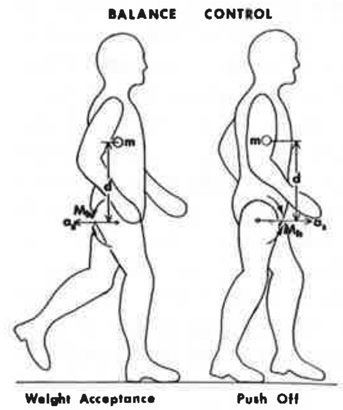

Figure 1. Model of H.A.T. as an inverted pendulum during stance showing unbalancing acceleration and hip balancing moments.

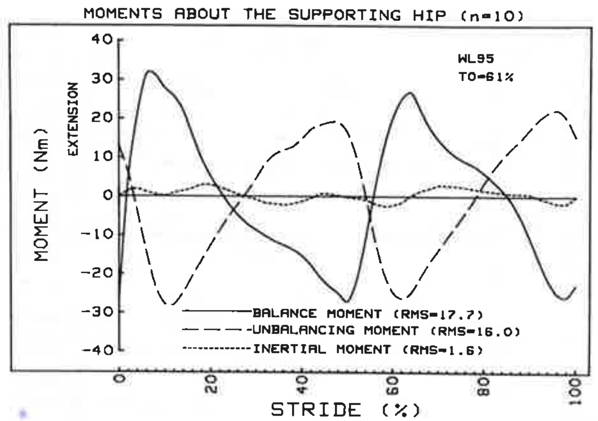

Figure 2. Average of 10 repeated trials of same subject showing the unbalancing moment, M_u, due to hip horizontal acceleration and the almost-cancelling balance moment, M_b, due to the hip extensors/flexors.

muscles. The high variability of the hip moment profiles during stance as seen in intra-subject averages [3] across repeat trials (minutes apart or days apart) has been attributed as a deterministic adaptation on each stride to balance H.A.T. [4]. This was confirmed by Ruder [5] who demonstrated that the hip moment pattern during stance was virtually equal and opposite to the unbalancing moment (Figure 2). These hip moment patterns were labelled balancing moments, M_b. The dynamic equilibrium equation of the inverted pendulum is:

$$M_u + M_b + mgx = I_h \alpha_h \qquad \text{Eqtn.1}$$

where: I_h is the moment of inertia of H.A.T. about the hip joint
α_h is the angular acceleration of H.A.T. in the plane of progression

In Equation 1 the gravitational term, mgx, is essentially zero for erect posture of H.A.T. Thus the angular acceleration of H.A.T. is low because M_b is almost equal and opposite to M_u. However, M_u varies on a stride-to-stride basis and thus M_b must also vary to regulate H.A.T.

However, this dynamic balance of H.A.T. by the hip flexors/extensors is not independent of the second regulating task, that of defence against a vertical collapse. Although the perturbations are orthogal in space the body's response is not orthogonal because of inter-limb coupling and the presence of biarticulate muscles. Twelve years ago a total limb synergy, called support moment, was identified [6] and is the summation of the ankle, knee and hip moments: $M_s = M_a + M_k + M_h$. Figure 3 shows these plane-of-progression moments for 9 repeat trials on the same subject over days. As can be seen the variability of these ensemble-averaged curves, as quantified by their coefficient of variation (CV) measures, was high at the hip and knee during the stance period. M_s, which is the sum of all three joint moments and represents the net pushing away from the ground by the total lower limb has a low CV, as does the ankle moment. Thus there must be some form of cancellation in the variability between the hip and

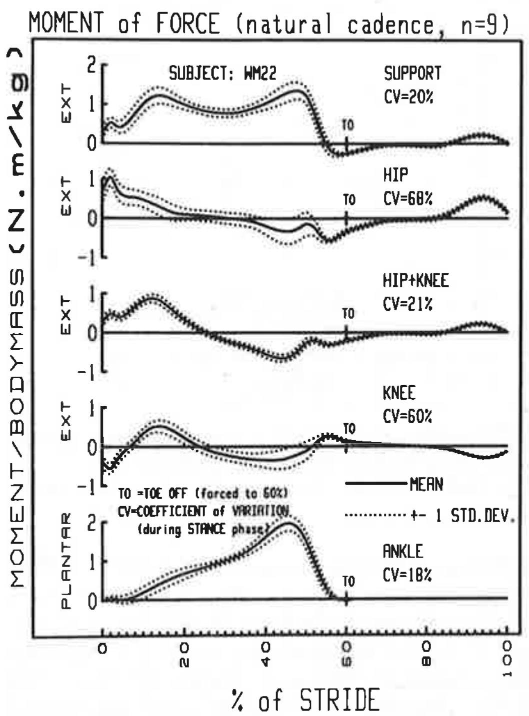

Figure 3. Ankle, knee, hip and support moments for 9 repeat trials of the same subject walking her natural cadence over nine days. See text for detailed discussion.

knee, and this was demonstrated by a separate plot of the ensemble average of the hip + knee moments: the CV score dropped to 21% from 60% or more for the hip and knee separately. Thus, a trade-off between the hip and knee moments is taking place on a stride-to-stride basis and that the knee is compensating with an almost equal and opposite variability. Thus the total moment acting on the thigh is quite invariable and that combined with the consistent ankle moments explains why the ankle, knee and hip joint angles are quite invariable over these trials - they vary ± 1 1/2° [3]. A single measure of the covariance between these moment patterns is a covariance score and for these repeat trials the hip-knee covariance was 89% [3,7]. When walking trials are compared minutes apart the variance at each joint drops drastically but the % covariance between the hip and knee remains about 65-75% [7]. In the elderly, this covariance score was seen to degenerate in the elderly to about 55% [8].

3.1 Assessment of Pathological Gait - Example of the Total Lower Limb Support

Because these synergies associated with balance and support apply equally well to those with gait pathologies it is critical to understand that any given patient may be adapting within the constraints of their pathology and that the compensating patterns may also appear to be "pathological". For example, knee surgery patients routinely experience knee pain and loss of knee extensor function. Thus it is common for them to unload their knee and not use their quadriceps during weight bearing. According to the principle of support moment the knee can also be controlled during stance by hyperactivity by the hip or ankle extensors. In a representative sample of knee surgery (patellectomy) both surgical and non-surgical limbs had the same support moment pattern (Figure 4) while M_a, M_k and M_h were total different. The primary problem at the surgical knee was the loss of knee extensor pattern for all of stance, likely due to inactivity of the quadriceps due to pain in the former pattelo-femoral joint plus loss of muscle strength and mechanical advantage. This patient compensated with hyperactive hip extensors and plantarflexors, thus the resultant gastrocnemii and hamstrings activity contributed to the observed knee flexor moment. These two adaptations in effect replaced the quadriceps and resulted in a stable support moment which was almost equivalent to that from the non-surgical limb. Only through a total analysis of all three joints of both limbs were we able to see this total neuromuscular compensation.

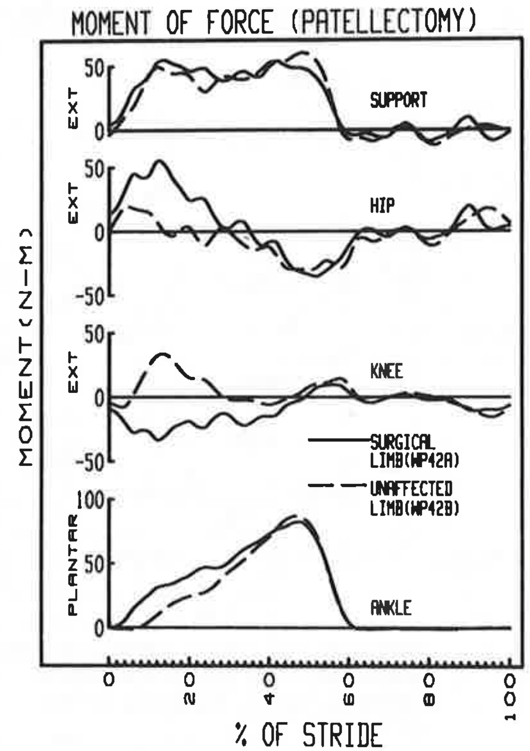

Figure 4. Ankle, knee, hip and support moments for the surgical and intact lower limbs of a patient who had a patella removed. Surgical limb demonstrates almost total unloading of quadriceps and a compensation by both ankle and hip extensors. The total support synergy for the surgical limb remained the same as the intact limb.

3.2 Multifunctional Role of the Same Muscles - Case Example of A/K Amputee

During normal level gait with an erect posture of H.A.T. the hip extensors during weight-acceptance have a dual role to balance H.A.T. and to assist in the control of knee collapse. In pathologies gait such as above knee amputees these dual tasks become even more important. Figure 5 demonstrates that the hip extensors must lock the knee in stance in order to prevent knee collapse during 80% of stance. However, this hyperactivity of the hip extensors also controls the dynamic balance of the amputee's H.A.T. If the amputee attempted to maintain normal erect posture his H.A.T. would be unbalanced posteriorly, thus he must adapt with a significant forward lean. Also, because this amputee has no push-off (the dominant energy generation phase by the plantarflexors in normal walking) he must adapt with extra generation elsewhere. Only his hip flexors and extensors have energy generating potential. During most of stance the hip extensors are concentrically active, thus in Figure 6 we see a large mechanical power generation phase (H1) followed by a short sharp "pull-off" power burst (H3) immediately prior to TO as the hip flexors unlock his knee and accelerate the total lower limb forward. The triple simultaneous role of the hip extensors is now evident only because a complete kinetic analysis was performed (both moments and powers).

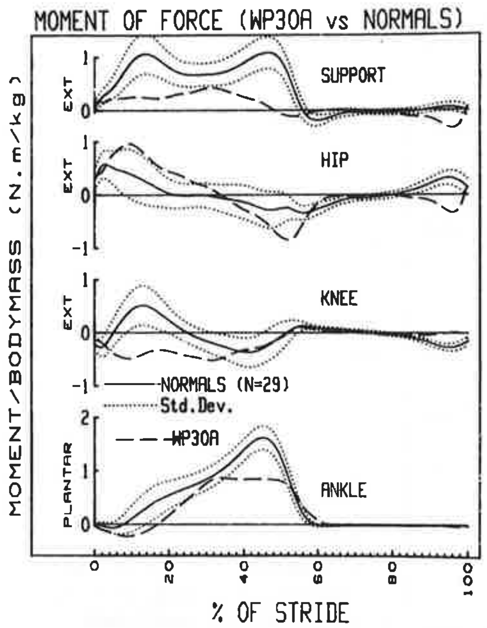

Figure 5. Ankle, knee and hip moments for an A/K amputee (dashed line) compared with the average of 29 normal subjects. See text for detailed discussion.

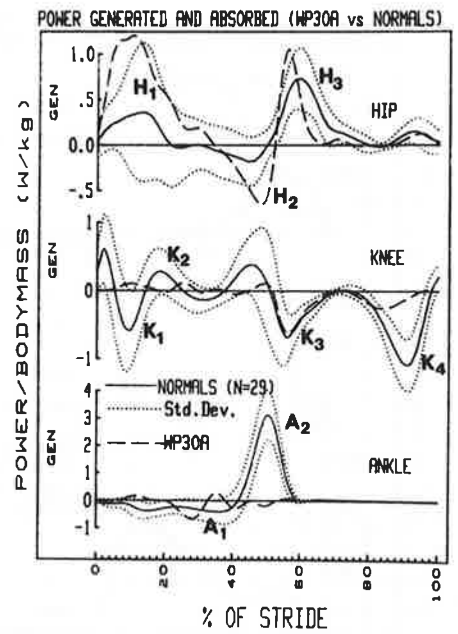

Figure 6. Mechanical power patterns at the ankle, knee and hip for the same A/K amputee as reported in Figure 5. See text for detailed discussion.

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DYNAMIC CHANGES OF ANTERIOR TRANSVERSE ARCH DURING WALKING

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Introduction

There are many deformities of the forefoot, such as hallus valgus and spread foot which affects the anterior transverse arch of the foot. The pathomechanism of the deformities of the forefoot is not yet clearly known. The purpose of this study is to measure the dynamic changes of the anterior transverse arch of the foot and to find out which factors influence these dynamic changes of this arch as a footstep to clarify the forefoot abnormalities.

Methods

We used two conductible rubber sensors for this experiment of 30mm in length and 5mm in width. In this study, we divided the anterior transverse arch into the medial part (A) and the lateral part (B) and two sensors were fixed accordingly. The electrical changes of the rubber sensors caused by the expansion of the medial and lateral part of the arch were recorded. Simultaneously, the three components of the ground reaction forces as well as the foot switch signals were recorded. Then we studied the relationship of the anterior arch with the ground reaction forces and foot switch signals. The foot switches were attached to the sole of the foot in this order, the heel, the 5th and 1st metatarsal head, and big toe. The foot switch signals were described as follows, the foot flat time as FF, heel-off as HO, and the 5th metatarsal-off as 5MO. Thirty-two healthy adult volunteers were selected for this study. Eighteen males and 14 females whose age range was from 24 to 40 years with the average of 29.3 years. The sensors were attached on the left foot and the subjects were asked to walk freely on a force plate embedded 7 m walkway for 15 to 20 times.

Results and discussion

In the medial and lateral part of the anterior arch, there were 3 turning points. In respect of time, the 1st being at about 25% of the stance phase which are A1, and B1, the 2nd turning point at about 45%, A0 and B0, the final turning point, at about 75%, A2 and B2. There were no significant differences between male and female. In the time relations during the 1st turning point, the time of A1, B1 and Z1 had no significant difference between them. In the relationship shows there were positive correlations only between A1 and Z1, and B1 and Z1. From these relationships we can suggest that both A1 and B1 were dependent on the vertical component Z1, that is the load. During the 2nd turning point, there were no significant time relations between A0, B0 and Zo. But correlations were seen between Z0 with A0 and B0, which is similar to that of the 1st turning point. From these relationships we can suggest that both A0 and B0 were dependent on the vertical component Z0, that is the load. During the final turning point, there were no significant differences of time between HO and B2, and between A2 and Z2. At the time of the late stance phase, that is just after the heel-off, A2 had a positive correlation with Y2 and Z2. B2 had a positive correlation with Z2 and HO. From these relationships we can suggest that both A2 and B2 were dependent on the vertical component Z2, that is the load. In addition, the counter-action of the lateral thrust (Y2) works on the 1st metatarsal head which forces the medial part to widen further. In conclusion from this study we can say that during walking, in the stance phase the expansion of the anterior arch is mainly due to the vertical force, that is body weight, and the other extrinsic forces acting on the forefoot, not the muscular activities.

STANDARDISATION OF PROTOCOLS FOR FUNCTIONAL EVALUATION OF POSTURE AND LOCOMOTION DISORDERS: CAMARC-II: A PILOT OF A EUROPE-WIDE NETWORK

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Among the factors hindering clinical application of Movement Analysis (MA) for the Functional Evaluation (FE) of postural and motor disorders, a crucial importance is related to the lack of communicability of the results among different clinical laboratories. This problem can be faced by the integration of the existing istrumentation and the development of proper software tools and, mainly, looking at the concertation of common clinical and experimental protocols (CEP) and at the definition of unified clinical reports (CR) among the interested clinical sites. This will be also the basis for connecting in a network centres aimed at the treatment of similar diseases.

The above topics have been treated in a structured way at the level of the EC by the Project CAMARC (Computer Aided Movement Analysis in a Rehabilitation Context), that ran under the Exploratory Phase of the AIM (Advanced Informatics in Medicine) Programme, from June 89 to May 91. It demonstrated the feasibility of a Network of Clinical Laboratories and Research Centres sharing agreed CEP, instrumental techniques and software packages, connected with the manufacturers of the relevant instrumentation and able to develop new tools and new disability oriented protocols.

CAMARC-II is the project, presently running under the CEC Programme AIM, aimed at the pilot implementation of the above network .It started the 1st January 1992, with a three years temporal perspective and a manpower of 70 man years. It gathers twenty-eight participants among health care, clinical , research institutions and industries from eight Countries of the European Community and from two EFTA Countries. Its main objectives are:

- to build-up a pilot of a Europe-wide network of clinical and research laboratories willing to practice MA in a standardised manner;
- to define, on the largest consensus basis, protocols for data capturing and processing, .comprehensive of the assessment of measurement and data processing methodology , the development of suitable new instrumentation and motor tasks disability-oriented;
- to integrate the existing and some new instrumentation by means of suitable hard and software interfaces ;
- to define a comprehensive Knowledge Base (KB) of the already assessed MA experience and to implement a suitable Representation of such a Knowledge;
- to constitute prototypes of KBs using quantitative data relative to clinical applications and to make them accessible at the largest interested community;
- to assess criteria for the definition of normative data for a conventional age-related classification of normality, impairment and disability with respect to the motor behaviour.

The implementation of the pilot implies the preliminar concertation among the clinical sites for the standardisation of FE procedures: five clusters of centres have been constituded, respectively devoted to the assessment of CEP for the FE of patients suffering from Cerebral Palsy, Stroke, Lower Limb Amputation, Degenerative Knee diseases and Multiple Sclerosis.

Network implementation is meant as a two step procedure: in a first phase a star-like configuration, with only one Centre responsible for the reliability and safety of the data., successively a truly distributed network will be built-up. The physical support will always be an already existing computer network.

WHAT DO THE STATIC MEASUREMENTS COLLECTED ON RADIOGRAPHS BECOME IN A DYNAMIC SITUATION - APPLICATION TO GAIT ANALYSIS

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In the case of lower limb pathologies, the patient is radiographed in frontal view. On this radiograph, the mechanical axis of the lower limb can be defined by a line dr~wn between the head of the femur and the center of the ankle joint. Displacement of this axis is measured by the normal distance of this line from the knee center.

A new method has been recently developed in our laboratory. It allows to locate the rotation centers of each joint (hip, knee, ankle) of a moving subject, using trajectories of surface markers fixed on the lower limb. Then, new measurements can be proposed in a dynamic situation corresponding to the stance phase.

The dynamic mechanical axis can be drawn in the three dimensional space, linking the rotation centers of hip joint and ankle joint. This axis moves during the whole stance phase, describing a ruled surface. A shifting vector can be drawn, at each time, between the knee center and the dynamic mechanical axis.

The component of this shifting vector in the frontal plane is a dynamic notion. It coincides with the normal distance measured in static on radiography. The component of this shifting vector in the sagittal plane is a new notion introduced by the dynamic situation of gait. These two notions allow the study of dynamic instability in the knee during walking.

The method is presented and discussed. Two examples are proposed: a normal subject and a patient suffering from osteoarthritis.

INTERMUSCULAR CROSS-TALK OR CO-ACTIVATION IN SURFACE EMG? FOURIER PREDICTIONS AND PARTITIONINGS

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Whilst surface electromyography (EMG) is a convenient procedure for the identification of the presence or imminence of muscular actions a potential confounding factor is the inadvertent pick-up of action currents propagated by volume conduction from muscles other than those of immediate interest. This lack of selectively in signal origin (i.e. cross-talk) becomes particularly relevant when the question of relative involvement of synergistic or antagonistic muscles is being examined. For example, on the basis of EMG findings it has recently been suggested that coactivation of quadriceps and hamstring musculature is both normal and desirable in aiding knee joint stability; however, it has also been suggested that such findings may to a large extent be accounted for by cross-talk.

Procedures for minimising cross-talk within a myoelectric signal recorded by means of surface electrodes have been previously documented and include reduced inter-electrode distance and double differential signal processing. However, both methods compromise to some extent the quality of the myoelectric signal of interest, and, under normal physiological conditions, it is not possible to quantify the extent of cross-talk reduction.

This paper outlines a Fourier analysis procedure whereby the degree of cross-talk between two or more simultaneously recorded myoelectric signals can be identified and signal purification can be achieved without additional instrumentation or compromise in the quality of the available signal.

The utility of this procedure has been demonstrated for myoelectric signals recorded from bipolar surface electrodes (2cm separation) overlying vastus lateralis, vastus medialis and biceps femoris muscles of healthy adult subjects, while they performed knee extension/flexion exercises on an Ariel computerised dynamometer. Data were analysed using TSA† software.

The removal of cross-talk is based on a non-parametric estimate of the transfer function associated with the cross-talk and uses the smoothed cross-periodogram. If the signal of interest is $y(t)$ and the signal which may be contributing cross-talk is $x(t)$, then the Fourier transforms

$$X(\omega)=\sum_{t=1}^{N} e^{i\omega t}\,x(t) \quad \text{and} \quad Y(\omega)=\sum_{t=1}^{N} e^{i\omega t}\,y(t)$$

are formed and the transfer function estimated as

$$\beta(\omega)=\frac{\sum_{\lambda} Y(\lambda)\bar{X}(\lambda)}{\sum_{\lambda} X(\lambda)\bar{X}(\lambda)}$$

where the sum is over λ in a narrow band of frequencies around ω. The corrected Fourier transform $\hat{Y}(\omega)$ is then $Y(\omega) - \beta(\omega)X(\omega)$. Inverse Fourier transformation then yields a corrected form of

$$\hat{y}(t)=\frac{1}{2\pi N}\sum e^{-i\omega t}\,\hat{Y}(\omega)$$

This procedure is analogous to fitting a regression model and then correcting for the effect of the independent variable. Doing this in the frequency domain automatically allows for time delays and frequency dependent attenuation of the cross-talk.

† Henstridge, J.D. *TSA: An Interactive Package for Time Series Analysis*, Data Analysis Australia Pty. Ltd., P.O. Box 630 Nedlands, 6009, WA.

GAIT ANALYSIS AS A METHOD OF EVALUATION OF CRUCIATE LIGAMENT RECONSTRUCTION

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The aim of this pilot-study is to evaluate the functional results of anterior cruciate ligament reconstruction by patellatendon replacement, by pre- and postoperative gait analysis. Gait analysis is performed by two objective methods: VMCMAS (Video Marker Computer Movement Analysis System) and CDG (Computer Dyno Graphics). The first method records the angles of hip, knee and ankle, the second measures temporal factors and foot reaction forces. All patients (N=4) walk on a motordriven treadmill with 4 variable speeds. For each patient the most comfortable speed has been determined (Si). The other given speeds are: S1=Si+20%, S2=Si+30% and S3=Si-30% .
Range Si (1,094-1,502 m/s).
Pre-operative results of the objective and subjective test are: Lachman test 8-12 mm (1-3+), Pivot shift grade 2-3, Anterior Drawer test 7-11 mm, Giving away feeling.
Post-operative are these results negative. All the patients had a diagnosis of a rupture of the anterior cruciate ligament. The average age of the patients was 28 years (range 22-32). Three were male and one, female. There were three right anterior cruciate ligament and one left anterior cruciate ligament reconstruction. During walking video recordings register the movements. These records are computed by the VMCMAS-method to obtain the flexioncurves of hip ,knee and ankle, where as the CDG system monitors the temporal factors, the vertical footreaction forces figured as a cyclogram. This cyclogram shows the position of the projection of the center of gravity force of the body.
The results of this study show in all (4) pre-operative recordings that the affected leg shows a shorter stancephase, a lower ratio double support/stance, a lower maximum of the footreaction force, a longer time to reach this maximum, and in the cyclogram a displacement of the projection of gravity to the unaffected leg.
The first post-operative measurements are made 10-14 weeks after the surgical intervention. The results show that the leg with the anterior cruciate ligament reconstruction shows a longer stancephase, a higher ratio double support/stance (0,2), and in the cyclogram a displacement to this leg.

GAIT ANALYSIS AS A METHOD OF EVALUATION OF CRUCIATE LIGAMENT RECONSTRUCTION

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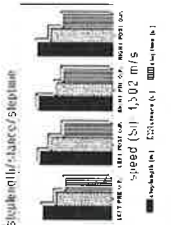

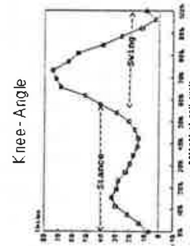

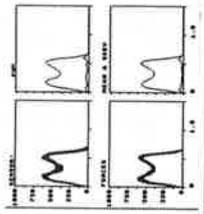

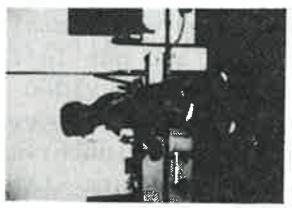

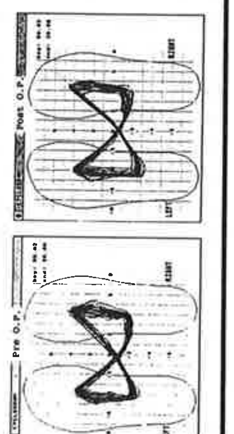

RIB CAGE-ABDOMINAL MOTION DURING RESPIRATION: 3D ANALYSIS

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The actual techniques for quantitative analysis of respiratory system, only allow a one dimensional measurement: the total inspired or expired volume. This quantity can not allow an analysis of the respiratory strategy. This work presents a new technique, absolutely non-invasive, which provides not only the measurement of the total volume of the air, but splits it into compartment sub-volumes (upper thorax, lower thorax, abdomen, right, centre, left).
The proposed method is based on the use of the ELITE motion analyser, able to compute the 3-D co-ordinates of several passive, lightweight and small markers applied to relevant body landmarks. The markers used for this application have a diameter of 8 mm.

The method has been validated for what concerns the accuracy in measuring air volumes, by a set of experiments in which the volumes were also assessed by a spirometer and a pneumotachograph. After this first phase performed in sitting position, the system has been applied to 5 supine subjects. Since these were also analysed while sitting, a comparison between the motor strategies of the same subject in these two postures has been made. In sitting position the contribution of the rib cage is predominant, while the same subject supine shows the predominance of the abdominal contribution to the total volume.

Also some pathological subjects have been considered. Two persons with ankylosing spondylitis have shown a clear reduction of the thorax mobility and thus volume. Three subjects with Duchenne's muscular dystrophy have shown a reduction in the abdominal component and a modification in the timing of espiration pattern.

These preliminary results seems to open interesting perspectives in the qualitative analysis of respiratory pattern. The clinical relevance involve the early diagnosis of fatigue of respiratory pump and the assessment of farmacological or ventilatory treatments.

This work has been supported by TELETHON Italia 1990

EFFECTS OF RE-EDUCATIVE TREATMENT OF HEMIPLEGIC GAIT: EVALUATION WITH DYNAMICAL SYSTEM APPROACH

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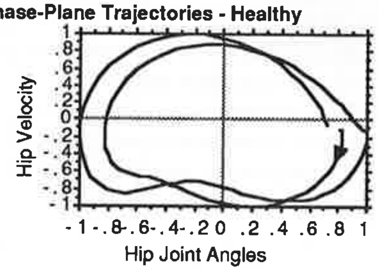
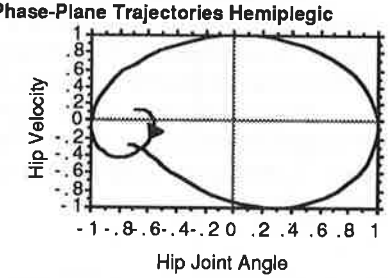
Physical therapy is considered effective in treating patients with alterations of motor behavior due to stroke. Effectiveness of treatment procedures, however, is generally referred in terms of variations in functional level rather than of actual changes in motor coordination of paretic limbs. Dynamic pattern theory, developed by Kelso JAS and collegues (1988), emphasizes how motor behavior is the result of the complex interaction of multiple subsystems, has identified sets of variables which could be used to assess a system's behavioral dynamics. We studied the characteristics of some kinematic variables of gait in eight patients (60-72 years) with hemiplegia due a cerebro-vascular accident (time interval from stroke 6-12 months) before and after re-educative treatment. Five healthy control subjects (62-71 years) were also studied. Subjects were requested to walk along a 9 m path at their own speed. Hip, knee and ankle joint angular amplitudes and velocities were sampled (50 Hz) with a video-based 3-D motion analysis system (ELITE). Twelve well performed trials were recorded in each subject. Off-line, for each trial and for each joint, angular amplitudes were plotted versus angular velocities thus obtaining time-independent representations (amplitude-velocity phase plane trajectories) in order to assess the variability of coordination patterns. In normal subjects, intra- and inter-individual variability of amplitude-velocity phase plane trajectories of hip, knee and ankle was small. On the other hand, patients always showed patterns quite irregular in their cyclic characteristics with marked intra- and inter-individual variability. After treatment, patterns displayed a reduced intra-individual variability in all the patients and five of them presented patterns similar to those recorded in normal subjects. Present results show that patients have difficulty in generating rhythmic movements and that physical therapy induce amelioration of gait characteristics.

Phase-Plane Trajectories - Healthy



Phase-Plane Trajectories Hemiplegic



Dynamic pattern theory of movement coordination have identified guidelines for identifying critical variables for assessing motor coordination and influencing pattern change. We have used the measures as the phase angle and relative phase as a order parameter in motor behavior description.

INITIAL GAIT ANALYSIS

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INTRODUCTION

A whole study of gait must includes the initiation of gait, the gait cycle and the stop of gait. The period of gait initiation is the transient period between early modification of upright posture and the stationary process of gait. Being involved with relatively high accelerations, the body has to apply a complicated system of muscular and neural control during the initiation period. By studying that period there is a good chance of identifying and diagnosing many disabilities in the locomotion system, including some of those that are not recognized in gait cycle. In this study, we aim at kinetic analysis of the center of gravity of the human body and the patterns of the reaction force and the center of pressure in the gait initiation.

METHODS

The subjects were 11 heathy, normal volunteers; 9 males and 2 female (23-29 y/o, average 25.5 y/o). The gait analysis equipment consisted of a 6 m walkway with a central located piezoelectric force plate (Model 9808, Kistler Instrument), an analog-digital converter (MP 280), and a computer system (IBM compatible 386 & 286 PC). All were barefooted during the test. At the beginning of each test the subjects was standing upright, balanced, arms folded within the force plate. Vocal order was used as a start signal. The force plate data were collected for 5 s. Each subject was tested three times at least.

RESULTS

In the beginning of gait initiation, the COP (center of pressure) moves backwards and towards the swing leg. The COP attains its maximum rightward movement (towards the swing leg) and then it begins to move towards the stance leg. The initial shift posteriorly and to the swing leg initially is slow, following which there are more rapid and uniform posterior progression. At the point of change of direction, there was a decreased rate of progression of the COP. The shift of the line from near the swing leg or near the stance leg was carried out rapidly. After which there was a fairly uniform forward progression of the COP until the point of toe-off of the stance leg. The trajectory of the late forward progression of COP of gait initiation is also similar to that of the gait cycle. The whole duration of gait initiation (1.205±0.124 s) is longer than that of gait cycle (0.739±0.069). At the beginning, the swing leg carried the major part of the load and the COG (center of gravity) was "falling" towards the stance leg and forwards the direction of progression. The deviation of medio-lateral displacement of COG was larger than those of fore-aft and vertical displacement of COG. The velocity of the direction of progression showed gradually increasing. At the end of gait initiation, there was about 95.8±19 cm/s in velocity, which was greater than the velocity of normal gait (88 cm).

CONCLUSION

At the beginning of the initiation of gait, COP moves backwards and towards the swing leg, but COG is falling towards the stance leg and forwards in the direction of progression. This unbalance phenomena takes the risk of falling, but is also essential to walk. The vertical displacement of COG in the period of initial gait is larger that in the period of gait cycle. From the pattern of reaction forces, the trajectory of the COP and the energy variation, the transition of the initiation occurs rapidly within first step.

LONG TERM AMBULATORY ASSESSMENT OF MYOELECTRIC SIGNAL ACTIVITY

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(2) Department of Biomechanics, University College of Health Sciences, Jönköping, Sweden.

A portable, battery powered, device capable of performing real-time, on-site, data acquisition and analysis of surface electromyographic (EMG) signals has been developed. The device, called MyoGuard, is easily carried by the waist. All signal conditioning and processing units needed to obtain Root Mean Square (RMS) and Mean Power Frequency (MPF) from the raw EMG signal are contained in a box of 200x110x56(mm) size. The RMS parameter gives information about the muscular involvement. Changes of the MPF parameter are a sign of localized muscle fatigue. Since these parameters are calculated in real time, it is possible, given the right criteria, to continously inform the subject about the current condition of the muscle being monitored. A monitoring session can be in full progress up to eight hours. The RMS and MPF parameters are stored on a semiconductor memory card every second. The memory card makes it possible to transfer the collected data to a personal computer for further statistical analysis and presentation. This enables us to collect data of muscle activity and local muscular fatigue from a whole working day.

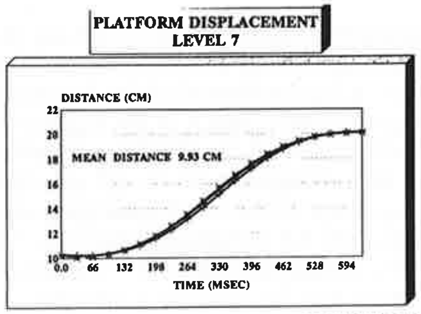
The EMG-signal is recorded by means of surface electrodes, amplified in a amplifier with automatic gain control (AGC), and converted to digital form to enable calculation of RMS and MPF. The MPF parameter is determined from the power spectrum, the result of a fast fourier transformation (FFT) of the EMG. The spectral information available from the power spectrum also facilitates detection of disturbances from the power mains. When such a disturbance is detected, its impact on the analysis can be reduced by means of an interpolation algorithm applied to the spectrum. The automatic quality control and correction algorithm makes the processed data presented by the equipment reliable.

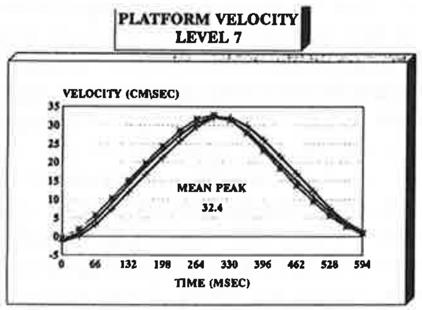
Application of the device is initially focused on static or semistatic work where strain is put on the neck and shoulder region, particularly the trapezius muscle since it is a common site for work-related myalgia. However, the method will probably apply to any superficially located muscle.

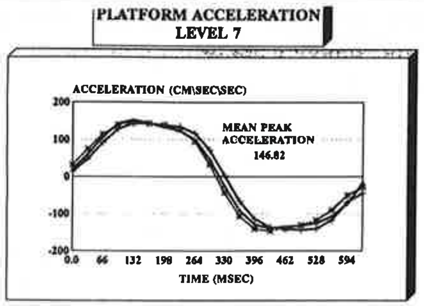
VALIDITY AND REPRODUCIBILITY OF MOTIONSPECtm PLATFORM MOVEMENT KINEMATICS

Anderson PA, Alon G, Smith GV; University of Maryland, School of Medicine, Department of Physical Therapy, Baltimore, Maryland, U.S.A.

This study was conducted to test the validity, reliability and accuracy of selected kinematic variables obtained during motion of the Motionspec™ Balance and Motor Control System. Three reflective markers each 5 mm in diameter were placed on the platform, one on the moving plate and two on a non-moving parts of the system. The movement of the platform was videotaped using 2 Magnavox video cameras. Three different movement profiles were filmed, each profile recorded 3 times. The 2 videotapes were then digitized using the Ariel Performance Analysis System™ and the data were transformed, smoothed by digital filter (4 Hz), and the linear distance, velocity and acceleration were calculated. Results indicated that Motionspec prescribed distance of 10 cm ranged between 9.77 and 9.93 and the peak velocities programmed as 10, 35, and 38 cm/sec were in effect 10.07, 32.4 and 35.5 cm/sec respectively. The prescribed peak accelerations of 35, 125, and 150 cm/sec^2 were in effect 66.97, 146.82, and 176.98 cm/sec^2 respectively. The data obtained for level 7 is illustrated in the 3 figures. The discrepancies between the prescribed and measured velocities and accelerations ranged 7.4 to 91.3 percent. They were attributed in part to the limited accuracy of the digitizing system. The coefficients of variation for displacement, velocity, and acceleration were .01, .09, and .01 respectively. Reliability of each movement profile was very high as indicated by intra-trial correlations coefficients ranging from r=0.9992 to r=0.9997. We concluded that within the limits of our testing the BTE Motionspec™ system movement kinematics are valid and highly reproducible.







MULTIVARIATE ANALYSIS OF MOTIONSPECr MOVEMENT KINEMATICS

P.A. Anderson, G. Alon and G.V. Smith - University of Maryland, School of Medicine, Department of Physical Therapy, Baltimore, Maryland 21201-1587 (USA)

Falls annually account for thousands of serious injuries and millions of health care dollars. Despite the magnitude of the problem, especially among the elderly and neurologically-impaired individuals, we are only beginning to understand the intricate neuromuscular processes involved in maintaining human equilibrium. Unfortunately, even this knowledge is difficult to transfer to the clinic, where valid, reliable and reproducible measures of balance heretofore have been arduous, if not impossible to obtain. The goal of our research over the past several years has been to develop clinically-useful means of measuring performance parameters related to balance. The present study is one such effort.

Previous studies conducted in our laboratory utilizing the MotionSpec[R] Balance and Motor Control System have employed kinematic variables of 10 cm displacement, 38 cm/sec peak velocity and 125 cm/sec^2 peak acceleration of the platform. These values were chosen arbitrarily on the basis that they were consistently able to interfere with the subject's balance. Previously, other combinations of kinematic variables were not tested systematically. The current study examined 60 combinations of displacement, velocity and acceleration and their ability to disrupt balance. Data were collected from twenty healthy subjects (10 male and 10 female, mean age 24.2 $\pm$ 8.2 years, mean height 64.2 $\pm$ 9.5 in, mean weight 157.0 $\pm$ 50.6 lbs) on the reaction times of the left and right foot when table movement elicited a momentary loss of foot contact. The subjects, wearing custom-made shoes with built-in pressure-sensitive electrical switches, were asked to stand on the platform perpendicular to the direction of movement. Computer-generated signals induced platform movements according to the pre-determined parameters. These movements, in the proper combinations, would elicit a stereotyped balance reaction. The raw data was subjected to a repeated measures analysis of variance (ANOVA) with an alpha of 0.05 employed for significance.

The data indicated the main effects of displacement (F=0.0095; p=0.0075), velocity (F=0.0025, p=0.0027) and acceleration (F=0.0001; p=0.0001) were significantly different. However, none of the interactions between these variables were significant. A Student-Neumann-Kuels Post Hoc analysis indicated that a 6 cm displacement was significantly different from a 12 cm displacement. Our analysis revealed that peak velocity was a direct function of the distance and acceleration. Additionally, the acceleration data indicated that an acceleration of 50 cm/sec^2 was significantly different from all other acceleration choices. Combinations of displacement, velocity and acceleration at or below 6 cm, 15 cm/sec and 50 cm/sec^2 respectively are inadequate to cause a loss of foot contact by a healthy subject on the MotionSpec[R] Table. Combinations above these values consistently produce balance disruption. These results suggest that the previous parameters of 10 cm displacement, 38 cm/sec velocity and 125 cm/sec^2 are well within the envelope of combinations necessary to elicit a stereotypical balance reaction.

KINEMATIC AND DYNAMIC STUDIES OF JOINTS IN WALKING PATIENTS FROM EXTERNAL SIGNAL TREATMENT

J. Dimnet, L. Cheze, J.P. Carret

Numerous systems exist alloying the analysis of human motion. They consist in markers (emitting or reflecting marKers) fixed on body segments. The marker trajectories in the fixed frame are computed through data treatment of several camera images viewing the markers. These marker trajectories are commonly used describing global gait of subjects. The goal of our laboratory in Lyon is an attempt to define kinematics and dynamics of internal joints from trajectories of markers fixed externally. The corresponding research covers two areas:
- the theoretical group applies techniques known in robotic identifying the instant analogue mechanism to the moving limb.
- the applied group is composed of medical doctors and biomechanical researchers. They study possible correlations between pathologies and observed disorders in motion.

The kinematic studies

Passing from external marker trajectories to internal three dimensional kinematics of joints. After defining an experimental protocol it is necessary to identify noises which disturb the trajectories. We identified 2 kinds of noises:
- the relative displacement of markers fixed on the skin versus the underlying soft tissues.
- the motion of soft tissues versus the adjacent bony structure. These two noises are treated separately. Then the curves of changes in the angles corresponding to the external motion are drawn. These curves disply high frequency noises. It is assumed that these noises are due to the interference of soft tissue displacements versus the underlying bone. A filtering process is applied giving both the angular changes of the internal joint and the noises due to deformable mass motions. The joint centers are then identified for each instant of time and the center paths in internal joint during motion are drawn.

The subject motion is then simulated

Bones have been taken from the same cadaver. The global shapes of each bone and insertion zones of muscular bundles have been identified an digitalized using a robot specially adapted for these measurements. A numerical data bank corresponding to both lower and upper limb has been constituted. It is assumed that the bony shapes in any subject are obtained from the data bank and adapted to the subject size using the kinematics laws obtained from the studied patient. For clinical applications in hip joint replacement we computed the femur displacements versus the fixed pelvis during the stance phase in gait. An analogue software will soon be available showing the knee joint motions during the stance phase.

The dynamical studies

Because of clinical applications we are mainly interested in calculating the articular forces in real time, while stance phase. This needs the knowledge of external load (at the level of a force plate) but also both the forces exerted by active muscles and the inertial forces. Presently our laboratory works in 2 separate directions:
- calculating the muscle forces
- estimating the inertial forces.

Calculation of muscular forces and articular forces

For each joint of the lower limb (hip, knee and ankle joints) we dispose of 3 curves of change (one per degree of freedom in rotation) let 9 W F for the entire lower limb. It is then possible to introduce nine pairs of muscles (agonist and antagonist). In a first step we assume that agonists muscles are only activated. So it is possible to compute, form the force plate signal nine muscular forces, when both instant positions of bony structure and instant geometric shape of muscles are known. For each position of the bony structure, it is necessary to define zones here muscular bundles insert bones and to model the geometric shape of muscular bundles as well for the monoarticular muscles as the biarticular muscles. From this modelling we can access the changing of the muscular length. This last one allows the calculation of the moment induced by the muscular force at the level of each degree of freedom of each joint. The resolution of an inverse problem gives the set of the nine muscular agonist forces necessary to resist to the external load at the level of the nine degrees of freedom. For clinical applications we computed the equilibrium of the hip joint while resisting to an external force of magnitude one applied at the level of the ankle joint. The software allows to take into account a set of three muscles chosen among a set of eight possible muscular bundles. For each combination of three muscles the program gives both the magnitude of muscular forces and that of articular force and corresponding point of application. In the near future inertial forces and action of antagonist muscles will be taken into account in dynamic modelling.

ANALYSIS OF GAIT PERFORMANCE: A MOVEMENT ANALYSIS SYSTEM IN HEMIPARETIC STROKE PATIENTS WITH A MODERATE LEVEL OF SPASTICITY

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U.S.L. 31 - Ferrara - Italy

The purpose of this study was to determine in 10 stroke patients the modification of the gait performance during the rehabilitation treatment aimed at reducing the spasticity of the lower limb. The patients (mean age of 60.3 years) had a moderate level of spasticity (between level 2 and 3 of the Ashworth Scale). The major gait abnormalities clinically seen in our patients in the sagittal and frontal plane were : (1) hip adductors spasticity causing a scissoring gait ; (2) rectus femoris/hamstrings cocontractions contributing to the stiff-legged gait and causing a reduction of the stride ; (3) an inappropriate foot pre-positioning in the initial swing ; (4) gastrocnemius-soleus spasticity causing foot equinus in stance.

The rehabilitation tratment in the acute phase was carried out by a theraphist and consisted in passive exercise and body postural alignment in bed. In the subacute phase the patients were included in a specific rehabilitation programme for the reduction of the spasticity of the lower limb by relaxation exercise with a Biofeedback Electromyographic device and for the reeducation of the gait. The patients who have been treated according to this rehabilitative protocol were evaluated with a stereophotogrammetric system (COSTEL) and with a force platform (BERTEC) .

Gait information available for the assessment includes : kinematic data and plots of the pelvis, hip, knee, and ankle in three planes (coronal, sagittal, and transverse) graphic presentation of ground reaction forces (GRF) generated during the stance phase of gait .

We are able to determine the following kinematic and dynamic variables : walking speed, stride lenght, time stance ; hip, knee and ankle angle at different phases of the gait.

The first evaluation was performed immediately as soon as the patients were be able to walk by themselves and the second evaluation was performed and 60 days after.

The results indicated the attractive possibility of employing movement analysis to the quantitatively assess of the gait improvement during relaxation treatment of Hemiparetic stroke patients.

ONE METHOD OF FAST MOTION ANALYSIS

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Behaviour and functioning of human locomotion was explored by many investigators especially for pathological cases. An exact analysis of locomotion in healthy subjects or in athletes is rather demanding or requires very sophisticated and complex instrumentation as well as methods of signal processing.

Locomotion produces various measurable signals which can be used for motion analysis, but nevertheless acquisition of these signals in fast, even normal motion is very complicated.

Here we present one method of fast motion analysis based on measurement and analysis of ground reaction force and myoelectric signals. Assuming a quality of performance to be a skill level criterion, acquired signals were mathematically processed in order to obtain quantitative indices of locomotor skill specific to the particular subject. This method of analysis allows us to create an intermuscular activities correlation matrix presenting sensitive indices of particular motoric skill level.

Taking as an example a typical gymnast locomotion, the backward somersault, proposed technique of analysis is explained and discussed in details. It is shown that using this relatively simple method of data acquisition and processing it is possible to quantitatively estimate the skill level of gymnast as well as to monitor the progress in training with high reliability.

A NEW TRIAXIAL ACCELEROMETER AND ITS APPLICATION AS AN ADVANCED INCLINOMETER

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In epidemiological studies of work-related musculoskeletal disorders there is a need for accurate and reliable methods to measure posture and movements.

The transducer consists of three commercially available miniature accelerometers (**Figure 1**) and can be taped to the skin (**Figure 2**). With a data logger (not shown in the figures), we can record accelerations in three directions for 20 minutes at 20 Hz. After transfer of data from the logger to a personal computer, amplitude distributions for acceleration in the three directions, as well as total acceleration is calculated. These distributions reflect the forces that act on the head due to gravity and dynamic accelerations. If the dynamic part is small, which is the case in sitting and standing work, we can calculate the inclination of the head (**Figure 3**).

The new triaxial accelerometer has a low weight and a high resonant frequency, it is accurate and easy to apply and has no restrictions concerning the range of movements. Since the total acceleration can be calculated, it is possible to test if the interpretation of the measurements as inclination is relevant, a feature not possible with an ordinary inclinometer.

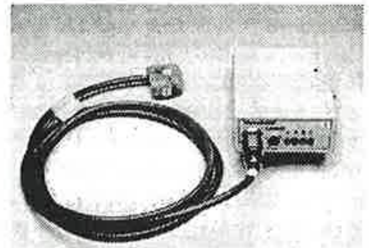

Figure 1: The transducer consists of three miniature accelerometers mounted perpendicular in a plastic cube. Each accelerometer senses both dynamic (due to velocity changes) and static acceleration (the gravity of the earth) in one direction. The battery powered electronics contain constant current sources for the accelerometers, instrumentation amplifiers, low pass filters and interface circuits to a data logger.

Figure 2: The transducer mounted with adhesive tape to the forehead. Since two reference postures are recorded, one with the head held straight and one with the head bent forward, the transducer coordinates can be transformed into head coordinates, and hence, the transducer may be mounted in an arbitrary position and orientation on the head.

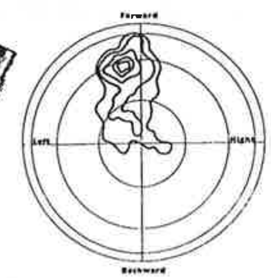

Figure 3: Contour plot of the two-dimensional distribution of head positions of a dentist treating a patient. Origin represents an upright position and the concentric circles an inclination of 22.5°; 45°; 67.5° and 90°. The graph shows that most of the time the head is bent steeply forward (40°) and slightly (10°) to the left, and that a short time is spent with the head in an upright position.

A VIDEO MARKER COMPUTER MOTION ANALYSIS SYSTEMS (VMCMAS)

A Video Marker Computer Motion Analysis System (VMCMAS).

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Many systems for motion analysis require large and expensive hardware systems, such as VICON and Selspot. For simple clinical and educational applications a VHS-video camera, a VHS video recorder and a personal computer with an overlay board are the components for a motion analysis system, which will work with software developed in an authoring system named TAIGA (Twente Advanced Interactive Graphic Authoring system).
The subjects are prepared with markers on the body segments which are studied. The movements are performed and recorded with the video camera. The recordings are displayed on the video recorder connected to the overlay board of the computer. In the computer the video signal is merged with the TAIGA software, thus giving the possibility of using the mouse as a pointing device to obtain the coordinates of the markers on the body segments. The video recorder has a still frame and a frame-by-frame function, for analysis of the individual video frames (25 frames per second). In the TAIGA software the angles between body segments can be calculated, and stick-diagrams can be drawn.
With this tool different motion analysis studies are performed in the field of ergonomics and gait analysis evaluations.

ELECTROMYOGRAPHY ANALYSE OF MUSCLES BICEPS FEMURIS (CAPUT LONGUM) AND SEMITENDINOUS IN MOVEMENTS IN PLANS DIAGONAL AND SAGITTAL

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F. Bérzin - Faculty of Odontology of Piracicaba - UICAMP.

The goal of this study was to compare the electromyography activity during the movements performed on the diagonal plan which characterize the patterns of the Kabat Method for the proprioceptive neuromuscolar facilitation with the movements performed on the sagittal plan usually indicated in the treatment and training plans. These movements were performed with and without the application of mechanical resistance by neans of an equipment called Double Pulleys System.
The lectromyographic records were obtained from the thigh semitendinous and biceps femoris (caput longum) muscles at 30, 60 and 90 degrees recorded by an electrogoniometer.
The patterns of movement in diagonal plan were: 1) extension, abduction and medial rotation of the hip; knee flexion, plantar flexion with eversion of the ankle flexion and adduction of the toes; 23 extension adduction and lateral rotation of hip, with knee flexion, plantar flexion with inversion of the ankle flexion and adduction of the toes. In both movements the volunteer was in a dorsal decubent position; 3) the movement performed on the sagittal plan was the knee flexion with the subject ventrally decubent.
Through this study it can be concluded that the thigh biceps femoris (caput longum) and semitendinous muscles show a greater electromyographic activy in the flexion movement of the knee on the sagittal plan in comparison with the movements performed on the diagonal plan.
The thigh biceps femoris (caput longum) and semitendinous muscles present a greater activity when submitted to load application.
During the three movements, the this biceps femoris (caput longum) and the semitendinous muscles present a greater activity at 90 degress when the angles affect is observed.
Of the movements use by the Kabat Method, the movements of extension, abduction and medial rotation of the hip, with knee flexion, plantar flexion and eversion of the ankle, flexion and adduction of the toes, presented a greater activity of the thigh long head biceps femoris (caput longum) muscle at the angle of 90 degrees.
For the semitendinous muscle, both patterns of the Kabat Method, showed similar activity both at 30 degrees and 60 degrees, and has a greater activity at 90 degrees in the M1 (extension, abduction, medial rotation of the hip knee flexion, plantar flexion with eversion of the ankle, and adduction of the toes).

ACTIVATION OF LUMBAR 1A INHIBITORY INTERNEURONS (IA IN) BY MAGNETIC CORTICAL STIMULATION (MCS) IN MAN

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Introduction : It is well known that spinal motoneurons are activated by MCS over the contralateral cortex via large diameter corticospinal fibres. This can be measured by the evoked muscular response in limbs. However, mainly based on animal studies, corticospinal fibres are known to project to various spinal interneurons cells, namely the 1A IN. The latter is assumed to mediate short latency reciprocal inhibition (RI). This study was designed to investigate the change in 1A IN (at the lumbar level) recorded in the lower limb after MCS of the motor cortex.

Methods : The MCS was studied in 12 healthy subjects, aged between 20 and 40 years. Physiological studies on their lower limb included a) EMG recording using a pair of surface electrodes on the TA, SOL muscle during the MCS; b) H-reflex recording (H/2) at rest (control) and during the MCS (H-reflex taken as time 0) given at various delays; c) measure of reciprocal inhibition (using modified TANAKA technique) a 2 ms interval delay between stimulation of the common peroneal nerve and H-soleus reflex at rest and during MCS (MAGSTIM MODEL 200 with coil 90 mm); d) every interval delays randomly tested from -6 to 2 ms.

Results and discussion : 1) MCS threshold to evoke TA, SOL muscle responses : 75 ± 15% output % display or 1.13 ± .25 telsa. 2) MCS output % display in this testing to facilitate the H-reflex: 70 ± 15% or 1.05 ± .25 telsa. 3) conduction time: for H-sol reflex : 31.8 ± 1.8 ms, for MCS to TA: 30.1 ± 1.7 ms, for MCS to sol : 30.9 ± 1.7 ms. The MCS induced an activation of the 1A IN (Fig. 1-2) in the lumbar spinal cord. The maximal effect was observed at -3.75 ± .3 ms by 27 ± 5%. But when we corrected for the residual reciprocal inhibition (mandatory when considering simultaneous facilitation of H-sol reflex) to the net cortical effect on 1A IN was 17.5 ± 5% ($P < 0.01$). However, the facilitation of H-sol reflex by MCS was observed within a mean delay of 2.25 ± .8 ms. that is 1.5 ms later then MCS on 1A IN.

The corticospinal fibres activate the lumbar 1a IN. This probably occurs earlier than motoneurons. This observation can be interpreted as being caused by the difference in the size of the spinal cells: the smaller interneurons could be fired by a single spike in the corticospinal fibres, while the large motoneurons if depolarized by a single spike, would need a summation to fire.

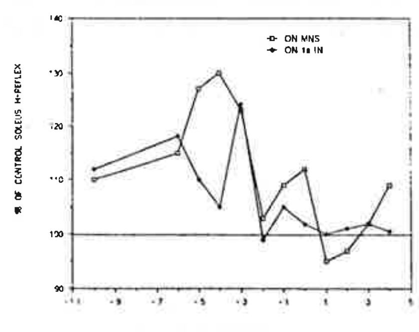

Fig. 1 - This figure shows (black squares) the facilitation of H-sol reflex by the MCS and (white squares) the effect of MCS on RI (Ia IN) coming from flexor antagonist muscle (TA) to the extensor muscle (Sol).

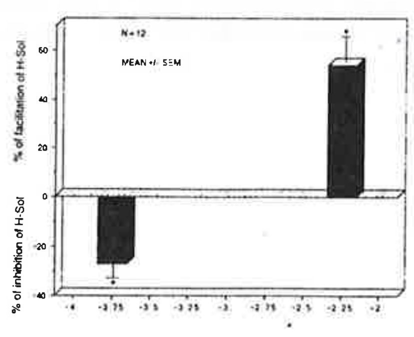

Fig. 2 - The upper panel represents the mean of the higher facilitation of H-sol reflex by MCS (54 ± 9%) and the mean of the delay produced (- 2.25 ± .8 ms).

The lower panel shows the maximal facilitatory effect on 1A IN by MCS (27 ± 5%) and the mean of the delay produced (- 3.75 ± .3 ms).

REFLECTIONS ON MOTOR UNIT ACTION POTENTIALS

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Abstract. Motor unit action potentials are recorded and observed with relatively little attention for quantitative aspects. Clinically some parameters are used for the detection of diseases. Can it be expected that quantitative analysis will be more relevant in the future? This question will be considered after presentation of recent data of single fibre action potentials. Additional data has to be gained before the message of motor unit action potentials can be understood.

1. Introduction

An active motor unit will generate a motor unit action potential (MUAP). The shape of a MUAP is determined by structural and electrophysiological characteristics of the motor unit, its constituent fibres and the muscle tissue and by electrode properties. Recordings of electrical activity of skeletal muscles contain MUAPs, but generally these action potentials cannot be observed directly. Nearly always the EMG signal is a compound one with activity of several motor units. The complexity of the electromyographic recording depends on the level of activity and on size and location of recording electrodes.

The study of MUAPs is potentially of interest when it is possible to record them at different levels of normal muscle activity. There are a number of techniques to isolate MUAPs from EMG recordings in human or animal muscle, even when several motor units are simultaneously active. A selection of techniques developed in the last decade will be described.

Quantitative analysis of MUAPs has been the subject of especially clinical papers, e.g. [1]. Clinically most relevant parameters appear to be the area, duration and number of phases of the MUAP. The main part of this paper consists of a critical evaluation about the processes and structure at the origin of normal, non-fatigued MUAPs, mainly based on the state of the art in research of single fibre action potentials. In the concluding part we focus on quantitative analysis of MUAPs.

2. Techniques to record motor unit action potentials

2.1. A variety of trigger methods

A signal at a single fibre electrode of one of the muscle fibres belonging to a motor unit can be used either to observe the MUAP directly (at low level of activity) or to calculate with triggered averaging the action potential of that particular motor unit.

Stålberg et al. [1] introduced the so-called scanning electromyography. MUAPs of one motor unit were followed by pulling a concentric needle in small steps upwards through the motor unit territory. Scanning electromyography provides a way to get information about global fibre density, fibre grouping and total dispersion in arrival time of the activity. The method is time consuming and needs cooperation of the person. Per person only one or a few motor units can be traced.

Hermens [2] used an intramuscular single fibre trigger to obtain MUAP surface recordings. Mono- and bi-polarly recorded MUAPs were compared. As expected the general shape of the monopolar recordings is triphasic and that of bipolar recordings biphasic. With monopolar electrodes also biphasic signals were found, probably due to end plate and tendon effects. The position of the electrodes has to be carefully chosen.

Masuda and De Luca [3,4] used triggered-averaging in combination with needle signal decomposition mainly to derive conduction velocities. The method also enables the observation of MUAPs in relation with the order of recruitment. Their figures display the general trend of small, slow MUAPs for early recruited motor units and larger, faster MUAPs at higher force levels, in agreement with the general ideas about motor unit recruitment and electrophysiological properties of the muscle fibres.

Using a multi array of electrodes on the skin above muscles the position of the endplate zone(s) can be found [3,4,5]. When representative samples of MUAPs should be obtained the endplate area has to be avoided.

2.2. Microstimulation of motor units

Andreassen and Arendt-Nielsen [6] applied a bipolar microstimulation method to evoke the activity of a single motor unit in large human muscles. By averaging, the MUAP and also the mechanogram of the motor unit were obtained. A fast-slow discrimination of the motor unit is then possible based on the twitch. Fatiguing patterns appear not to be suited for further characterization. MUAP characteristics have not been studied.

In contrast with natural activity there is a preference with this method for large motor units because of their relatively large axons. In fact this method gives a good opportunity to match quantitative analysis of the MUAP to mechanical properties (fast or slow) of the motor unit.

2.3. Decomposition of myoelectric signals

There are several techniques to decompose myoelectric signals in MUAPs, e.g. [7]. The decomposition itself is still subject of study. The techniques are promising for reliable derivation of MUAPs from compound signals and allow MUAPs to be studied quantitatively.

3. Structural and electrophysiological aspects of motor units and MUAP calculations

3.1. Structural aspects

The spatial distribution of fibres belonging to one motor unit, e.g. [8] is an important structural aspect for the understanding of MUAPs. The only direct way to observe the fibre positions of one motor unit is the glycogen depletion method [9]. An example of the position of fibres belonging to a motor unit is given in figure 1.

Fibres of a normal motor unit are not compactly grouped together. In small

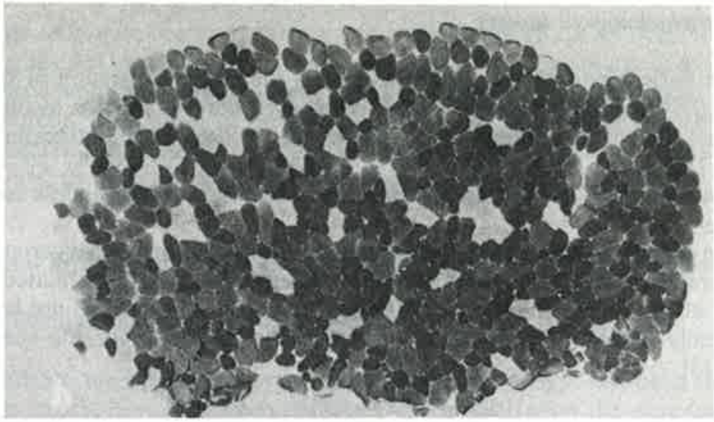

Figure 1. A cross-section of M. extensor digitorum longus (EDL) of a mouse. The glycogen of the muscle fibres of one motor unit has been depleted (light coloured fibres). An improved histochemical procedure enhanced the contrast between glycogen depleted and non-depleted fibres [10].

animal muscles they are sometimes found in the whole muscle cross-section. In large human muscles the diameter of the motor unit territory can be up to 10 mm. [1].

Each fibre is wrapped in a thin layer of connective tissue, while groups of fibres are surrounded by layers of connective tissue of variable thickness. Blood vessels and nerve bundles occur frequently; their sizes are variable.

The extracellular volume fraction plays a major role in current distribution. Freezing, drying of the sections after cutting and staining procedures may contribute to an enlargement of the extracellular volume. The derivation of reliable values of it is troublesome. The extracellular volume fraction in cryosections without staining is estimated to be about 10% in our material; this value is in agreement with other data [11].

For a three-dimensional reconstruction of a motor unit in a muscle a large set of cross-sections is necessary. The cross-sections have to be stained with different protocols to get information about the position of endplates (longitudinal spread in the position of the endplates, multiple presence of endplate zones along the longitudinal axis of the muscle), the length of the muscle fibres between endplates and the tendon at both sides of the muscle and the presence of connective tissue. In figure 2 some structural aspects of a motor unit are schematically drawn.

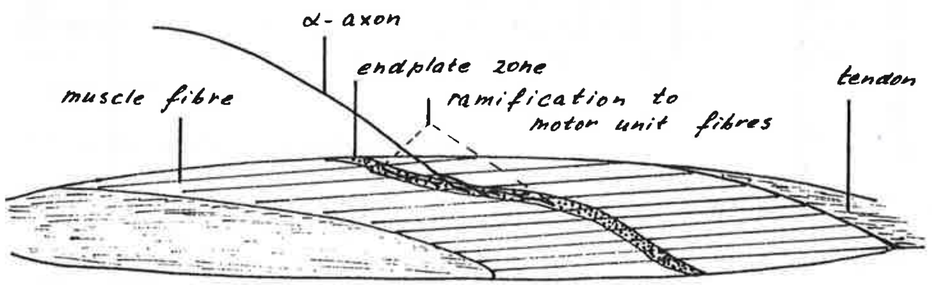

Figure 2. A scheme of main structural aspects of muscle for MUAPs.
The lateral side of the M. extensor digitorum longus of the rat has been taken as an example. The ramification of one axon has been drawn to illustrate the superficial area of one motor unit. The width of the endplate zone is about 0.3 mm. The muscle fibre length at optimal twitch length is about 15 mm.

3.2. Electrophysiological aspects

The most basic aspect of the MUAP concerns the current injection of the muscle fibres in the surrounding tissue. In the often used core conductor assumption the membrane current is related to the second derivative of the intracellular action potential [12].

An example of a propagated intracellular action potential and its second time derivative is given in fig. 3.

An action potential is propagated along the muscle fibre with a certain velocity. It is generally assumed that the conduction velocity is positively correlated with the diameter of the muscle fibre (e.g. [13]), but the exact relation is still not known [14]. Theoretically the conduction velocity is an important factor for the shape of the MUAP [15].

3.3. Calculation of motor unit action potentials and comparison with experimental data

The motor unit territory, fibre density, a simple current injection per fibre and a constant conduction velocity are part of the MUAP model of Nandedkar et al. [13]. They analysed the contribution of the fibres closest to the electrode, and the effects of model assumptions on the area and the duration of the MUAP. These both aspects are related to the contribution of many fibres.

MUAP calculations for known positions of fibres with respect to the electrode are rare [16]. Here good agreement with experimental data has only been found for a fixed point of time at which the fibres located close to the electrode contributed to the signal and for an anisotropy value in volume conduction of 5 (found by trial and error to be optimal for the ratio of longitudinal and transversal bulk conductivities).

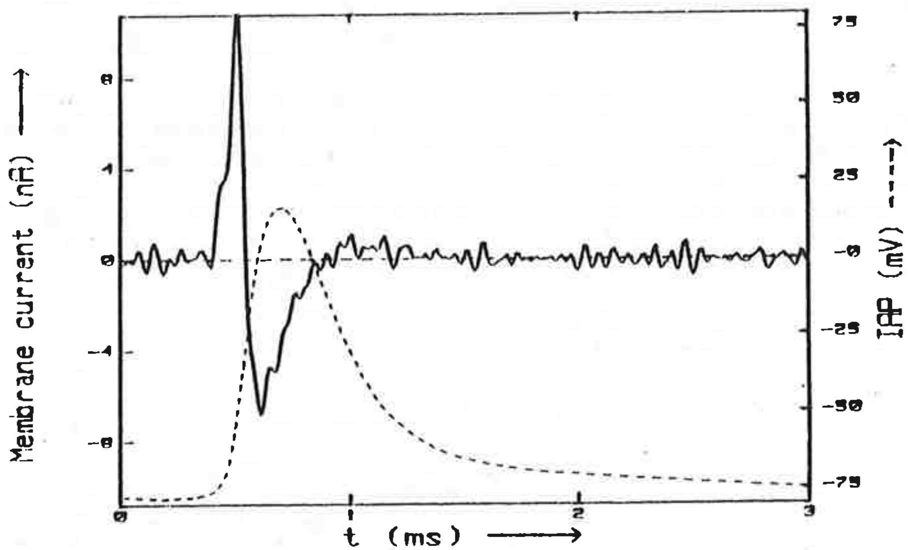

Figure 3. The dashed line shows a measured intracellular action potential, it is a representative example for a fast muscle fibre in the M. extensor digitorum longus of the rat (in vivo condition). The solid line displays the corresponding transmembrane current, computed from the intracellular action potential according to the core conductor theory.

Main differences between the recorded and simulated curves occur in the start and end of the MUAP. This has mainly been caused by the assumptions about the intracellular action potential (a simple triangle representation resulting in a tripole current source). Two lines of experimental research have been triggered by the MUAP-research of Griep [6] in our laboratory, one to explore intracellular action potentials and the other to study volume conduction properties of skeletal muscle.

4. Additional experimental data and its effect on single fibre action potentials

4.1. Intracellular action potentials and membrane currents

To gain reliable recordings of intracellular action potentials it has been pursued to measure in muscle fibres of known character (determined after the experiment), in a normal propagating condition (activity evoked via the endplate and recorded in the middle between the endplate and the tendon) and with small or no disturbance caused by activity of surrounding fibres. Such conditions are fulfilled when action potentials are recorded with fine-tipped, flexible micropipettes in a muscle fibre at body temperature and at in vivo circumstances [17], stimulating a motor unit [18]. The intracellular action potentials of fast muscle fibres are steeper in comparison with those of slow fibres in the rising and declining phase and have a higher mean amplitude. Main improvement with respect to older data (e.g. [19] is the fact that the action potentials are not disturbed by stimulation current.

When the second derivative of the intracellular action potential represents the membrane current, the core conductor theory is used. The extracellular medium is, however, not a very good conducting fluid in the in vivo condition. High extracellular potential recordings probably arise by poorly conducting medium around the fibres. Very exceptionally values up to 25 mV have been found [20]!

Recently a loose patch clamp technique has been adapted to measure the transmembrane current during a propagated action potential (figure 4) [21]. This transmembrane current has been measured at in vitro conditions, on a superficial fibre of a small mammalian muscle at room temperature, in a Ringer bath. Unfortunately it is not feasible to record transmembrane currents at in vivo condition with this method. But the membrane current during an action potential of a mouse muscle fibre in vitro offers the opportunity to estimate the membrane current of fibres in vivo. Its reliability cannot be indicated, because the effects of temperature and of in vitro versus in vivo condition are not known. The general picture of the membrane current in vivo is expected to be similar to the plot in figure 4 with a compression of the time axis with a factor 3.5 (based on comparison of the intracellular action potentials in both circumstances).

4.2. Volume conduction in skeletal muscle

Volume conduction descriptions nearly always concern a macroscopic approach. Fibres close to the electrode are mainly responsible for the peak-to-peak amplitude of the MUAP. So the microscopic volume conduction properties are important, at least for intraterritorial recordings. It is not easy to realize microscopic observations of volume conduction. Gielen et al. [23] developed a technique to derive properties at a scale of 1mm.: then volume conduction has a capacitive component. Current is partioned between the intra- and extra-cellular volume components dependent on the conduction properties of the components and their volume and passes fibre membranes. So it is

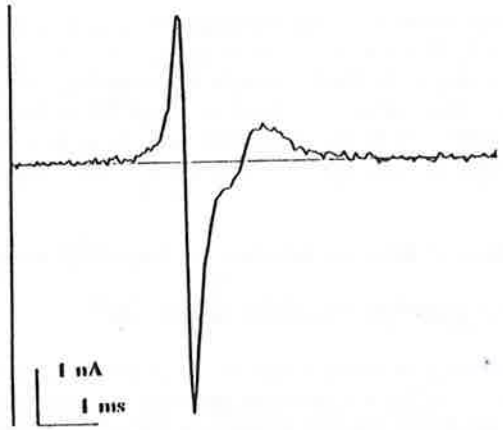

Figure 4. An example of a membrane current during a propagated action potential. The membrane belongs to a superficial fiber, normally surrounded by the other fibres of the muscle. The recording has been made with the loose-patch clamp technique [22]. Temperature 22° C, in normal Ringer solution.

not astonishing that the membranes introduce capacitive elements in the volume conduction [24].

4.3. Modeling of single fibre action potentials and comparison with experimental data

The effects of microscopic volume conduction properties [24] and of the shape of the intracellular action potential [25] have been tested in simulations and checked experimentally at single fibre action potentials [26].

Introduction of small scale volume conduction properties has an apparent effect upon the amplitude of the single fibre action potential when the recording is made within 300 μm from the center of the active fibre. Very close the amplitude is 1.2 [27] till 1.5 [24] times enlarged due to capacitive properties of the membrane, dependent on the chosen model parameter values.

Until recently the connective tissue has not been introduced in models. This is now possible in a finite element model in which defined parts may have deviating volume conduction properties [25]. A connective tissue layer close to the recording electrode may change the peak-to-peak voltage of the single fibre action potential up to about 30 % [25].

The effects of different representations of the source of the electrical activity on the peak-to-peak amplitudes of single fibre action potentials is shown in figure 5.

The simulation using the adapted measured transmembrane current gives the best result. Figure 5 shows the large effect of small changes in the representation of the source. It has to be recommended to pay more attention to the parameters of intracellular action potentials than seen in several papers e.g. [15].

An unexpected experimental result is that the single fibre action potential can be above the noise level (about 20μV) up to recording distances of up to 2mm [25], clearly in contrast with the simulation results shown in figure 5.

In needle recordings (animal experiments [25], human muscles [29]) it has also been observed that the potential stays high at large distances. This cannot have been caused by the background activity of other motor units in [25]. These observations may have great impact for motor unit action potentials. It means that the contribution from fibres far away from the electrode is relatively large to the MUAP.

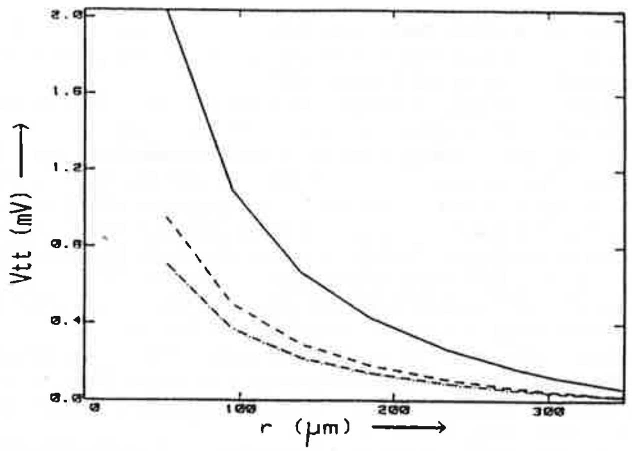

Figure 5. Peak-to-peak amplitude of simulated single fibre action potentials of rat muscle (37°C) as a function of recording distance, for three approximations of the source:

————	**source based on measured membrane current, adapted to in vivo temperature**
– – – – –	**analytical Rosenfalck-representation [28] of membrane current belonging to an in vivo measured intracellular action potential**
–·–·–·–	**second derivative of a representative in vivo measured intracellular action potential.**

5. From single fibre to MUAP modeling and quantitative analysis of MUAPs

In motor unit recordings the recording distance is large in general. Probably, it is not necessary to introduce in MUAP models the capacitive component of the volume conduction [29], but we have to take into account the anatomy of the muscle and the mean source representation [30]. Simulations with the membrane current source, as shown for single fibre calculations in fig. 5, have not been made for the motor unit condition. MUAP simulations were bad in early rising and late declining phases of the action potentials [16]. Smoother phases of the membrane current compared to the second derivative of the intracellular action potential will contribute to a better fit.

At first in the evaluation of the meaning of quantitative analysis of MUAPs a rather negative opinion arises. The origin of the MUAP is very variable: each motor unit has its own structure, its fibres are embedded in a unique way in the muscle, muscular structures influencing current distribution are irregular and the contribution per fibre cannot easily be tested. The fact that the current injection by the fibres has a dominant role in MUAPs independently whether they are measured with small or large electrodes, inside or outside the motor unit territory attracts the attention especially to the fundamental study of membrane currents.

Perhaps quantitative analysis is worthwhile in the future if it is clear that

- the MUAP presents other information about the motor unit than other measurements, e.g. the source and the volume conductor effects can be discerned
- the MUAP measurement is easier than other measurements at motor unit level
- the MUAP recordings can be made without pain for the person or patient and within a short time.

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THE SOURCE OF ELECTRICAL ACTIVITY IN SKELETAL MUSCLE

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Introduction. In simulations of single fibre action potentials (SFAPs), motor unit action potentials (MUAPs) or interferenced electromyograms the source of the signals is often represented as the second derivative of the intracellular action potential (IAP). Nearly always the IAP is approximated with a simple analytic description. Its second derivative is triphasic and the first phase has the largest voltage. In contrast, the second phase is the largest in measured triphasic SFAPs. This presentation concerns new experimental data of the source: the current through the muscle fibre membrane and its use in SFAP-modeling.

Methods. Mouse hindlimb muscles were excised and bathed in Ringer-solution at room temperature. Both the current and the potential were measured with the so-called loose patch method (Almers et al., 1983) under potential and current controlled clamp conditions respectively. The recordings were made with pipettes positioned on superficially located muscle fibres (Wolters et al., 1991). The IAP was measured with a flexible micropipette. SFAP-simulations were done with the radially bounded muscle model of van Veen (1992). As source activities the second derivative of IAPs and the membrane currents were used.

The contribution of sodium and potassium current in the total membrane current was detected using the channel blockers TTX and TEA.

Results. The time course of the potential and current recording was identical when measured at the same membrane spot for two subsequent propagated action potentials. The resemblance between the second derivative of the IAP and the current and voltage recordings was apparent. Therefore, in our experimental condition the representation of the membrane current by means of the second derivative of the IAP is acceptable. The membrane current showed three phases, the middle one was the largest as in triphasic SFAPs.

On the basis of measured data the membrane current at body temperature was estimated. Application of this source resulted in an enhanced simulation of SFAPs in comparison with other inputs for the source.

With channel blockers the membrane current was splitted in three components. When sodium and potassium channels were blocked, the membrane current equalled the first derivative of the IAP. This component of the membrane current was identified as a capacitive current. Other components in the normal triphasic membrane current were an inward sodium current and an outward potassium current.

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SOUNDMYOGRAM AND ELECTROMYOGRAM COMPARED ANALYSIS TO STUDY MOTOR UNIT ACTIVATION PATTERN

C. Orizio, F. Esposito, A. Veicsteinas.

Aim of this work was to compare the influence of the motor units activation pattern on the properties of the surface electromyogram (EMG) and of the soundmyogram (SMG) during isometric contractions of the elbow flexors. At this purpose the time and frequency domain parameters of the two signals were investigated. In seven healthy subjects - age 34 ± 5 yr (x±SD), maximal voluntary contraction (MVC) 282 ± 59 N - the EMG and the SMG were picked up by two silver bars electrodes, 1 cm spaced, and a piezoelectric contact sensor, HP 21050 A, placed over the belly of the biceps brachii. After antialiasing filtering, 2-400 Hz and 2-100 Hz for EMG and SMG respectively, and A/D conversion the two signals were sampled at 1024 Hz and stored on-line on the personal computer hard disk. Each subject performed 4 s steps (constant force) at 20, 40, 60, 80 and 100% MVC, in randomized sequence. Between one effort and the other 3 minutes of recovery were allowed. The arm of the subject was placed in an anatomically shaped stirrup allowing 115° between arm and forearm. The exerted force was recorded by a load cell strapped perpendicularly to the subject wrist. The same computer gave on its monitor the force target, and the visual feedback of the output force. The EMG and SMG time and the frequency domain analyses were carried out by means of the root mean square (RMS) calculation and of the maximal entropy spectral estimation (MESE). The processing were carried out on the 2048 middle data window.
Time domain results: EMG and SMG RMS were 101 ± 50 μV and 1.2 ± .4 mV (20% MVC), 239 ± 110 μV and 2.5 ± .8 mV (40% MVC), 291 ± 110 μV and 3.48 ± .8 mV (60% MVC), 332 ± 140 μV and 2.56 ± 1 mV (80% MVC), 480 ± 180 μV and 2.0 ± 1.4 mV (100% MVC). Frequency domain results: EMG and SMG mean frequencies (MF) were: 79 ± 18 and 8.4 ± 2 Hz (20% MVC), 90 ± 24 and 11.2 ± 1.9 Hz (40% MVC), 93 ± 23 and 13.9 ± 1.9 Hz (60% MVC), 92 ± 20 and 16.7 ± 2.7 Hz (80% MVC), 80 ± 18 and 21.2 ± 6 Hz (100% MVC).

The greater the muscle output force the greater the number and the firing rate (FR) of the active MUs. In particular in biceps brachii, the MU recruitment (REC) is used to control force up to about 70 % MVC. To generate force beyond this level of effort an increase in the FR of the recruited MUs is necessary. The muscle sound detected at the muscle surface can be considered as a compound signal related to the pressure waves generated by the dimensional changes of the muscle fibres of the active MUs. This can explain the reduction of the SMG RMS beyond 70% MVC when a fusionlike situation, due to the high MUs' FR, takes place. This phenomenon does not affect the EMG RMS that still increases at high contraction level. From low to high levels of effort the recruitment process involves progressively larger MUs with higher conduction velocity (CV). This can explain the EMG MF increase up to 60% MVC reported in our study. In 70-100 % MVC range the motor unit FR increment is the sole tool to increase force and no changes in CV takes place, as a consequence the EMG MF v. force relationship levels off. At the same time the SMG MF v. force continues to increase. On the contrary to EMG the SMG time and frequency domain characteristics are greatly affected by the MUs dominant FR. In fact the higher values of the latter at high contraction intensities are monitored by a clear SMG RMS reduction and a by a steeper SMG MF v. force relationship. Conclusion: In biceps brachii the different way in which REC of larger MUs and FR changes affect the EMG and SMG time and frequency domain properties, allows to get informations about the MUs activation pattern by the comparison of the two signals RMS and MF v. force behaviors.

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SYNCHRONIZATION OF MOTOR UNIT FIRINGS IN SEVERAL HUMAN MUSCLES

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Synchronization of motor unit (MU) firings was investigated in six human muscles. EMG signals were recorded during isometric contractions at approximately 30% of maximal voluntary contraction level. Firing times of individual MUs were identified using the specialized myoelectric signal acquisition and decomposition technique referred to as Precision Decomposition. Cross-interval histograms were constructed for pairs of MU action potential trains by measuring the first order forward and backward recurrence times (the nearest forward and backward firing times) of the alternate MU with respect to each firing of the reference MU. A rigorous statistical technique was used to assess the presence of peaks in the cross-interval histogram. The location of the center of the peak was measured. A synch index was developed to quantify the percentage of firings that were synchronized. The percentage of pairs of concurrently active MUs that contained synchronized firings was measured. Synchronization was defined as the tendency for two MUs to fire with dependent latencies relative to each other more often than would be expected if the firings of the two MUs were independent random processes. The following conclusions were drawn:

1. Synchronization among MU firings may not have a significant physiological purpose, but instead it may be a by-product of other physiological mechanisms.

2. Our results argue against the hypotheses of common presynaptic fiber branches and feedback mechanisms that have previously been used to explain the occurrence of synchronization. A new hypothesis was introduced whereby oscillators within the central nervous system drive motoneurons to fire in synchrony. This hypothesis requires no common physical connection.

3. Synchronization was observed to occur in two modalities. The short-term modality was seen as a peak in the cross-interval histogram centered about zero delay (0.5 ± 2.9 ms) and with an average width of 4.5 ± 2.5 ms. The long-term modality was seen as a peak centered at latencies ranging from 8 to 76 ms. Long-term synchronization occurred far less frequently than short-term synchronization.

4. An average of 4.0% of the firings were synchronized in the six muscles that were tested. The synch index was statistically indistinguishable ($p=0.07$ to 0.89) among subjects.

5. The percentage of pairs of concurrently active MUs that showed short-term synchronization was 60% across all the muscles. This value was significantly lower in larger muscles than in smaller muscles, and in two out of 11 subjects.

6. The synch index was found not to be dependent on the recruitment threshold of the reference MU or the difference between the recruitment thresholds of any two MUs that displayed synchronization.

EMG OF FINE MOTOR CONTROL OF OPPONENS POLLICIS AND TIMING OF WRITING AT DIFFERENT AGES

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To better evaluate motor difficulty, this work attempts to determine and compare the quality of fine motor control (FMC) of Opponens pollicis (OP) and the timing of performance which could be affected by the degree of difficulty brought by writing mediums (WM) and by the age of subjects. From a previous anatomic study of the functional constitution of OP, two neuromuscular "divisions" (bundles of muscular fascicles individually insert and innervated separately (1)) are chosen for an EMG biofeedback testing of FMC (2). The introduction of electrodes in the OP is made from bony landmarks in the deep central, a fan shape division of this muscle which may act in several directions of the thumb's opposition, and in the trapezo-metacarpal, a rectangular division made of transverse muscle fibers which may aid the flexion, opposition and stabilizer function of the thumb. Two groups of active subjects participate. They are Young (Y) of 20 to 40 years (n=5) and Senior (S) of 60 to 86 years old. They are comfortably fixed in an orthotic system to eliminate effects of gravity at best and to standardize positions and motions of the thumb - the S1, S3 and post S4 : at an upward writing and the S2 : at a downward writing postures. "Slow" writing downward (D) and upward (R) of 5° carpo-metacarpal motions are also required. Subjects are asked to single out a motor unit into one division of OP and to try to hold this chosen FMC action during the entire testing. In both groups, results show three levels of FMC : 100% (a pure single motor unit repetitive action), 80% (a transfer of motor unit action or a complete inhibition), and 60% (an evocation of a light activity). The means of the levels of FMC and the frequency distribution for each level demonstrate that each WM used, requiring a different type of contraction of OP, have its own degree of success which is similar for both groups (Table). The age of active subjects do not affects the mean of FMC success of both groups (85%/85%). The younger group shows a tendency to accomplish the voluntary writing faster (24%) than the older group (Table). The effort of FMC seems to affect the timing of production but not the levels of FMC success in the older group, a matter of basic interest in the evaluation of motor inability. Granted from CAFIR, U. de Montréal.

%	$\bar{X}$ of FMC		f distribution of FMC levels :			sec	$\bar{X}$ of timing	
	Groups :						Groups :	
WM	Y	S	100%	80%	60%	WM	Y	S
S1	92	98	74	25	0			
D	82	83	16	72	11	D	10	18
S2	86	89	32	68	0	S2	15	20
R	74	72	5	46	48	R	13	12
S3	82	83	32	41	26			
S4	91	95	42	46	11			
$\bar{X}$	85	85				Total	38/50 : 24%	

Table - Influence of writing mediums (WM) and groups of different age on the mean of FMC, on the frequency distribution of levels of FMC and on the total mean of timing WM to obtain the optimum FMC result.

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ESTIMATION OF THE NUMBER OF MOTOR UNITS IN A MUSCLE BY MEANS OF SURFACE EMG RECORDINGS

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Abstract

The purpose of this study was to investigate a new way to estimate the total number of motor units in a muscle, using the variance of the EMG signal. This parameter can be estimated during voluntary contraction (Vv) and during electrical stimulation of the muscle (Vs).

The variance of an EMG signal depends on the number of motor units, the variance of the MUAP's and their firing frequencies. In case of unsynchronized motor unit activity, Vv will increase with the square root of the number of motor units whereas with electrical stimulation Vs will increase linearly with the number of motor units. As a consequence, assuming equal MUAP's and equal repetition rates, it can be shown analytically that the ratio between the Vs and Vv equals the number of motor units.

Using a simulation model, that was developed before (Hermens et al. 1992) we investigated the possibilities of this relation to be used to estimate the total number of motor units.

The following results were obtained:

- the simple relation can be disturbed considerably when synchronisation occurs during the recording of the EMG signal during voluntary contractions. Simulation shows that synchronisation results in a strong increase of the variance of the EMG signal. The role of synchronisation in voluntary contraction is still largely unknown. Yet an indication of the presence of synchronization can be obtained from changes in the power spectrum of the EMG signal. Synchronisation will result in a decrease of the median frequency.
- variability between the MUAP's considering their amplitude or duration, described in terms of density functions, does not influence the relation.

Considering the experiments the following is required:

- the mean firing rate of the motor units has to be estimated from the EMG signal obtained during voluntary contractions. With most muscles this is possible by detection of the first peak in the power spectrum.
- the muscle has to be stimulated supramaximally, without disturbing the EMG registration.

Presently we are carrying out the first experiments. The results will be presented at the congress.

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THE BEHAVIOR OF M.E.P. AT DIFFERENT DEGREES OF MAXIMAL VOLUNTARY ACTIVATION

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It is well known that a moderate muscular activation (10-20%of the maximal voluntary activation or MVA) causes an amplitude increase and a latency reduction of the motor evoked potentials (MEP) by transcranial magnetic stimulation (TMS). So far, however, the behavior of these parameters for different degrees of muscular activations,particularly for submaximal or maximal contractions, has not been investigated. We performed TMS in 14 healthy right-handed subjects (10 males; 4 females) whose mean age 31 years (range 21-38 years), recording from the right abductor digiti minimi (ADM) and biceps brachii (BB), during isometric contractions of increasing intensity until they reached the MVA. In each subject we recorded the surface integrated EMG before every evoked muscular response and, in order to confirm its relationship with the muscular strength we obtained in 4 subjects the surface integrated EMG at 20%,40 %, 60 % ,80 % and 100%of MVA, measured with a dinamometer and controlled by means of a visual feed-back.
For the BB, MEP amplitude and area mean values were higher when the muscular activation ranged between 61% and 80%of MVA than for submaximal or maximal (81-100 %) contractions. For the ADM the increment of the area was proportional to the intensity of the muscular activation, while the amplitude did not change significantly for degrees of contraction over the 60%of MVA.
Our data suggest that in healthy subjects, in the BB, an increasing muscular activation is not paralleled by a progressive facilitation of MEPs. This phenomenon could depend by an increment of the number of spinal motoneurons in refractory period, therefore not excitable by TMS-induced volley, caused by increasing muscular activation. In the ADM this trend is moderate and involves only and MEP amplitude. The different behavior betyween proximal (BB) and distal (ADM) muscles could depend by the different ratio of the two types of motor units which are present in the two groups of muscles.

EFFECTS OF SKIN DESENSITIZATION ON MOTOR UNIT BEHAVIOR IN MAN

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The effects of reduced input from cutaneous afferents upon motor unit (MU) recruitment threshold and firing rate during isometric voluntary contractions in man has been the object of this study. A specially designed quadripolar needle electrode was used to record three independent channels of myoelectric signals from the right First Dorsal Interosseous muscle. Linearly varying isometric contractions were performed by the subject at regular intervals of time (15 min) for one hour. Topical anesthesia was achieved by means of an anesthetic (lidocaine 10%) directly sprayed on the skin of the right hand and forearm on either volar and dorsal aspects. The extent and duration of skin anesthesia was evaluated using Von Frey's method. Control and anesthetic experiments have been performed in different days on 6 subjects and the activity of 18 and 23 MUs in the two conditions respectively has been analyzed. Individual MU action potential trains have been assessed using an interactive decomposition program (Mambrito B. and De Luca C.J., Electroencefalogr. Clin. Neurophysiol., 58, 1984). A significant ($p<.05$) increase of the excitability of the low threshold (below 20% MVC) MUs resulted from the application of the anesthetic, while the high threshold (above 30% MVC) MUs were inhibited. Mixed results have been obtained on the intermediate threshold MUs. The mean firing rate showed an inverse relationship with the recruitment threshold; when recruitment increased mean firing rate decreased and viceversa. Electrical stimulation of cutaneous receptors has been shown (Masakado et al., Exp. Brain Res., 86, 1991) to alter MU behavior in a completely opposite fashion. On the basis of these results, two major conclusions can be drawn. 1) There is a strict relationship between MU recruitment threshold and its stationary firing rate which, to our knowledge, has not been previously considered and 2) Different motoneurons in the same pool receive from skin afferents different proportions of excitatory and inhibitory inputs.

DIAGNOSIS OF NEUROMUSCULAR DISORDERS BY NONINVASIVE ELECTROMYOGRAPHY OF SINGLE MOTOR UNITS

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Introduction

The diagnosis of neuromuscular disorders is essentially based on the information about the single motor unit (MU) activity. In clinical practice the needle-EMG technique is used because the conventional surface-EMG does not contain information about the activity of single MUs. But this method is invasive and painful. We have developed a new noninvasive EMG technique allowing the recording of single MU activity [1].

The noninvasive EMG-procedure

The new EMG-procedure depends on the use of a multi-electrode array in combination with a spatial filter processing [1]. This spatial filter amplifies the signals of MU located beneath the center electrode of the filter and reduces the signals of more distantly located sources. Thus, from one spatial filtered channel it is possible to characterize the activity of one MU. A series connection of such spatial filters allows the determination of the excitation spread along single muscle fibers. In this way, it is possible to nonivasively measure the conduction velocity in single MUs, to determine the firing rate and to localize the site of the endplate region [1,2].

Clinical validation

The noninvasive electromyography of single MUs was examined in volunteers and patients suffering from different kinds of neuromuscular disorders. From the investigations in healthy children it was possible to derive the conduction velocity in single motor units depending on the age of the children. Additionally, a significantly reduced conduction velocity in patients with Duchenne muscle dystrophy was found.

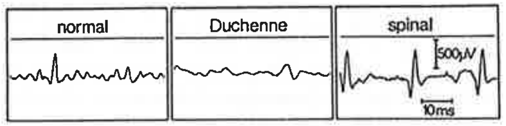

Fig. 1: Spatial filtered EMG-signal of children with different kinds of neuromuscular disorders.

The EMG-patterns (Fig.1) recorded with the new noninvasive EMG-procedure and their typical parameters (for example, entropy or variance of the amplitude a.s.f) of healthy children, patients with muscular disorders or neuronal disorders, respectively, were compared. The results show that there is a distinct difference in the EMG-shapes and their typical parameters between the different groups of patients (Fig.1).

Conclusions

Using the spatial filtering EMG-procedure it is possible to differentiate noninvasively between patients and volunteers on the one hand and between patients with muscular and patients with neuronal disorders on the other hand. Thus, the new EMG-procedure is suitable as a **noninvasive** tool for the diagnosis of neuromuscular disorders.

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EMG POWER SPECTRA FREQUENCIES PROFILES OF ELBOW ANTAGONIST MUSCLE PAIR DURING LINEARLY AND STEPWISE INCREASING CONTRACTIONS

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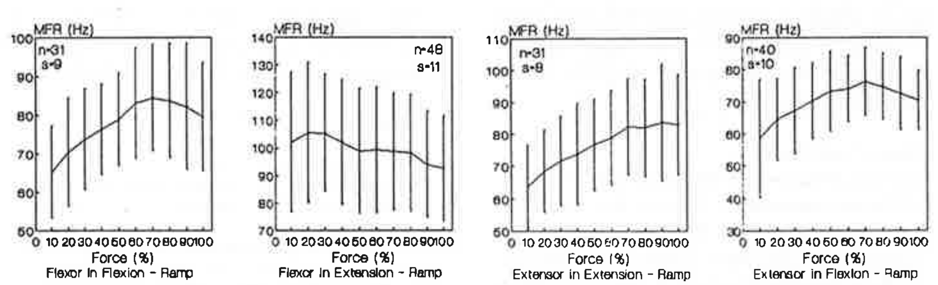
A previous report has suggested that changes in the myoelectric signal's median frequency can indicate the motor units recruitment control strategy employed by the muscle (Solomonow et al., J. Appl. Physiol., 1990). This technique has been used to study the elbow extensors during linearly (ramp) and stepwise (step) increasing contractions (Bilodeau, Eur. J. Appl. Physiol., 1991). It is of interest to compare the control strategies employed by antagonist muscle pairs in these types of contractions.

A bidirectional load cell was used to measure elbow flexion and extension force. Differential amplifiers were used to obtain the myoelectric signal from the biceps brachii and the triceps brachii. The subjects performed ramp contractions which increased from 0% TO 100% Maximum Voluntary Contraction in three seconds. The step contractions were separated into eleven levels of 0, 10%, 20%... to 100% MVC, performed in random order. The median frequencies obtained through an FFT Algorithm for each muscle at each force level were plotted for each type of contraction. Some examples are shown in **Figure 1**.

It was concluded that the control strategy employed by muscles is dependent upon the type of contraction they are performing. When acting as agonists during ramp contractions the MF of both biceps and triceps increased up to 70% MVC. During step contractions, the MF increases only up to 50% MVC, and then levels off or slightly decreases. In both step and ramp extensions, the flexor acting as antagonist exhibited a decreasing MF pattern. In contrast, during flexion the antagonist triceps showed generally increasing MF pattern in step contractions, and an increasing-decreasing pattern during ramp contractions.

Among the possible explanations for the MF patterns observed is the differences in neural input and thus control strategy for ramp (dynamic) versus step contraction (static), which may account for differences in the agonist MF patterns. The antagonist control strategies are probably linked to compensation of various external and internal disturbances such as gravity vector, muscle moment arm, joint angle, etc..., as well as the conflicting needs of maximizing net joint torque while maintaining joint stability.



PROPERTIES OF SURFACE MOTOR UNIT ACTION POTENTIALS

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1: Introduction

Allthough there is a considerable experimental knowledge of MUAPs recorded with needle electrodes, there is hardly any knowledge on the properties of MUAPs as can be recorded with surface electrodes.
Yet simulation clearly shows that especially the surface EMG power spectrum strongly depends on MUAP properties like shape and duration (Hermens et al. 1992). This paper discusses these statistics and their implications evolving from empirical data recorded through a technique of triggered averaging.

2: Methods

MUAP's were isolated by using the signal of a single fibre electrode to trigger the averaging process of the surface EMG signal (averaging window: 100 ms; 1024 samples).

Bipolar recordings with two interelectrode distances were realized by an electrode configuration with three surface electrodes, aligned in the direction of the muscle fibres, halfway the motor endplate region and the distal tendon (inter-electrode distances: 20 mm). For monopolar recordings the middle electrode was used separately, with a reference electrode placed on the elbow.

The EMG signals were amplified, bandfiltered (surface signal: 10-1000 Hz; single fibre signal: 0.2-20 kHz) using a two channel Mystro system (Medelec). The experiments were carried out at the m. biceps brachii of 7 male healthy subjects, 20-35 years of age. Subjects were asked to exert a low level constant force (range 10 - 40 % of maximal force). The single fibre electrode was manipulated randomly until a single fibre signal (amplitude > 0.2 mV, rise time < 0.3 ms) was found. The negative slope of this signal was used as a trigger. Each averaging process consisted of at least 250 subsequent responses and was carried out twice. The MUAP was accepted only if the outcome of both averaging procedures was approximately identical. The accepted MUAP's were transferred from the MYSTRO system to an IBM micro computer for further data processing.

3: Results

Generally MUAP's recorded with a monopolar electrode configuration (51 recordings) could be characterized as tri-phasic shapes with the second phase having the largest peak voltage (65%, figure 1a). In most cases the peak voltage of the third phase was larger than that of the first phase. With approximately 25% of the monopolar MUAP's, the peak voltage of the first phase was so small, compared to the peak voltage of the other phases, that a characterization of the monopolar MUAP as biphasic was more appropriate (figure 1b). With approximately 50% of the biphasic and triphasic monopolar MUAP's relatively sharp peaks were observed in the last phase of the MUAP (figure 1c). The remaining monopolar MUAP's (10%) were characterized as monophasic.

The MUAP's derived with a bipolar electrode configuration were characterized as slightly asymmetrical biphasic shapes (figure 1d,e). Only with some MUAP's (< 10%) additional phases were seen (figure 1f). No differences in MUAP shape could be observed, with different interelectrode distances.

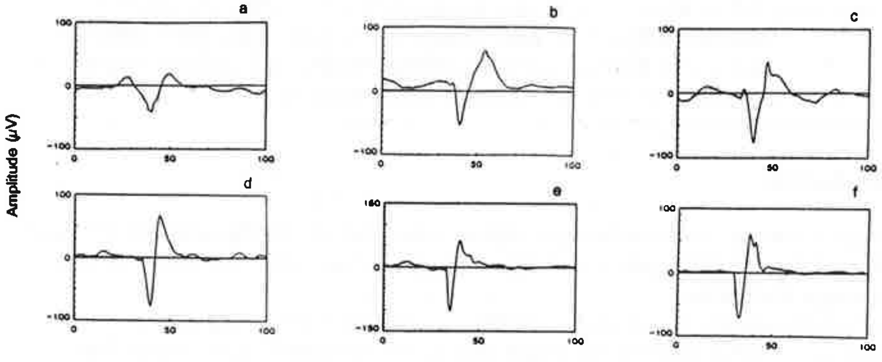

Figure 1. Examples of MUAP's obtained by needle triggered averaging of the interference pattern, using monopolar (a,b,c) and bipolar (d,e,f) surface electrode configurations. The points show the result of a repetition of the averaging process.

In order to study these biphasic MUAP's in a more quantitative way, the peak-peak voltage (V_{pp}) and the peak-peak time (T_{pp}) were determined for both inter electrode distances (20 mm el.: 51 MUAP's; 40 mm el.: 62 MUAP's).

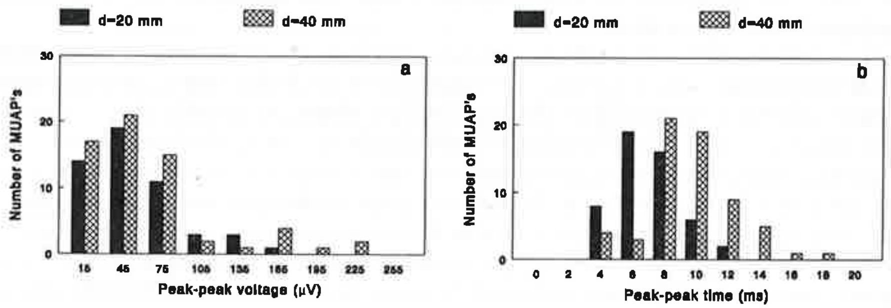

Figure 2. Histograms of the peak-peak voltage (a) and the peak-peak time (b) of MUAP's, recorded with a bipolar electrode configuration. Interelectrode distance d=20 mm and d=40 mm.

Both histograms of V_{pp} are skewed. They overlap largely (mean: d=20 mm: 54 μV; d=40 mm: 64 μV. St. dev.: d= 20 mm: 37 μV; d=40 mm: 50 μV). No significant difference between the two population means was found (t-test, 0.05 level of significance).

The histograms of T_{pp} are more symmetrical and show less overlap (mean: d=20 mm: 6.9 ms; d=40 mm: 9.7 ms. St. dev.: d=20 mm: 2.0 ms; d=40 mm: 2.8 ms). A significant difference between the two population means could be detected (t-test, 0.01 level of significance).

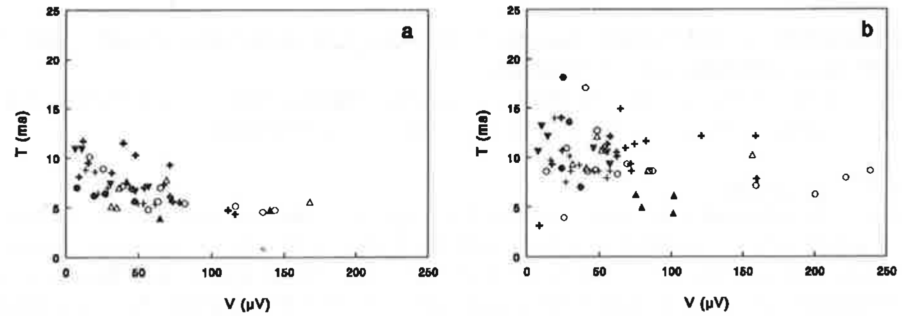

Figure 3. Peak-peak time (T) versus peak-peak voltage (V) of MUAP's recorded with a bipolar electrode configuration for two interelectrode distances (d=20 mm: a; d=40mm: b). The different marks refer to different subjects. The drawn line shows the relation as predicted by the model.

With the 20 mm interelectrode distance data, a significant correlation was found (t- test, linear curve fitting, p<= 0.01)) between V_{pp} and T_{pp} of the pooled data and with 4 of the 7 subject data. With the 40 mm interelectrode distance, the relation between V and T is much less clear. This was also confirmed by the statistical tests, which indicated for both the pooled data as well as for each subject an absence of significant correlation (p> 0.1).

4: Discussion

Our data, on 51 shapes, support the characterization of the monopolar MUAP shapes as triphasic, which is in accordance to theoretical and experimental findings (e.g. Griep et al 1982, Gootzen et al. 1991).

A peak in the last phase as observed in about 50% of the cases is probalbly caused by the arriving of the action potentials at the tendon (Gootzen et al. 1991). Our results show that the MUAP, recorded with a bipolar electrode configuration generally is biphasic without additional peaks. This finding reflects that the middle phase of the monopolar MUAP is the most dominant in characterizing the bipolar MUAP.

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SYMMETRY OF GLUTEUS MAXIMUS ACTIVATION IN BELOW-KNEE AMPUTEE GAIT IN COMPARISON TO NORMAL.

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INTRODUCTION

Often rehabilitation clinicians assume that improving the symmetry of an amputee's gait would inherently result in a more optimal gait pattern and that it would reduce the energy cost of ambulation (Hannah and Morrison, 1984). Winter and Sienko (1988) indicated that there is no scientific justification for this assumption. As a contribution to this debate the symmetry of muscle activation of the gluteus maximus was studied empirically both in normals and experienced below-knee amputees.

METHODS

Computer Averaged Surface Electromyographic Profiles (CAEPs) were determined of the Gluteus Maximus muscle of both legs in two groups of subjects at comfortable walking cadence using the instrumentation and methods described by Kleissen et al. (1989). One group consisted of 9 male below-knee amputees, age range 28 to 70 years (mean age 52 years) and mean mass 84.6 kg. These subjects had been using their prosthesis for longer than 1 year without complaints or problems, and had a normal walking function in daily life. Amputation had been performed after trauma, without affecting quadriceps and hamstrings. All subjects were tested using their own prosthetic device and shoewear. The normal group comprised 10 male subjects, age range 28 to 71 years (mean age 43 years) and mean mass 79.6 kg. Also the normals used their regular shoewear. Surface EMG data of at least 15 cycles of gait collected in the CAEPs.

RESULTS

Figure 1 presents the CAEP of the Gluteus Maximus of the right limb for the normal group. Figures 2 and 3 give the CAEP for the amputee group for the prosthetic and healthy limb respectively.
Statistical analysis of the data yielded, among others, the following observations:
- The symmetry of the profiles for left and right limb, expressed in the Pearson Correlation Coefficient, proved to be significantly lower for the amputee group (Wilcoxon test, $p < 0.05$).
- The total EMG activity over the gait cycle was significantly higher in the prosthetic leg than in the healthy leg. (Signed Rank test, $p < 0.05$).
- The EMG activity in the prosthetic leg during early stance was nearly twice the activity in the healthy leg.

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Fig 1: CAEP of Gluteus Maximus
Right Limb, Normal (n=10)

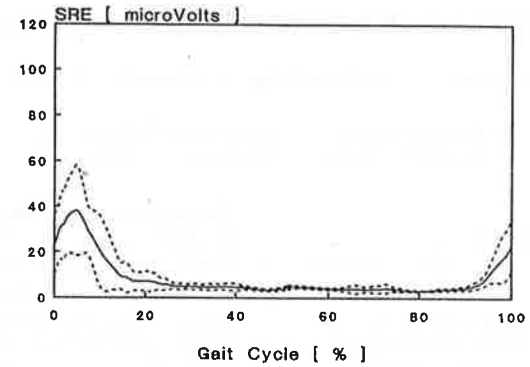


Fig 2: CAEP of Gluteus Maximus
Prosthetic Limb, Amputee Group (n=6)

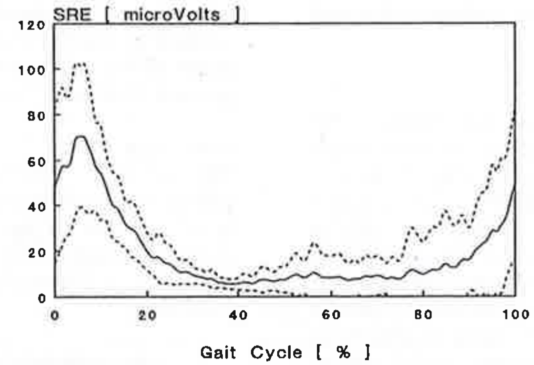
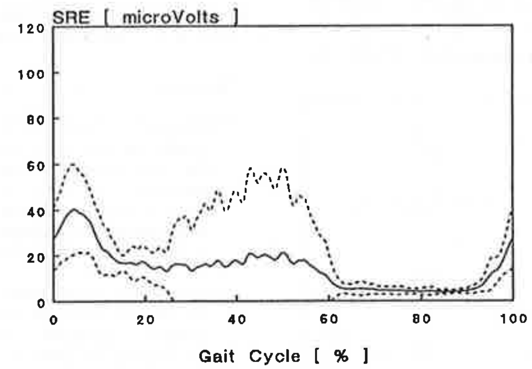


Fig 3: CAEP of Gluteus Maximus
Healthy Limb, Amputee Group (n=6)



RECORDING OF SINGLE MOTOR UNITS BY NONINVASIVE ELECTROMYOGRAPHY WITH HIGH SPATIAL RESOLUTION

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Introduction

Due to the low spatial resolution of the unipolar set-up, the conventional surface EMG technique is useless in the measurement of single motor unit activity which is of interest in the diagnosis of neuromuscular disorders. Consequently, the needle-EMG is used for this purpose, an invasive and painful method. We have developed a noninvasive EMG procedure with a high spatial resolution which allows the consideration of the activity of a single motor unit (MU).

Spatial filtering

A bipolar lead with an interelectrode distance in the mm-range is the most simple configuration of a spatial filter [1]. However, the spatial filtering of the bipolar lead is not sufficient to discriminate the activity of single MUs (Fig.1, B). For the optimization of the spatial filter procedure five unipolar leads are used which are arranged crosswise (Fig.1, D). This filter, called Normal-Double-Differentiating Filter (NDD-Filter), amplifies the electromyographic signals of sources directly under the center electrode and reduces those of more distantly located sources (Fig.1, D). In this way the activity of only a few MUs located on the surface of the muscle and beneath the center of the filter becomes clearly distinguishable in the signal course.

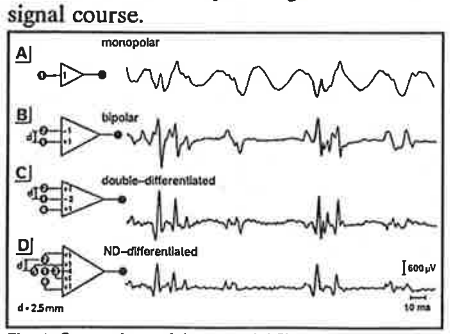

Fig. 1: Comparison of three spatial filters

The multi-electrode array

A series connection of those NDD-filters which are arranged in a consecutive row allows the observation of the excitation spread along the muscle fibers of single motor units [1,2]. For this application a multi-electrode array is used consisting of 32 gold-covered pin electrodes. The electrodes have an equal interelectrode distance which is adjusted to the size and depth location of the muscle [1,2]. For investigations of the biceps muscle an interelectrode distance of about 5 mm and for measurements of the m. abductor pollicis brevis a distance of about 2.5 mm is used.

Conclusions

The use of the multi-electrode array in combination with the NDD-filter processing is well suitable for the noninvasive examination of single motor units. In the proposed way, it is possible to determine noninvasively:

- ***the conduction velocity in single MUs***
- ***the firing rate of specific MUs***
- ***the muscle fibre orientation***
- ***the site of the endplate region***

The method is practicable even at high levels of voluntary contraction of superficial muscles.

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INVESTIGATION OF HUMAN MOTOR DISTURBANCES USING COMBINED KINEMATIC AND EMG ANALYSIS

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The development of computer based motion analysis systems has made it possible to study human movement, including motor dysfunction occurring as a result of neurological disease, in ways that were not previously possible. In human subjects where physiological studies must obviously be retricted to using non-invasive techniques, recording motor output - kinematics, EMG signals, and forces - can provide information about the underlying physiological processes which generate the movement. By combining these measurements of motor output from patients with neurological disorders with the very precise anatomical information concerning the location and extent of lesions in the brain or spinal cord now provided by modern imaging techniques, we can develop new insights concerning the function of different components of the nervous system. Motor function analysis also has potentially important applications in the field of rehabilitation where there is a pressing need for reliable, quantitative methods to measure functional recovery following damage to the nervous system and to evaluate the reponse to various types of therapeutic interventions.

This introductory review will look at several areas where kinematic analysis combined with recording of EMG or ground reaction forces has contributed to our understanding of motor pathophysiology. Specific examples will include analysis of dysmetria and decomposition of movement in patients with cerebellar dysfunction and analysis of abnormal patterns of locomotion in patients with Parkinson's disease or hemiplegia.

1. Kinematic analysis of cerebellar dysmetria

Patients with damage to the cerebellum show a number of characteristic abnormalities of voluntary movement which are often most evident during reaching or pointing tasks requiring precise coordination of movements of several different limb segments. These abnormalities which include errors in the direction, velocity, and amplitude of target directed limb movements were described in detail more than 50 years ago by the

British neurologist Gordon Holmes. He used the term "decomposition of movement" to describe the loss of the ability to integrate the different components of a complex movement into a smooth continuous sequence, and documented these abnormalities with a simple but effective form of kinematic analysis, having his subjects perform movements in a darkened room with a small flashing light fixed to the fingertip. The path of the finger was recorded by exposing a single sheet of photographic film.

Modern systems using two or more special video cameras and computer software to perform three - dimensional reconstruction of movement parameters have made it possible to begin to examine some of the underlying mechanisms responsible for the movement abnormalities associated with cerebellar damage. In a recent study(1) we examined the coordination between angular changes at the shoulder and elbow joints during a pointing movement. Starting from a position in which the upper arm was vertical and the elbow flexed to 90 degrees, subjects were required to reach out to a target located at arms length in front of them, a movement requiring simutaneous flexion of the shoulder and extension of the elbow. In normal subjects performing this type of task movements at these two joints are coordinated in a manner which results in the finger following an approximate straight line path to the target. Movement at the shoulder and elbow begins at almost the same time, and plots of elbow angle versus shoulder angle show a smooth quasi - linear relationship. In patients with cerebellar damage the finger follows a very irregular path which shows considerable variability from trial to trial. Analysis of angular data shows uncoupling of the tight relationship which normally exists between movements at the elbow and shoulder joint. Often movement at one joint starts well before the other and angle - angle plots show very irregular patterns. Some of these changes could be interpreted as representing a fundamental abnormality in the timing of different components of complex movements which could be responsible for the decomposition of movement described by Holmes.

2.Analysis of gait patterns in Parkinson's disease

Many neurolological and musculoskeletal disorders produce abnormalities of gait, and analysis of kinematic and EMG activity together with force platform data can help us understand the complexities of these abnormal gait patterns. In Parkinson's disease a common problem is difficulty with initiation of gait. In normal subjects there is a well defined pattern or sequence of events associated with the process of initiating forward walking(2). The initial event is a sudden inhibition of EMG activity in the soleus muscles which are normally active during quiet standing, followed by prominent bursts of activity in the tibialis anterior muscles of both legs. Force platform recordings reveal that the center of foot pressure is displaced initially backward and toward the swing leg - that is the leg which is going to take the first step - before transfer of weight on to the stance leg. It would seem that these events are important for initiating the anterior displacemnt of the center of gravity and the forward fall which is required to generate propulsion.

In Parkinson patients EMG recordings reveal abnormalities in the motor program for gait initiation (3). Often there is tonic activity in both the soleus and tibialis anterior muscles during quiet standing which persists when the subject attempts to start walking. Failure to turn off the tonic activity in the soleus is a characterisitic feature, and the phasic bursts in the tibialis anterior are either absent or poorly defined. As a result there is little or no backward displacement of the center of foot pressure. Kinematic recordings confirm that, even though Parkinson patients often start from a forward flexed posture, the forward displacement of the upper trunk prior to taking the first step is reduced in comparison to normal subjects.

Reduced arm swinging is another common feature of Parkinsonian gait which can be analysed with kinematic and EMG recordings. Cross correlation of joint angle signals from the shoulder and hip have shown that the normal phase relationships between arm movement and leg movement are disrupted in Parkinson patients (Crenna and Lee, unpublished observations). In addition, the phasic bursts of EMG activity in the posterior deltoid, which are a constant feature of normal locomotion, are often absent.

3. The problem of data reduction

One of the problems with kinematic and EMG analysis, particularly when it is applied to human locomotion, is that the equipment and computer systems which are now available are so efficient at sampling and storing analog signals that the investigator can very quickly become completely overwhelmed with data. Traditionally, gait analysis has focused on description of obvious and easily measured variables such as cadence, stride length, timing of EMG bursts in multiple muscles, and force platform dynamics. However, to gain a better understanding of pathological gait patterns and to quantitate changes which may be occurring over time during the evolution of disease processes, new methods are needed to reduce the number of variables and to compress large volumes of data into more functionally meaningful proportions.

Promising results have been obtained from preliminary studies using the technique of principal component analysis to look at gait patterns in hemiplegic subjects (C. Mah et al, unpublished data). It was shown that after starting with 8 channels of raw joint angle data, over 95% of the variability in the original data could be accounted for in the first three principal components. Furthermore, a reliable reconstruction of the original signals could be made from the principal components, indicating that this approach does not result in significant loss of information. Comparisons of principal components from hemiplegic and normal subjects showed consistent differences in configuration, particularly in the second component. Each principal component includes information contributed by all of the original signals. Therefore, it is possible that principal components could provide insights into gait synergies which are not immediately obvious from inspection of raw kinematic data.

4. Conclusions

These few examples are described to indicate that with presently available technology it is possible to supplement clinical observations of abnormal movement patterns in neurological patients with quantitative analysis which may lead to a better understanding of the pathophysiology of the underlying processes. A major challenge for the future will be to develop appropriate methods for interpretation and display of the data which our machines are now capable of providing.

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IMPAIRMENT OF ISOMETRIC FORCE CONTROL IN THE HAND IN PATIENTS WITH HEMIANESTHESIA FOLLOWING A CEREBRAL LESION

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Introduction

It is generally aknowledged that the sensory input plays a crucial role in several aspects of motor behaviour, although some purposive movement can be correctly performed in deafferented invividuals.

Clinical data seem to suggest that one of the major motor difficulties in patients deprived of somatosensory feedback is the inability to control the production and maintainance of isometric force. These alterations may cause disability in everyday-life activities.

Methods

We performed a quantitative evaluation of isometric force control in the hand in 4 patients with cerebral vascular damage (two patients had a right-sided lesion and two had a left-sided lesion).

They were affected by somatosensory disturbances at the contralateral upper limb, without any clinical and CT scan evidence of involvement of the motor pathways.

The ability to control isometric force with both the affected and unaffected hand was tested in a tracking task involving the maintainance of a constant given level (680g) of isometric force (Slow isometric task).

The task was performed under two conditions:

a) Visual feedback condition (patients were provided with visual information about their force production);

b) No visual feedback condition (no supplementary information about motor output was provided).

Two patients were also tested in a ballistic task involving the production of rapid changes in isometric force from "O" to a level corresponding to 10% of maximum force (Fast isometric task). Forces were measured with strain-gauged transducers interfaced to a microcomputer. A monitor displayed a force time-curve as visual feedback.

The constant error (a measure of the tendency to produce a response greater or smaller than required) was calculated to assess the patients' performance; the measurements were expressed in grams (Tab.1).

Results

Slow isometric task. The patients performance with the unaffected hand was not influenced by the presence/absence of visual feedback. On the contrary, three of them presented a decrease in accuracy (showing a tendency to produce force smaller than required) when the visual information was not provided.

Fast isometric task. The patients showed a tendency to overshoot the target in the no-visual feedback condition as compared to the visual feedback condition with both the affected and unaffected limb.

SLOW ISOMETRIC TASK

	UNAFFECTED LIMB		AFFECTED LIMB	
	VISUAL FEEDBACK	NO VISUAL FEEDBACK	VISUAL FEEDBACK	NO VISUAL FEEDBACK
CASE 1	34.4	14.4	40.0	-308.0
CASE 2	38.0	30.0	52.8	- 70.4
CASE 3	10.8	- 2.8	8.0	- 27.6
CASE 4	11.8	-34.8	- 9.8	17.7

FAST ISOMETRIC TASK

	UNAFFECTED LIMB		AFFECTED LIMB	
	VISUAL FEEDBACK	NO VISUAL FEEDBACK	VISUAL FEEDBACK	NO VISUAL FEEDBACK
CASE 3	4.9	17.2	9.2	27.1
CASE 4	4.6	22.0	- 1.7	19.4

TAB.1 - MEAN CONSTANT ERRORS (BLOCKS OF 5 TRIALS)

Conclusions

These results agree with previous findings concerning motor disturbances in deafferented individuals; the partial dissociation in the performance of the two tasks seems to support the hypothesis of two different systems involved in the production and control of fast and slow isometric contractions.

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EMG-MONITORING AND ANALYSIS FOR EVALUATION OF SPASTICITY IN MS-PATIENTS

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INTRODUCTION

In the MS clinic where this investigation is being carried out, much effort is made to improve the autonomy and the quality of life of MS patients. Most of the multiple sclerosis patients suffer from spasticity and related symptoms, often impairing their available strength and motor abilities. Spasticity is a complex syndrome characterised by exaggerated stretch reflexes and muscle hypertonia. Quantification of spasticity is very important for objective evaluation of antispastic drugs and therapies. Clinical evaluation of spasticity is mostly based on Ashworth scale. Although of great clinical importance, this characterisation has some disadvantages : during the examination the muscles are at rest and the accuracy is limited. Some investigators have used neurological tests or performed biomechanical analysis of limb movement but no satisfactory objective and precise measure has as yet been found. A difference between spasticity in passive and active situations has also been reported.

This study is aimed at the observation and the measurement of spasticity. EMG-measurements are made on spastic muscles of the lower limbs of MS patients quoted at different degrees of Ashworth scale. Two different techniques were used: integrated EMG with a portable recorder and real-time EMG analysis with FFT.

EXPERIMENTS

In a first series of experiments, we used an ambulatory EMG recorder. The measurements were first conducted by a recorder developed at our institute and later by a two-channel ME 3000 portable system for muscle testing (Mega Electronics Ltd, Finland). Integrated EMG activity (IEMG) is registered with over 24-hours registration capability. Different patient markers are available. After registration, the data are transferred to a computer for storage and analysis. We were interested in muscle activity of MS patients during night. Different registrations were made on the tibialis anterior and gastrocnemius of healthy persons and MS patients. In all measurements monitoring electrodes of about 1 square cm were used.

We observed a large spread in IEMG amplitudes, and MS patients generally showed small amplitudes compared to healthy persons. Spastic movements however often show large amplitudes making calibration problematic. MS patients in general show more peaks during night than healthy persons and this is attributed in most of the cases to spasmodic activity. In spite of this the night registrations are difficult to interpret because of artefacts, not following instructions, etc.

In another series of experiments we measured the raw EMG signal by means of a MEDELEC MS91 two channel monitor connected to a CODAS data acquisition system. The resolution is 12 bits and the maximal sampling rate amounts to 50kHz. An on-board processor has advanced signal-processing capabilities as smooth screen scrolling, waveform compression and amplification, Fourier transformation. The EMG-monitor was set to a frequency band from 2-10000Hz and an anti-aliasing filter was used with a -3dB-frequency of 500Hz. The measurements were carried out with 4000Hz sampling frequency. We always used variable end point FFT.

Our investigation was carried out on eleven MS patients situated on different degrees of Ashworth scale (ranging from 0 to 3) and on healthy persons for comparison. The subjects were asked to lift their foot a few times and EMG was recorded by surface electrodes from tibialis anterior and sometimes from gastrocnemius at the same time. Different lifting cycles of a patient showed practically identical traces. In spectra from healthy persons and MS-patients during a voluntary movement we observe an initial part of climbing with a broad maximum around 80 Hz, followed by a descending part with a slope of about 0.06 dB/Hz. In the case of a short spasmodic movement the maximum is situated at a lower frequency range and the slope is increased. In the case of clonus, a pulsed waveform is observed in the time domain and a wavy spectrum appears.

DISCUSSION

We cannot make a distinction between spectra of normal movements in MS patients and healthy people. However the amplitudes of the EMG from the patients are often smaller. Spectra of short spasms seem to have a steeper roll-off at high frequency , and also a maximum situated in the lower frequency range or sometimes no clearly visible maximum at all. In the case of clonus, particular tracings are obtained as well in the time domain as in the frequency domain.

CONCLUSIONS

IEMG registration can be used to study spasms in MS patients but the interpretation of the night registrations is difficult. Moreover calibration problems and artefacts often occur. Registration of complementary information on movements could solve part of the problems.

Full-bandwidth EMG registration with FFT is valuable for studying activated spasticity and provides a lot of information. Much work has to be done for extracting parameters from the recorded data. More elaborate studies on a great number of patients will be conducted in the future. An electromechanical apparatus will be built to impose limb movements under controlled conditions.

ANALYSIS OF GAIT PARAMETERS REVEALS SIMILIARITIES IN COMPENSATORY MECHANISMS IN PATIENTS AFTER STROKE AND PATIENTS WITH HUNTINGTON'S DISEASE

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Clinical observation reveals characteristic gait patterns for different motor disorders, e.g. Huntington patients and hemiparetic patients after stroke. Modern gait analysis systems allow to describe those abnormalities in great detail. However, it is often difficult to decide, whether the observed parameters are due to specific lesions or due to compenstory strategies employed by the patients. Nevertheless this distincition is important for clinicians and physiotherapistst monitoring progress during therapy.
The aim of this study was to show, whether data obtained by a simple method of gait analysis - recording of pressure patterns and step time parameters during walking - help to distinguish between different strategies employed by patients with different motor disorders.

We investigated two patient groups with clinically easily distinguishable changes of gait: 12 patients with Huntington's (HD) disease (mean age 48 years, mean duration of disease 4,6 years) and 25 patients with hemiparesis (HP) after stroke (mean age 45 years, mean duration of disease 8 months). 12 healthy subjects served as controls.
Gait parameters were registered using overshoes with 8 in-built pressure sensors (registration frequency 50 Hz). Patients were asked to walk for 20 seconds using their own comfortable speed, walking distance was measured. Data were recorded on a portable registration unit carried by the patient. After data conversion pressure patterns and changes of pressure patterns over time could be illustrated. Walking speed, step frequency and step symmetry were displayed and duration of stand and swing phase including single and double support phases were calculated.

Pressure patterns showed an increased variability of mean pressure lines on both sides in HD and HP patients.

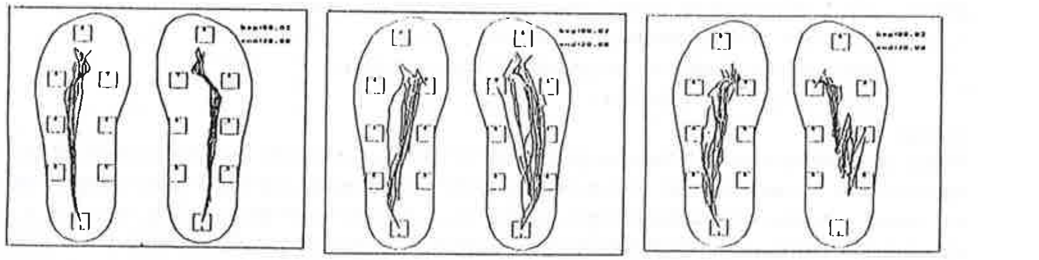

Fig 1: Pressure patterns of a healthy subjects (left side), of a patient with longstanding Huntington disease (middle) and with hemiparesis after stroke (right side).

Pressure patterns of HD patients and of stroke patients on the unaffected side displayed many similarities (fig. 1). This is presumably due to compensatory mechanisms för involuntary movements in HD patients and for impaired control of balance (e.g. decreased hip control) in HP patients. Analysing numerical data, HD and HP patients showed a reduction of walking speed. In addition, in HP patients duration of stand and swing phase were altered on both sides. Double support phase on the non-paretic side was mainly prolonged in stroke patients with predominant motor deficit.
Further analysis revealed, that gait patterns change according to severity of disease (HD and HP) and that in stroke patients those gait patterns are specifically affected by motor, sensory and tone disturbances. In stroke they are sensitive to changes over time.

The study shows, that with a simple gait analysis system only some specific parameters could be observed. Similar pressure patterns - presumably due to compensatory mechanisms - occur in clinically very different patient groups.
As the method is sensitive to changes of gait patterns over time especially in stroke patients, it opens up the opportunity to study those changes in more detail. This may help to evaluate the benefit of different methods and strategies used in physiotherapy.

QUANTIFICATION OF PATHOLOGIC GAIT IN PATIENTS WITH BI-AND UNILATERAL CEREBELLAR DYSFUNCTION

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Introduction

Surface electromyography (EMG) has been used to evaluate muscle activation patterns in patients with gait disorders. Some researchers have commented that there is no diagnostic function involved in an assessment of pathologic gait. We believe that simultaneous recordings from a set of muscles can be used to assess aspects of pathologic gait, e.g. changes with time, side-to-side differences, coordination abnormalities. In the present study we investigated muscle coordination in 4 patients with uni- and bilateral cerebellar dysfunction

Methods

The EMGs were recorded from the right and left m. biceps femoris, m. vastus lateralis, m. gastrocnemius medialis, and m. tibialis anterior during treadmill walking. Pressure-sensitive transducers monitored the pressure profile under the foot. The period of time between two heel contacts was considered 100% stride time. The EMG was full-wave rectified followed by a low-pass filtering and profiles were calculated. The amplitudes of the individual EMG profiles were normalized to the mean EMG amplitude and the mean amplitude was adjusted to 100% in order to minimize the inter-individual variances. During the experiment the gait velocity was chosen individually. EMG and video recordings were compared. The patients walked for approximately 30 seconds at each of the following conditions:

- Both hands were used by the patients to support walking.
- Free walking without support.

Results

When the patients changed from supported to free walking a number of unsynchronized steps occurred and in most cases m. gastrocnemius showed inadequate activity. For the patients with uni-lateral affections the predominant coordination changes were found in the affected side, although the normal side also showed major changes in coordination pattern compared to normals. This demonstrates that so-called "normal" sides do not show normal patterns as substantial compensation takes place during gait. In some cases the EMG showed altered activation profiles and coordination patterns from one side although the video recordings indicated almost normal and symmetrical walking. It is assumed that one way to improve pathologic gait is to investigate the coordination pattern.

Conclusion:

Different coordination patterns could be detected in patients with uni- and bilateral cerebellar dysfunction. This indicates that EMG can be used as a sensitive technique to quantify pathologic gait.

THE PENDULUM TEST IN THE EVALUATION OF HEMIPLEGIA

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Introduction: The pendulum test of the leg is gaining acceptance as a simple method for evaluating velocity-dependent spasticity of the quadriceps muscle. Numerical evaluation by this test is however only in its early stages and limited to the observation of the relaxation index (Bajd and L. Vodovnik 1984).
Moreover, methods shown by literature require some kind of mechanical instrumentation to be put on the knee joint such as electrogoniometers and velocity sensors.

Methods: A group of patients with ischemic hemiplegia (10 right side and 8 left side)were tested with the pendulum test. Time since vascular accident varied from one month to eleven years. The test was implemented with a gait analysis system (Elite) with passive markers on the throcanter, center of the knee and lateral malleolus of each leg. Kinematics of the oscillation was recorded together with E.M.G. of the quadriceps muscle both for the affected and controlateral leg. The test was implemented first monolaterally and then bilaterally,that is with both legs tested at the same time.

Results and discussion: E.M.G. and kinematics (angle versus time) exhibited a characteristic pattern during the first descent in the bilateral test, consisting in the appearance of the stretch reflex for the affected leg, and an abnormal E.M.G. activity, subsiding during the descent, for the unaffected leg, see fig.1. Since the relaxation index loses its meaning if the stretch reflex is not elicited, we could not use it for the controlateral leg. We there fore defined a new index called "Deceleration Index" shown in fig.2. The Deceleration index was out of normal range for the controlateral leg of 15 patients out of 18, even in very recent patients whose affected leg had not yet had time to develop spastic behaviour.

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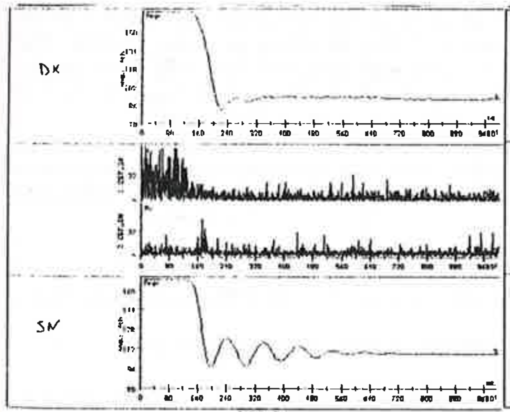

FIG.1

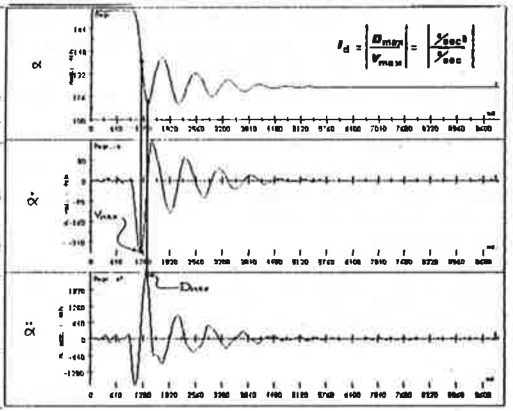

FIG.2 INDEX OF DECELERATION

QUANTITATIVE ANALYSIS OF MOTOR FUNCTIONS IN PATIENTS WITH PARKINSON'S DISEASE

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The isokinetic test is usually used in Orthopedic and Sports Medicine for an objective evaluation of multiple parameters regarding muscle function. The application of this method has only recently been widened to include neurological pathologies (Multiple Sclerosis, Strokes). At present there are very few studies regarding the use of isokinetic in the evaluation of motor function in Parkinson's Disease (P.D.)(3,4).

OBJECTIVES: The objectives of the study were: 1) to evaluate the variation of performance of P.D. patients in the isokinetic test during the ON phase (under drug therapy) and the OFF phase (after a period of 12-16 hours of wash-out), 2) to determine up to which degree of P.D.(mild, moderate, severe) the isokinetic test can be used as a reliable index of motor performance.

SUBJECTS: The group examined was composed of 16 subjects (9 males and 7 females, mean age of X=65,2 years, S.D.=8,2, Range=42,1-73,2) with P.D. at various levels of neurological severity (Modified Hoehn and Yahr staging: 1-4)(1,2) and average duration of disease X=7,1 years, S.D.=2,8, Range=2-11. None of the patients presented with other major diseases. All were out-patients and all volunteered after giving their informed consent.

METHODS: Preliminary testing with the Unified Parkinson's Disease Rating Scale (U.P.D.R.S.)(1) was performed in both ON and OFF conditions to determine the severity of the disease. All subjects were evaluated objectively during the execution of concentric exercises and isometric exercises using the modified Cybex 340 Equipment (Lumex, Bayshore, New York). The subjects were required to recruit the maximum muscular force during the concentric test in flexion-extension of the knee (angular velocities: 90-120-150°/s) and the elbow (angular velocities: 60-90-120°/s) and during isometric contractions (angular velocity: 0°/s) of the elbow flexor muscles and the quadricep muscles of the thigh. The dominant side was usually tested, but when there was a significant difference between the two hemisoma the more affected side was studied. The tests were repeated in three different occasions: the first trial was a learning situation aimed at preparing for the correct execution of the test, the second trial was during the ON phase, the third trial during the OFF phase. During concentric contractions the following values were determined: Peak Torque, Total Work, Average Power; during isometric contractions the values for Peak Torque and Average Torque were determined. The obtained results were analized using Student's T test (paired and unpaired sample test).

RESULTS: At the U.P.D.R.S. all patients showed significant score differences during phase ON and during phase OFF (ON better than OFF). An analogous variation was shown by analyzing the Motor Examination Subtest of the U.P.D.R.S.. In the isokinetic test the patients showed different levels of performance that were correlated with neurological severity. As a result we were able to identify two groups: the first group consisted of 11 patients with Hoehn and Yahr rating from 1 to 2,5; the second group consisted of 5 patients with Hoehn Yahr from 3 to 4 (between the two groups there was no significant difference in age and length of illness but there was significant difference for neurological severity: P<.001). The subjects of the first group showed significant improvement of motor performance during the ON phase in all the isokinetic tests (always P<.01 except: Peak Torque, Total Work and Average Power at 120°/s in elbow extension, P<.05). The subjects of the second group showed no significant difference from ON and OFF. The second group obtained levels of force production significantly inferior to the first group especially during phase ON (P<.01).

CONCLUSIONS: From our results it can be concluded that the isokinetic test is not reliable in patients with sever P.D., while it can be used in patients with mild to moderate disease to monitor the efficacy of drug treatment, to evaluate the benefits of rehabilitative programs and to follow the clinical course of the disease.

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A CORRELATION OF DYSPHAGIA AND TRUNCAL BALANCE WITH RESPECT TO SECOND COMPONENT OF THE BLINK REFLEX IN PATIENTS WITH SUPRATENTORIAL STROKE BY USE OF THE DEVISED METHOD

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Amid the controversy regarding the pathways of blink reflex (BR), trials of clinical applications of the reflexes to intracranial lesion have been investigated. There remain the problems of eliciting and recording these reflexes especially of the second component (R2), which use to be difficult in determining its latency due to the fluctuation by many causes such as conscious level, supratentorial influences that modify the BR responses.

We investigated the correlation between the R2 of BR and dysphagia and or truncal imbalance in stroke patients with supratentorial lesions by using our devised method to get R2 latency more clearly.

Sixty-nine chronic patients with stroke and 18 healthy controls were participated in this study. Patients with dysphagia (fluoroscopic examination was applied along with clinical symptoms for diagnosis) and truncal imbalance were classified into three degrees according to the severity.

Electrical stimulation of the supraorbital nerve was applied through bipolar surface electrodes and consisted of squared pulses with around 40mA, 0.2msec duration. The signal was filtered (band pass : 0.5 - 3000 Hz). The activity of the orbicularis oculi muscle was recorded bilaterally by silver disc surface electrodes. A paired pick up electrodes were fixed at just below the medial / lateral eye angle of lower eyelid. The impedance was reduced to below 5 kohms. A large T-shaped silver plate was horizontally fixed over the bilateral supraorbital margins for ground.

The electrical stimulations were given at random by manual switch. While watching the oscilloscope monitoring, subjects were kept in awake in order to give the stimulations to the very moments the background EMG noise was at its minimum level. The recordings were made by single stimulation or electronic signal averaging for 10 to 20 times and each sweeps were doubly superimposed to measure R2 latency.

Significant correlation was revealed between the degree of dysphagia and truncal balance and also between the former two factors and R2 latency (more prolonged latency in proportion to hither degree of dysphagia / truncal imbalance). The R2 latency is difficult to determine in severe stroke cases due to its fluctuation by usual methods. These are presumably due to a lowered excitability of the brain stem trigeminal and or facial systems and reticular formation associated with the conscious level. The alterations and decrease of R2 components in stroke patients could be explained by a loss of facilitatory hemispheral influences to both sides of the brain stem due to a lesion of cortico-bulbar fibers. In control, the fluctuation of R2 latency is within 2 msec in awake condition (our Lab. data). In strokes, the alterations increase in proportion to severity of disability. It may be partly due to drowsy prone condition during the electrodiagnostic testing. With the manual averaging technic while watching patients condition carefully, the determination of R2 onset became much easier.

OPTIMUM STIMULUS PARAMETERS FOR CONSCIOUSNESS LEVEL ESTIMATION

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We have been studied the sleep monitor for monitoring the various data on central nervous system and circulation system during sleep(1). The microcomputer system analyses the data on EMG, EEG,ECG, EOG,temperature, body movement and voice. The system estimates the sleep stage, and operates data communication system or air controller. By this system,it possible to help the muslte function of the handicapped or aged people with mustle disorder. Application of the ABR(auditory brainstem response(2))to this system, should make this monitor useful for the consciousness disorder.

In this paper, we report optimum stimulus parameters ; sound level, duration and interval for the ABR with sharp peaks and with shorten latency, and the 3-D model which shows relation between parameters for such response. Normal ABR waveforms which were recorded at sound level under 95 HL, duration between 0.1 and 1.0 ms , and interval under 50ms (between 20 Hz and 80Hz), analysed by Fourier series. Regarding coefficients of the expanded terms of Fourier series as a vector,Auto-correl -ation and cross-correlation of given ABR waveforms were caluculated.

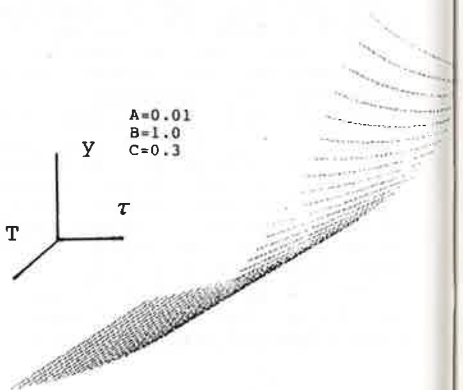
The waveform is initially flat at the lower sound level, 1.0ms duration and at 80Hz but sharp peak gradually appears and the peak latency shows delay and nonlinear shift tendency with the increase of the sound level the interval and with the decrease of the duration. The auto-and cross-correlation reflect these affairs at any average counts. And it conform -s the ABR shows shorten latency and saturation tendency at near 80 dB HL sound level, 0.1ms duration and at 50ms interval.

In modelling on optimum stimulus parameters,we consider follow -ing equation.

$$y = C\,exp\left(\frac{B\tau}{1-AT\tau}\right) \quad (1)$$

y : ABR latency. A,B,C : arbitrary **constant**. T : stimulus click interval τ : click duration(ms). $1-AT\tau \neq 0$

This equation establishes the 3-D model of nonlinearity of the peak latency corresponding to stimulus parameters,which may show neural delay circuit by mainly synaptic function.



the 3-D model

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REACTION TIME PERFORMANCE IN POST-ACUTE STROKE PATIENTS

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INTRODUCTION

Some texts point out that the most part of post-stroke functional recovery happens within six months. Functional recovery of post-acute stroke (PAS) patients has been related to mental processing (e.g. learning) rather than to spontaneous recovery. The simple reaction time (SRT) performance is considered the simplest psychomotor performance and the archetype of volitional act.

PURPOSE

This study was carried out in order to get more information on psychomotor performances on 6 months post-stroke period (PSP).

METHODS

29 consecutive PAS patients admitted to our rehabilitation center were studied.

All the patients had single cerebral lesions and were examined at a post stroke period from 33 to 180 days (1 to 6 months).

Their age ranged from 51 to 66 years. We divided them into two groups: short-PSP and medium-PSP. In the short-PSP group were included the patients with a post stroke period from 33 to 90 days.In the medium-PSP group were included the patients with a post stroke period from 90 to 180 days.

We excluded patients with: important visual impairment, severe aphasia, severe neuro-psychological impairment, daily therapy with psycho-active drugs.

In the first week of admittance SRT was calculated by a computer-aided system. The stimuli appeared on the screen of a personal computer and the patients had to press the space bar of the keyboard using the hand homolateral to the side of the cerebral lesion.

RESULTS

31% of all patients (9 of 29) had pathological SRT at the admittance. Only 17,6% of the short-PSP group (3 of 17 patients) showed pathological SRT. On the contrary, 50% of the medium-PSP group (6 of 12 patients) showed pathological SRT.

DISCUSSION

Judging from the present set of results, a high percentage of PAS patients admitted to our rehabilitation center had pathological psychomotor performances.

Surprisingly the highest percentage of pathological SRT was found in patients with PSP of 3 months or more. We have no clear explanation for this.

Anyway, this study suggest that it could be advisable to evaluate psychomotor performances in clinical setting of post-acute stroke patients.

THE ASSESSMENT OF L-DOPA EFFECT ON PATHOLOGICAL WALKING PATTERN AND MOBILITY OF PARKINSONIAN PATIENTS

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Introduction. The abnormalities in Parkinsonian walking pattern are well known. Using surface EMG recording, it is possible to quantify the various kind of pathological pattern of walking. The present study intended to reveal the effect of L-DOPA treatment on pathological walking pattern in Parkinsonian in time course during long-term treatment.

Methods. To quantify the pathological walking the muscle activation pattern from m.biceps femoris, m.vastus lateralis m.gastrocnemius medialis and m.tibialis anterior on both legs was studied using EMG recordings. Analysis of video recording were also included. Pressure-sensitive transducers were mounted on both feets in order to monitor the pressure profile. The time period between two heel contacts was considered 100%. EMG signals were full-wave rectified followed by a low-pass filtering and stored in the computer. Later on the profiles were calculated. In order to minimize the inter-individual and intrasubject's variances the amplitudes of the individual EMG profiles were normalized to the mean EMG amplitude and the mean amplitude was adjusted to 100%. The recordings were made every half hour before and after L-DOPA treatment during four hours. EMG and video recordings were compared. The patients were required to keep walking for approximately 60 sec.

Results. Analysis of the walking pattern during free-speed walking indicates that LDOPA caused a considerable change in walking and mobility. The amplitude of leg's movement were found to increase after the L-DOPA dose. Strides were found to increase in length. Several different pattern of control of the movements in relation to on/off phases of disease and effect of treatment could be distinguished. The amplitude and frequency of associated movements in upper limbs remain unchanged during time course.

Conclusion. The EMG activation and coordination can be used to quantify the effect of treatment on the pathological walking pattern in Parkinsonians.

THE USE OF CAD TECHNIQUES FOR THE PREOPERATIVE PLANNING OF INTERTROCANTHERIC OSTEOTOMIES

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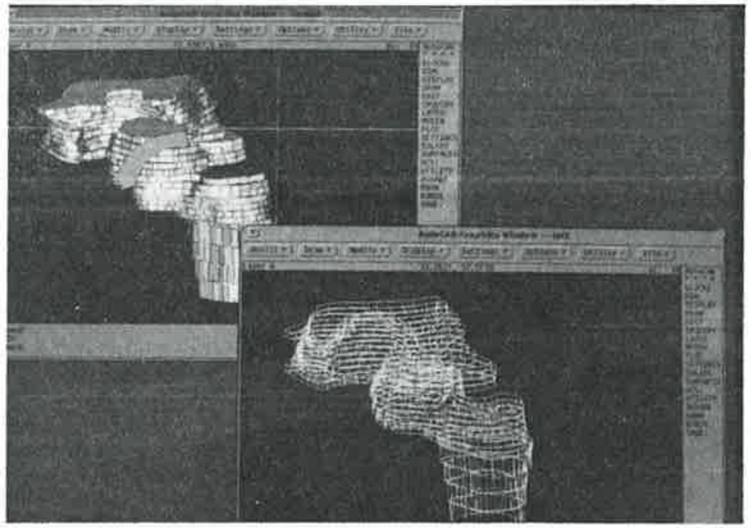
Keywords:CAD,Image Processing,Orthopaedics

In the last ten years Tomographic techniques (TAC, NMR etc.) has been allowed to visualize in all 3 dimensions the internal structure of human body.
These innovations will be used currently in the next future for the radiological diagnosis; furthermore they might be even more useful in a new application field like "Preoperative Planning","Surgical simulation" or "Computer Assisted Surgery".
These keywords mean all those techniques that, on the basis of 3dimensional anatomic informations, allow to simulate, plans or assist surgical intervention. Of course a full simulation or planning is very hard because they would require the knowledge of all the possible events that can occur in the operating theatre.
We noticed that CAD techniques also could be used in the "Computer Assisted Surgery" applications; of course the informations source is always a tomographic set of images, but in this case, through appropriate calculations, a "CAD" solid model is generated starting from the volume based model.
Of course while the volume contains all the available informations, the solid model concerns only the requirements of the surgical assistance.
The solid model can be used either to get morphometric informations about the bone structure, or to perform modelling operations like cut, join, move and so on.Several AUTOCAD commands are available to change the view point and to query morphometric informations about the solid model.(Fig)



Acknowledgments
This work is financially supported by C.N.R. and M.U.R.S.T. Many thanks are due to Mrs Sandra Thompson for the cooperation in the english translation.

AN ERGONOMIC APPROACH TO THE ASSESSMENT AND PREVENTION OF DEGENERATIVE DISEASES OF THE SPINAL COLUMN IN WORKING POPULATIONS

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In many industrialized countries, degenerative alterations of the rachis, especially those of the lower back and the joints of the upper limbs are certainly on the increase.
pIn order to confront this problem increasing in diffusion also in Italy, an interdisciplinary research unit was set up involving the Institute of Occupational Medicine in Milan, the Department of Electronics at Milan Polytechnic, the Faculty of Engineering, the Bioengineering Center of the "Pro Juventute Don Gnocchi" Foundation and the medical service of Milan's Local Health unit.
Within this research unit, occupational doctors, physiotherapy graduates, other qualified physiotherapists, industrial designers, ergonomists, bioengineers and experts in anthropology were working together.
The main aim was essentially to analyze and evaluate work spaces, describe them in quantitative terms as much as possible in an attempt to identify any associated risks. Then, clinical examinations was performed on workers in order to evaluate what these risks might lead to in terms of injury of the spine.
A series of laboratory instruments were developed in association with the bioengineers and a procedure for clinical examination has also been developed which allows information to be collected without radiographic examination of the spinal column.
Furthermore, it has been possible to transfer this methodology to a series of research units which can be found within the medical services departments of both public and private bodies.
Seventeen of these units have been set up in local health authority offices ;as far as universities are concerned, it has been possible to transfer this method to six occupational medicine departments of the University and also the health services of three major Italian companies have shown considerable interest and brought together thousands of workers.
It was so possible to carry on a series of research studies on various people(more than 10.000) whose work requires intense physical effort (quarry workers, grave diggers, porters) and on others who are exposed to mechanical vibrations transmitted to the entire body (drivers of heavy vehicles or motorcycles).
In other cases, the research is concerned with subjects suffering afflictions caused by excessive sedentarianism - such as workers who use a video screen, telephone operators, goldsmiths,etc.
In some cases our data as enabled us to formulate hypotheses regarding a cause and effect relationship between risk and injury: in other cases, the data permits us even greater certainty. For example, a study carried out on a group of approximately 350 video terminal workers makes it reasonable to suppose that one of the factors involved in the incidence of low back pain is incongruous posture.

AEROBIC AND ANAEROBIC METABOLISM DURING LOCOMOTION WITH TWO DIFFERENT WHEELCHAIR TYPES

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Wheelchair design is extremely important in order to improve efficiency of locomotion and reduce physical stress in subjects whose muscular and cardiopulmonary fitness are impaired.
Purpose of this study was to evaluate the effect of different wheelchair design on the aerobic and anaerobic metabolism during locomotion at different speeds in paraplegic subjects.
The experiments were carried out on a group of 5 male paraplegic subjects (25±3 years; body weight 65±7 kg) during locomotion on a roller ergometer (Sopur, Ergotronic mod.) at 3-4 different speeds from 2 to 9 km/h. At each speed oxygen consumption and heart rate were determined after at least 6 min of exercise. Lactic acid (LA) venous blood concentration was evaluated before and at the 5th min of recovery and lactate production was calculated. The oxygen equivalent of LA was assumed to be 3,15 ml O_2 per kg body weight for an increase of blood LA of 1 mmol/l. For each subject the test was repeated using two different types of daily use active wheelchairs: type A, foldable, 13,95 kg; type B, demountable, 13,35 kg. The main difference in size was in the horizontal location of the wheel axle, in seat height and in hand-rim diameter.

Results indicate that:

a) oxygen consumption increased linearly with speed being: 2050±350 ml/min and 1780±270 ml/min at 9 km/h for wheelchair type A and B, respectively;
b) lactic acid concentrations were significantly higher, at a given speed, while using wheelchair type A than B (at 9 km/h; 7,4±1,5 mmol/l and 6,0±1,6 mmol/l, respectively);
c) the total energy required, aerobic and anaerobic, increased linearly with speed and was 15-20% higher with wheelchair type A than B at all speeds;
d) the energy cost of locomotion at a given speed was in the 15-25% range higher for wheelchair A than B;
e) at corresponding oxygen uptake, heart rate and pulmonary ventilation were not different with the two wheelchair types.

The main results of this study concern the large difference existing in the energy cost of locomotion and in the lactate production in the same subject when two different wheelchairs, even if apparently similar, are used. In particular the much higher lactate production suggests that wheelchair design affects the limb and trunk movements in such a way that the metabolism of some muscle group requires a greater participation of anaerobic mechanism of energy supply, this leading to early onset of muscular fatigue. Further studies, in particular the combined biomechanical analysis of user and wheelchair during locomotion, are required to increase the optimum fitting of wheelchair-user interface.

TONIC NECK REFLEX IMPROVEMENT OF THE WORK CAPACITY IN MAN

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Tonic neck reflexes (TNR) are often assumed to be included in the building of voluntary motor programs. The well-known dynamic postures illustrated with photographs by Fukuda (1979), strongly suggested that these reflexes play a role in the postures which are spontaneously adopted in various sport activities. Hellebrandt et al. (1956) also demonstrated that there was a strong improvement of the work capacitiy -or resistance to fatigue- when subjects turn their heads in a way in accordance with the facilitating tonic neck reflex posture described by Magnus in the case of decerebrate cats. Nevertheless, in Hellebrandt et al. studies, subjects were not only instructed to turn their head in a given way but were allowed to progressively adopt complicate and probably facilitating postures when fatigue arised. Thus one must ask: were it no changes in the posture would the performance be facilitated by the only rotation of the head? The present study was designed to answer this question.

Subjects were set in an experimental situation close to the original one used by Hellebrandt et al., but in a fixed restrained posture from the beginning to the end of each experimental session. They repeatedly lifted weights by extending the elbow with the head rotated either towards the side of the active upper limb or towards the opposite side. Frequency of the movements was .5 Hz, and initial amplitude 30 degrees. Successive series of 15 movements separated by a rest time of duration equal to the preceding series of movements were performed by each subject until exhaustion, i.e. until they were unable to repeat the movement at the initial frequency. The mechanical work performed during each series of movements was calculated. Surface EMGs from triceps brachii, anconeus, and biceps brachii muscles were recorded and then integrated over the whole duration of each series of movements.

With successive series of lifting, movements amplitude progressively decreased under the influence of fatigue. Thus there was a decrement in the work produced during the successive series. In agreement with Hellebrandt et al. data as well as with the expected effect of a TNR, the decrement of work was less important when subjects turns their heads towards the moving limb. However, in contrast with the results from these authors, the facilitation was very low and only statistically significant for the two last series. During these series, the integrated EMG values from all the recorded muscles where also slightly higher with the head in the "facilitating" position. However, the gain in EMG amplitude was not statistically significant for all the subjects.

From these results, we concluded that TNR can slightly influence the amount of work that a subject can produce. But a direct facilitation exerted by TNR on the prime mover muscles is lower than the one exerted through compensatory postures.

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JOINT LOAD AND LUMBAR COMPRESSION CAUSED BY MANUAL MATERIAL HANDLING ON A MOVING SURFACE - A COMPUTERIZED ANALYSIS

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Work on a moving platform often exposes the worker to accelerations that involve high forces and moments on the body segments. Additional external load caused by e.g. manual material handling enhances the forces further. It is assumed that these forces increase the musculo-skeletal load implying a risk of injuries. In the present project the effect of ship motions on the joint load during manual work on a fishing boat is studied.

The motions of a Swedish medium sized fishing vessel were registered in three degrees of freedom by means of accelerometers and a pendulum. Simultaneously the posture of a fisherman, with reflecting markers on nine joint centres, was recorded on a video tape. Three simplified working tasks, common in fishery, were studied; standing erect unloaded (I), standing erect holding a basket of 21 kg (II), and repeatedly lifting this basket from the deck onto a bench in front of him, and down again (III). The two dimensional coordinates of the nine joint centres were digitized and synchronized with the ship motion data at a sampling rate of five Hz.

All data were processed in a biomechanical model, developed within the project. The moment and vertical and horizontal forces at seven joint centres - elbows, shoulders, C7, L4/L5, hips, knees and ankles, and the lumbar compression at the L4/L5 level were calculated. The model was two-dimensional in the sagittal plane, symmetric and dynamic. Necessary parameters - centres of mass, moment arms, and radius of gyration - were calculated from the subjects weight and the length of the body segments. Acceleration of the segments was calculated using coordinates of three consecutive projections. Joint moments and lumbar compression were also calculated for a static posture with the subject standing erect and unloaded in still conditions.

In spite of mild weather conditions, moments and lumbar compression forces causing considerable musculo-skeletal strain were calculated. Standing unloaded on the ship (sequence I) induced lumbar compression of similar magnitude as when standing in still conditions. The compression in sequence II was more than twice as high as in sequence I. In sequence III the range of compression values was wide with upper values more than twice as high as those in sequence II. The highest compression values were observed when the subject was lifting the basket from the bench (approximately 4000 N).

In the present study it is shown that the musculo-skeletal load does increase during work on a moving surface. Furthermore, it is likely that muscular over-stabilization of the joints occurs. It is therefore probable that the model gives conservative estimates of the lumbar compression. The project includes checking the amount of antagonistic muscle activity by EMG techniques. In spite of the small motions, high values of moment and lumbar compression were present, and thus it is justified to assume that in conditions of prolonged and large surface motions, which is common in fishery, lumbar compressions will result which create a definite risk of back injuries.

The model used is two-dimensional and thus acceleration in the sagittal plane is considered. At present the model is being developed to comprise three dimensions - ship motions in six degrees of freedom is registered using six accelerometers and the three dimensional coordinates of 16 joint centres are measured.

MUSUCULAR LOAD IN CASH-REGISTER WORK

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In spite of ergonomic improvements (1), musculoskeletal problems are still a serious occupational hazard (2). The aim of the present study was to evaluate the myoelectric activity in some shoulder and arm muscles in cash-register workers during work at their normal work place. Twelve healthy female cash-register workers aged 19 - 51 years volunteered in the study.

Electromyographic activity was recorded while they were handling and registering goods placed on the conveyor belt. The cash-register machine was equipped with a scanner standing vertically beyond the conveyor belt. One task was to handle and register goods which were piled up on the conveyor belt randomly, even upside down, forming a "mountain". The cashier had to lift every piece in order to find the bar codes to be registered. The other task was to handle and register goods placed by the customer on the conveyor belt in line, in most cases with the bar codes towards the scanner. Each task was performed for 20 minutes. The conveyor belt and goods were running from left to right. Thus, the goods were mainly handled by the left arm of the cashier.

Electromyographic signals were recorded bilaterally from the trapezius muscles, and on the left side from the anterior part of the deltoideus, the common belly of the biceps brachii, the extensor carpi radialis brevis, and the flexor digitorum superficialis muscles. Bipolar surface electrodes were used. The myoelectric signals were transmitted by telemetry and recorded on a FM tape recorder. The root mean square (RMS) values of the myoelectric signal amplitudes during the work tasks were analysed using a computer.

It was shown that the mean RMS values were significantly lower ($p<0.01$) in all the examined muscles on the left side when the goods were placed in line compared to the situation where the goods were piled up randomly on the conveyor belt.

It is apparent that by asking the customers to arrange the goods properly on the conveyor belt, it is possible to reduce the muscular load of cash-register workers.

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MULTIFACTOR ANALYSIS OF WHEELCHAIR HAND PROPULSION

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The aim of this study is to define a method for functional evaluation of wheelchair propulsion motor strategies using biomechanical and electromyographic (EMG) approaches. In particular, the effects of wheelchair geometric modification on motor strategies have been evaluated. The final goal is to optimise wheelchair configuration with respect to individual users.

Ten paraplegic subjects performed several trials on a roller ergometer with the same wheelchair model, characterised by different geometric configurations. A multifactor measurement system consisting of motion analyser (ELITE) and 8 channels of EMG was used to obtain anatomical and wheelchair landmarks coordinates, and EMG activity of upper limb and trunk muscles. Upper limb joint angles, trunk attitude and wheel rotation have been computed. Identification of temporal phases was performed via a specific algorithm [1]. In a separate acquisition session on the same subjects, vocational EMG [2] was used to monitor the amount of each muscle effort during a five minute trial.

Figure 1 shows elbow (a) and shoulder (b) joint excursions and EMG activity of medium deltoid, biceps and triceps muscles together with time markers of different phases during a complete cycle. Propulsive phase can be divided into two subphases: pull phase, identified by elbow flexion and characterised by biceps activity, and push phase, in which elbow extends, with an initial role played by triceps, and shoulder flexes. Anterior deltoid begins to act at the end of push, probably because of the camber wheel angle, and continues to abduct the arm allowing a safe backward movement of the upper limb during recovery phase.

Figure 2 shows the graphic output of vocational EMG data as an histogram of time percentage of muscle activation versus activation level, as a percentage of the maximum voluntary contraction.

By comparing results obtained with different wheelchair geometric configurations it is possible to identify the optimal one. Next step is the correlation analysis between optimal configuration and both subject anthropometric parameters and level functionality.

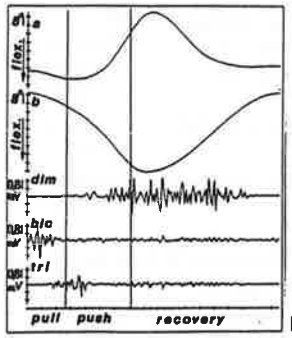

Fig.1

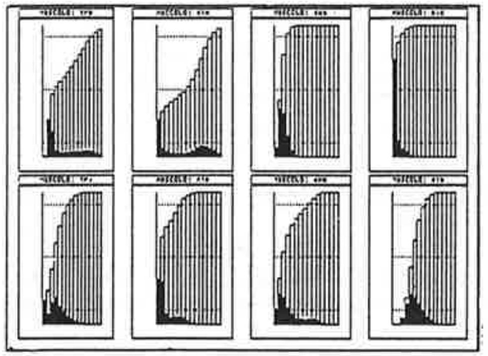
Fig.2

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EMG ACTIVITY OF THE UPPER TRAPEZIUS AS A FUNCTION OF ORGANIZATION OF ASSEMBLY LINE WORK

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Introduction: A mismatch between workstation design and anthropometric measures has often been shown to increase muscular load. For example, working height above elbow height increases the load on trapezius. Thus, adjusting the workstations are common intervention measures taken by ergonomists. Another factor, which may also influence the muscular load but has not been considered extensively by ergonomists, is the success of production engineers in creating a balanced and fast flow along the assembly line. This factor is investigated in the present study.

Materials and Methods: Fifteen subjects from two assembly lines, a Chinese (CAL) and a Swedish (SAL), participated in this study. Working height and elbow height were measured and compared. Organization of assembly lines was evaluated by the MTM (Methods-Time Measurement), and by observations on buffer conditions and allocations of utility men. These three aspects are generally considered being important to the balance of the assembly line. EMG activity of the upper trapezius was picked up during normal assembly work. The signals were normalized according to the signal level during a reference voluntary contraction (RVC) (Hägg, et al. 1987). The cumulative distribution of the EMG amplitudes during work (APDF analysis) was computed (Jonsson 1982) to obtain a measure of static (probability level P=0.1), median (P=0.5) and peak (P=0.9) load levels. A new method (EVA) described by Mathiassen and Winkel (1991) was also used to evaluate the variation of muscular activity.

Results: At both plants, too high working levels were observed. This may contribute to the high static muscular load level of about 20%RVC (≈3%MVC) observed at both plants using the APDF analysis (Figure 1). The APDF analysis showed similar loads at the median and peak levels at the two plants as well.

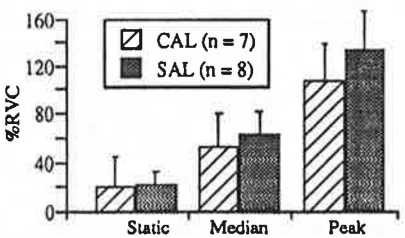

Figure 1. Comparison of Muscular load on the upper trapezius between groups (Mean + SD)

According to the results of the MTM study, the workstations at SAL had more uniform cycle times than at CAL (Figure 2). In addition, buffers were larger at SAL (about 1 hour) than at CAL (a few minutes). Finally, at SAL a B-side arrangement was constructed along the line where utility men could work whenever necessary. No such arrangement was observed at CAL. Consequently, a better product flow was seen at SAL than at CAL. This may well explain the longer periods of low muscle activity at CAL as compared to SAL (Figure 3).

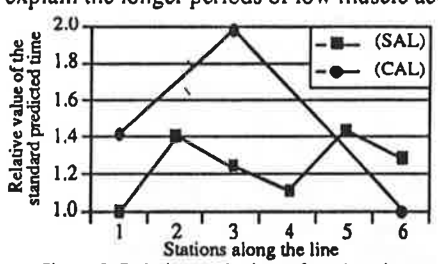

Figure 2. Relative cycle time of workstations along the two lines

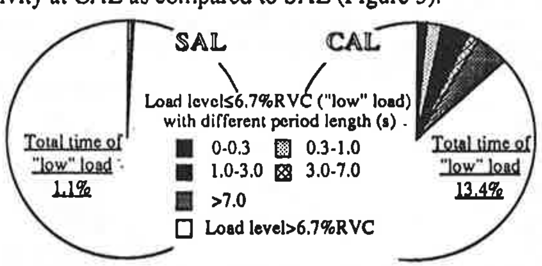

Figure 3. Proportions of "low" muscle activity between the two groups (EVA)

Conclusions: The success of production engineers strongly influences the load pattern on the upper trapezius in assembly line work. This should be considered in ergonomics interventions.

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EFFECTS OF SITTING WITH ANTERIOR CHEST SUPPORT ON MUSCULAR ACTIVITY AND SUBJECTIVE RATING OF EXERTION DURING MICROTOME SECTIONING

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The aim of the study was to analyse the effect of anterior chest support on muscular activity in the posterior neck muscles, shoulder and low back muscles in sitting subjects simulating microtome sectioning.

Methods: Seven females, six laboratory assistents, participated in the experimental part of the study. The level of muscular activity in different sitting work postures was recorded, using surface electrodes, as full-wave-rectified and low-pass filtered EMG, and normalized. Six of the subjects also rated the degree of exertion in four different body regions during their ordinary microtome sectioning work, comparing the use of anterior chest support with the use of their usual chair with lumbar support.

Results: Working with anterior chest support in a vertical trunk posture reduced lumbar back muscle activity and increased shoulder muscle activity. Arm movement increased the back muscle activity in all postures. The subjects rated exertion higher in nape, shoulders and thoracic spine when using the anterior chest support chair than when using their ordinary chair.

Conclusions: The use of anterior chest support reduced the activity in the lumbar back but increased it in the neck and shoulder. Perceived exertion in the neck-and-shoulder region increased. Anterior chest support does not seem to solve the problem with neck and shoulder pain during the work of preparing laboratory sections.

METHODS FOR MEASURING ISOMETRIC ENDURANCE OF SHOULDER FLEXORS AND CERVICAL SPINE EXTENSORS IN PATIENTS WITH CHRONIC NECK-SHOULDER-ARM PAIN

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The aim of the study was to develop and describe methods for measuring isometric endurance of shoulder flexors and cervical spine extensors in patients with chronic musculoskeletal pain.

Methods: 23 female patients (mean age 38 years) with chronic neck-shoulder-arm pain participated. The median pain duration was 19 months. The shoulder flexor test was performed in the sitting position with the right arm kept straight and horizontal. A weight of 25 N was applied close to the wrist. Exertion in the shoulder flexors was rated on a 0-14 scale every 15 s and at cessation. The cervical spine extensors were tested with an extra load of 50 N on the head (a helmet plus applied weight) and with the cervical spine 45 degrees flexed. Neck muscle exertion was rated in the same way, but here the test was broken off after 180 s. The values obtained were compared with those of a randomly selected reference group (Schuldt et al 1991, Harms-Ringdahl et al 1991).

Results: The patients' (n=23) average flexor endurance was 41.9 s (right) and 45.1 s (left) before the programme; 52.3 s and 54.6 s at the end and 50.1 s and 52.0 s three months later. The corresponding endurance time for a female reference group (n=29) was 95 s. Rated exertion (on a 0-14 scale) in the shoulder muscles at cessation of the endurance test was 13.7 (right) and 13.0 (left) before the programme; 13.4 and 13.3 at the end; and 13.2 and 13.0 three months later. The corresponding rated exertion of a reference group was 13.3.
The patients' mean cervical epine extensor endurance time was 97.7 s before the programme; 78.5 s during the last week, and 89.7 s three months after the end of the programme. The rated exertion in the neck at cessation of the test was 13.3 (before), 13.5 (last week) and 13.3 (3 months later). The corresponding endurance time for a female reference group (n=85) was 173.3 s (max time allowed 180 s) and the rated exertion was 10.3 on the same 0-14 scale.

Conclusion: The two endurance tests showed much lower values for patients with neck and shoulder pain than for a randomized reference group. The rated exertion at cessation of the tests was the same for the shoulder flexor test in both groups and much higher for the neck extensor test in the patient group. Shoulder flexor endurance increased after exercise.

INSTABILITY OF SURFACE EMG RECORDINGS FROM THE UPPER TRAPEZIUS MUSCLE

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Introduction. Bipolar surface EMG recordings from the upper trapezius muscle are often used as an indicator of shoulder muscle load in occupational work tasks. The standard recording position is a symmetrical placement of the electrodes about the midpoint ("central lead point") between acromion and the spine of 7th cervical vertebra ("lead line", Zipp 1982). An instability of the EMG amplitude in this position has been noticed, as exemplified by a 3-fold difference in the EMG amplitude at maximal force in shoulder elevation, compared to 90° arm abduction (Westgaard 1988). A central depression in the EMG amplitude when moving the electrodes along the length of the trapezius muscle fibres was recently described (Veiersted 1991), and a possible relation between this phenomena and the instability of EMG recordings was examined.
Methods. Eleven subjects participated in the study. A 16-ch bipolar array electrode was placed along the lead line, spanning the centre point. Each set of bipolar electrodes had an interelectrode distance of 20 mm and each electrode a diameter of 3 mm. The distance between adjacent sets of electrodes was 5 mm, resulting in a total recording span of 7.5 cm. EMG signals were recorded at maximal force in isometric shoulder elevation with a sling placed over the acromeon, alternatively with the arms in 90° abduction and the sling just proximal to the elbow. Continuous force-EMG calibration curves in shoulder elevation were generated. The EMG signals were full-wave rectified and integrated at 0.2 s.
Results. The EMG amplitude along the trapezius muscle fibres of all subjects showed a characteristic profile: relatively low near the acromion end, increasing at more medial locations to reach a plateau, about 2 cm wide, just lateral to the central lead point. A reduction in amplitude of about 80 % consistently occurred near the central lead point, increasing again at more proximal locations. In the midpoint the EMG amplitude at maximal force would vary considerably between shoulder elevation and 90° abduction, often with a ratio of 4. This ratio was near unity at the amplitude plateau lateral to the midpoint. The force-EMG calibration curve in shoulder elevation would sometimes have a triphasic shape near the midpoint, and the submaximal EMG signal was proportionally more elevated: a force of 10 %MVC corresponded to 16% of maximal EMG amplitude near the midpoint vs. 11% (mean values) at the lateral amplitude plateau.
Discussion. The central depression in EMG amplitude is most likely a consequence of a differential EMG recording spanning the muscle endplate region. This placement may result in cancellation of EMG signal components, due to symmetrical muscle action potentials spreading in both directions from the endplate. The variation in EMG amplitude at this position may be caused by a movement of the surface electrode relative to the underlying muscle, and introduces an uncertainty when attempting to evaluate muscle load in occupational situations demanding flexion or abduction of the upper arms. Consequently, the EMG surface electrodes for evaluation of load on the upper trapezius muscle should be placed lateral (ca. 15 mm) to the midpoint between acromion and the 7th cervical spine.

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SPECTRUM ANALYSIS IN DELAYED MUSCLE PAIN INDUCED BY ECCENTRIC WORK

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The aim of the present study is to estimate, through power spectrum analysis of EMG signals, the modifications in muscular contractions that are induced by eccentric strain. 7 healthy volunteers (5 male and 2 female) between the ages of 24 and 35 years were tested. The subjects performed a step test in such a manner as to force the right quadriceps muscle to perform only concentric contractions, while the controlateral quadriceps muscle performed eccentric contractions. Before performing the test, immediately after, and every 24 hr thereafter for the next 4 days, the volunteers performed isometric contractions of the quadriceps muscle at 30% of maximal force, which had previously determined. During this contraction, EMG signals were recorded from the rectus femoris muscle- using superficial electrodes. Beginning with the attainment of the requested level of contraction, recordings were made for 400 msec. The tracings thus obtained were subjected to spectrum analysis by means of the Fourier Fast Transform (FFT)!. The median frequency (FMED) of the power spectrum of the EMG signal was taken as the measurable parameter. Concurrently, the following were assessed: the intensity of pain, which was assessed by means of the Analogical Visual Scale of Scott-HusKisson; the algogenic pressure threshold of the quadriceps femoris muscles of both legs; and the range of motion of the knee joint in relation to the type of muscular work performed. Compared to baseline, the FMED of the side that was subjected to concentric contraction did not demonstrate significant variations in subsequent evaluations. In contrast, the FMED of the side that was subjected to eccentric contraction was found to be significantly elevated immediately after ($p<0.04$), 24 hr after ($p <0.005$), 48 hr (p 0.02),and 72 hr after ($p< 0.05$) the test. The intensity of perceived painwas found to be maximal at 48 hr. Motility was found to be significantly reduced immediatly after ($p<0.02$), 24 hr after ($p<0.001$) until 72 hr after the test.It normalized after 96 hr.

TRAPEZIUS MUSCLE ACTIVITY DURING PSYCHOLOGICAL STRESS

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Introduction. The present pilot project was aimed at evaluating the importance of psychological stress with respect to trapezius muscle activity, using an established set of laboratory stressors and evaluating stress by means of stress hormone measures as well as physiological measurements.

The trapezius muscle in of particular interest to study in this context, since this muscle is not only a prime mover in the shoulder, but also a site for work related myalgia, a common syndrom in workers exposed to, e.g., light assembly tasks or work with visual display units. Previous studies (1,2) have indicated that there might exist a connection between psychological stress and tension in the upper part of the trapezius.

Material and method. Seven female university students aged 20-33, participated in the study. EMG from the upper part of the trapezius muscle was recorded bilaterally using surface electrodes. The subjects were exposed to a series of mental and physical stressors (arithmetic test, high force hand grip, stroop test, cold water test). Each test lasted for 1-5 minutes. Two minutes rest was allowed between successive tests. Urine sampling was taken before and after the experiment and analysis was done for stress hormones. Blood pressures and cardiac pulse rate were recorded. EMG activity level (RMS) was computed before, during, between and after tests and compared with activity before commencement of the provocation. Test contractions (elevated arms) were performed, and r.m.s. and mean frequency measures were computed.

Results. The results showed significant effects on blood pressure and pulse, but not on stress hormones. Nevertheless, there was a higher trapezius activity in most situations during stress provocation as well as in the intermediate rest periods, compared with rest before start of the provocation. For instance, this effect was significant at group level in the left trapezius, resting state ($p<0.05$). Here, the activity increased by 13 microvolts r.m.s. (about 30 per cent) on the average. However, there were vast differences between individuals in the EMG data. There was no correlation between stress hormone values and EMG activity augmentation. Nor were there any effects on r.m.s. values or mean frequency during test contractions during the experiment.

Conclusion. Experimental stress increased tension in the resting trapezius muscle. However, in some subjects this effect was marginal or even non existing.

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INVESTIGATION OF CORTICO-CORTICAL INTERACTIONS WITH MAGNETIC BRAIN STIMULATION

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Abstract

Connections to the motor cortex from other areas of brain have been investigated using two transcranial magnetic stimulators connected to focal figure-of-eight stimulating coils. A conditioning-test design was used in which the conditioning magnetic stimulus was applied over one area of brain, and the second magnetic stimulator was then used to test the excitability of the motor cortex at different times afterwards. Changes in the size of the test shock indicate that the conditioning stimulus has influenced the excitability of the cortico-spinal pathway at some level. In order to identify whether inhibition occurred at the cortex rather than the spinal cord two additional experiments were performed. In the first, we used H-reflexes in forearm flexor muscles to evaluate monosynaptic excitability at different times after the conditioning shock. In the second set of experiments, we compared the effect of the conditioning stimulus on responses elicited by low intensity anodal electric stimulation of the motor cortex. Lack of an effect of the conditioning stimulus on either H-reflexes or responses to anodal electric stimulation suggests that inhibition of a magnetic test volley occurred by influencing motor cortical excitability. Inhibitory connections from cerebellum to motor cortex, contralateral to ipsilateral motor cortex and cortico-cortical within the motor cortex itself have been described. Changes in the excitability of these connections can be observed during different types of voluntary movement, and in certain groups of neurological patients.

The advent of transcranial magnetic stimulation has made stimulation of the human brain, particularly the motor cortex, commonplace in clinical neurophysiology. With the introduction of the figure-of-eight design [1], more focal stimulation is possible than when using large circular coils, and this has allowed the motor strip [2], and to some extent the visual cortex [3], to be mapped topographically in some detail. Such focal stimulators have another advantage. Because the area which they activate is relatively small, we can now employ two such stimulators to investigate the connections between the motor cortex and other parts of the brain. Experiments have been described showing

inputs from muscle stretch receptors in the periphery to motor cortex [4], and from other areas of the brain including the cerebellum [5], the contralateral motor cortex [6] and neighbouring areas of the ipsilateral motor cortex itself [7]. In this report, I shall concentrate on the latter two effects as examples of this technique.

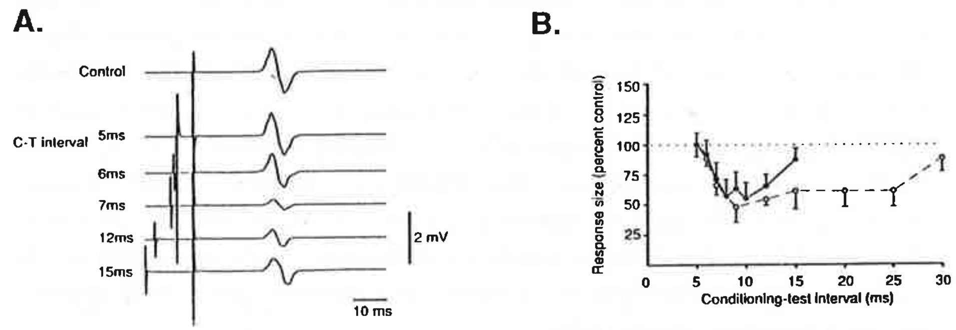

Figure 1
A, Effect of a conditioning magnetic stimulus delivered by a figure-of-eight coil (peak magnetic field rating of 2.4T) over the left motor cortex on EMG responses in the left first dorsal interosseous muscle produced by a magnetic test shock delivered by a second identical coil over the right hemisphere. The top trace shows the response to the test shock given alone. The lower five traces illustrate the effect of a conditioning shock given 5, 6, 7, 12 or 15ms before the test shock (C-T interval). The traces are aligned to the onset of the test shock. Each trace is the average of ten single trials. The subject was relaxed throughout. The intensity of both the test and the conditioning stimulus was 55% of maximum output. B, Mean ±1SEM time course of interhemispheric inhibition in six different normal subjects using two different intensities of conditioning shock. Responses were recorded in the relaxed FDI muscle, and the size of the conditioned response at each interval has been expressed as a percentage of the size of the test shock given alone (=100%). The conditioning shock was set to be about 10 (●) or 25 (○)% above the threshold for producing responses in contralateral relaxed hand muscles. As in the subject in A, both test and conditioning stimuli were delivered through figure-of-eight coils. (From Ferbert et al, 1992, with permission.)

In order to investigate transcallosal connections between the two sides of the brain, two magnetic stimulating coils are used. One, (the conditioning coil) is used to activate, say, the right motor cortex. The other stimulator is used to test the excitability of the left motor cortex at different times after the conditioning shock. It turns out that a stimulus to the right side of the brain can inhibit responses evoked from stimulating the left motor cortex (or vice versa) at intervals of 6-7ms or longer. The EMG traces from the right first dorsal interosseous muscle in the top of figure 1 illustrate the effect in the raw data from one subject. The first trace shows a control response to the test stimulus (of the left motor cortex) given on its own. In the lower traces, the test stimulus has been preceded by a conditioning stimulus given at different times beforehand to the opposite side of the head. As can be seen, when the conditioning stimulus was given 6 or 7ms before the test stimulus, the test response was inhibited. There is quite a large change, and it is not necessary to average many trials to see the effect.

The lower graph is the time course of inhibition averaged from seven subjects. In this graph, the size of the control response given alone is set to be 100%. Values less than this indicate that the test response was inhibited by the conditioning shock. The x-axis indicates the interval between the conditioning and the test stimuli. Inhibition begins at about 5 or 6ms, and then it begins to recover by 15ms or so. The duration of inhibition depends upon the intensity of the conditioning stimulus. The higher the conditioning intensity, the longer the period of inhibition. The intensity of the test shock is also important. The larger the response to the test stimulus, the more difficult it is to observe inhibition. The inhibition is quite a specific effect. If the conditioning coil is moved away from the hand area of the cortex, either more medially or more laterally, then the amount of inhibition is reduced.

It is important to know at what level in the central nervous system (spinal cord or cortex) this inhibition is occurring. For example, the conditioning stimulus could be inhibiting spinal cord mechanisms so that a descending test volley produced a smaller EMG response in muscle because the spinal cord was less excitable than usual. On the other hand, the conditioning stimulus may have inhibited the cerebral cortex itself, so that a given test stimulus evoked a smaller descending volley from cortex to cord than in the control situation. We think that the latter is true, because the same conditioning stimulus has no effect on test responses produced by an anodal electric stimulus instead of a magnetic stimulus. This is illustrated in Figure 2. These experiments were carried out in active muscles, so that the intensity of the test shocks could be as small as possible. The upper EMG traces show examples from one subject of the effects. The lower graph shows the average effects from four different subjects. Responses to electric stimulation of the brain were not inhibited by a conditioning stimulus to the opposite hemisphere; responses to magnetic stimulation were suppressed considerably. From this result, we conclude [see arguments in refs. 2, 5 and 6] that the conditioning stimulus was having an inhibitory effect on the excitability of the contralateral motor cortex. One other piece of evidence supports this conclusion. H-reflexes in relaxed forearm muscles are unaffected by the conditioning stimulus even though the same stimulus can suppress responses evoked in those muscles by a magnetic stimulation of the cortex.

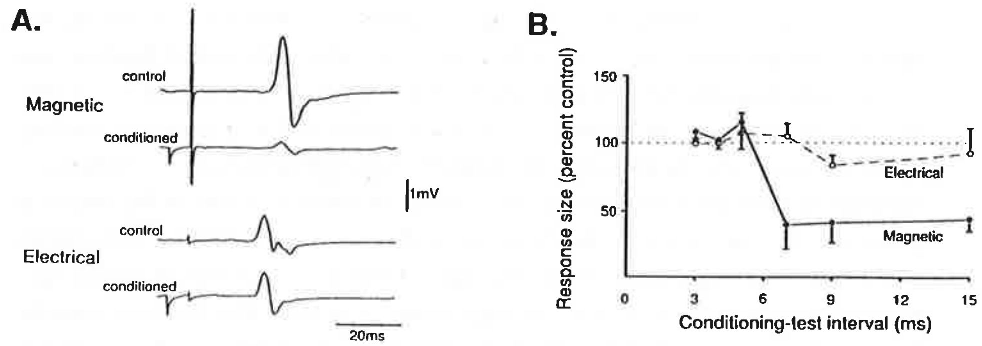

Figure 2
A, Example in a single subject of the effect of a magnetic conditioning stimulus over one hemisphere on the size of EMG responses evoked by magnetic (upper two traces) or electrical (lower two traces) stimulation of the opposite hemisphere. Each pair of traces shows the average (of ten trials) response to the test shock given alone (control) or with the response conditioned by a preceding magnetic shock (conditioned). The conditioning-test interval was 9ms in both pairs of traces. The response to magnetic test stimuli is inhibited, whereas the response to electrical test stimuli is almost unchanged, apart from loss of a second positive component which had been present in the control. The four experimental conditions were applied pseudo-randomly in one bock of trials. B, Comparison of the time course of interhemispheric inhibition from responses elicited by electrical or magnetic test stimuli in four subjects. Points are means ±SEM (paired t-test of comparing mean inhibition of electrically and magnetically evoked test responses showed significant ($P<0.05$) differences at intervals of 7, 9 and 15ms). Magnetic conditioning stimuli were given via a figure-of-eight coil having a maximum output rating of 2.4T. Stimulation intensity ranged from 50-67%. Magnetic test stimuli were given through a large round coil having a maximum output rating of 2T. Stimulation intensities ranged from 45-74%. The mean peak-to-peak size of the control EMG response evoked by magnetic stimuli (2.7mV) was slightly larger than control responses evoked by electrical test stimuli (2.2mV). (From Ferbert et al, 1992, with permission.)

The final piece of the jigsaw is to answer the question as to the pathway via which this inhibition is mediated. That is, how did the effect cross from one hemisphere to the other? Because of the time interval at which inhibition becomes noticeable, it is likely that the pathway involved the corpus callosum. There are very many fibres which link the two hemispheres of the brain across the corpus callosum, and we think that the inhibitory effect seen in these experiments was conducted via this pathway. Indeed, we have examined one patient with agenesis of the corpus callosum and in her the inhibition was absent [8].

It is not necessary to use a test magnetic stimulator to observe transcallosal inhibition. A conditioning stimulus to one hemisphere can also inhibit ongoing voluntary activity of the contralateral motor cortex, producing a silent period in the EMG ipsilateral to the conditioning hemisphere. This effect is particularly prominent in distal muscles, but is more difficult to observe in proximal muscles such as the biceps or deltoid [6, 9].

Having demonstrated the effect, and the pathway by which it is mediated, two questions remain about transcallosal inhibition. First, what is its normal function, and second, how does this go wrong in neurological disease? As an answer to the first question, it is possible to investigate how transmission in this transcallosal pathway changes during the performance of different types of movement. Preliminary experiments show that the amount of transcallosal inhibition is greater if the subject is performing a movement with the hand contralateral to the conditioning hemisphere, whilst keeping the other hand relaxed, as compared with a task in which the subject uses both hands at the same time [6]. In other words, it appears that the brain controls bimanual hand movements differently than unilateral hand movements. It may be that the transcallosal inhibitory pathway is useful in preventing movement of one hand when we use the other.

This transcallosal pathway also appears to go wrong in neurogical disease. Patients with myoclonus have muscle jerks which may occur in response to a sensory stimulus, or may occur when the patient tries to move. In some cases the jerks affect the whole body, so that, for example, a tap to the arm on one side will cause a massive jerk of both the arms and legs of the patient. In some of these patients, it appears that the somatosensory stimulus provokes an abnormal discharge on the contralateral side of the brain which then spreads via the corpus callosum to drive activity in the other hemisphere [10]. The net result is a bilateral twitch of the body. In this sort of patient, transcallosal inhibition seems to be absent. Indeed, in some cases the normal inhibition is replaced by a facilitation, consistent with the known pathophysiology of these patients.

A second example of the use of two magnetic stimulators to explore connections between different parts of the brain is the demonstration of possible cortico-cortical inhibitory connections between different parts of the motor cortex of the same hemisphere. The experiment is a variation on the transcallosal test described above. However, in this case, the conditioning coil is placed at the vertex of the skull in order to stimulate the leg area of the motor cortex. The test coil is again on the hand area of motor cortex. Figure 3 illustrates the experimental arrangement and some typical EMG records from a normal subject. In these experiments, the intensity of the test shock was set to be such that it evoked responses of about 1-2mV peak-to-peak amplitude in contralateral relaxed hand muscles. The intensity of the conditioning shock was set so that it would elicit responses in leg muscles, but is incapable of producing any responses in hand muscles, even if the subject activated the hand.

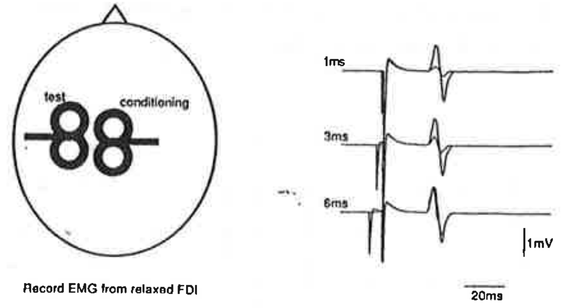

Figure 3
Probable cortico-cortical inhibition from a conditioning stimulus placed over the vertex of the skull to a test coil placed over the hand area of motor cortex. The right hand traces show averaged EMG records from the relaxed first dorsal interosseous muscle of the left hand of a typical normal subject. Each trace consists of two superimposed records: the response to the test stimulus given alone, and the response conditioned by the vertex stimulus at different times beforehand (numbers on left of each trace). The control response is the larger of each pair.

Under these circumstances, if the conditioning stimulus was given 1ms to 6ms before the test stimulus, the test response was inhibited. If the conditioning stimulus was moved away from the vertex, then this inhibition disappeared. In addition, we can demonstrate that the inhibition is probably occurring at a cortical level if we compare the effect of the vertex conditioning shock on responses evoked by anodal electric test stimuli to the lateral part of the motor cortex. As shown in Figure 4 responses to anodal test stimuli were unaffected by conditioning stimuli applied to the vertex, whilst the responses to magnetic stimulation were suppressed. In companion experiments, we verified that the conditioning stimulus also had no effect at these intervals on H-reflexes elicited in relaxed forearm flexor muscles.

As with transcallosal inhibition, a conditioning stimulus at the vertex may also produce suppression of ongoing voluntary EMG activity. Such suppression has been noted by many authors using very low intensity stimulation of the hand area itself [11]. Recently, Caramia et al [12] reported that vertex stimulation also was capable of suppressing EMG activity in active hand muscles without producing any direct motor response.

This type of inhibition is clearly quite different from the transcallosal inhibition described previously. It begins at very much shorter interstimulus intervals, and is elicited from a different position of the conditioning coil. It may be a type of "surround inhibition" of one area of cortex by another.

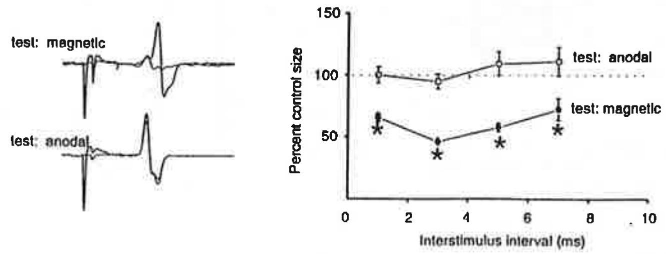

Figure 4
Comparison of the effect of a vertex conditioning shock on responses evoked by a magnetic or anodal electric test shock in the active first dorsal interosseous muscle of one subject. Traces on the left are raw data showing the effect at a conditioning-test interval of 2ms. Each trace consists of two superimposed records: the average (of 10) response to the test stimulus given alone and the response when conditioned by the vertex shock. In each trace, the control response is the larger of the two. Note that the response to the magnetic test shock is almost completely suppressed by the conditioning stimulus, whilst the response to anodal stimulation is virtually unaffected. The graph on the right shows the time course of this effect. Each point is the mean (±1SE) of 10 observations. Asterisks indicate a significant ($P<0.05$) difference in the effect of the conditioning stimulus or the two types of test response.

Again, the question is how does the circuitry go wrong? We have not yet had an opportunity to investigate this question yet in detail. However, one natural condition to examine is epilepsy, in which patients are thought to have abnormalities of both excitatory and inhibitory mechanisms within the brain. Recent data collected in this laboratory suggests that abnormalities of the inhibitory circuit may exist in certain groups of epileptic patients. However, disorders of this inhibitory circuit are not specific to epilepsy: abnormalities also are evident in basal ganglia disease, perhaps caused by changes in the tonic input to motor cortex from basal ganglia structures.

Acknowledgements

The work described in this article is the result of a collaboration with very many colleagues over several years: Drs Brian Day, Philip Thompson, Kaz Ugawa, Andreas Ferbert, Alberto Priori, Jim Colebatch, Peter Brown, Takashi Kujirai, Alain Maertens de Noordhout and Professors David Marsden and Pat Merton. Mr Richard Bedlington provided the firm bedrock of background support.

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PRESENCE OF NEUROPHYSIOLOGICAL SIGNS OF SPASTICITY WITH NORMAL CONDUCTION IN THE FASTEST FIBRES OF THE CORTICOSPINAL PATHWAY IN BABIES DEVELOPING CEREBRAL PALSY

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Normal conduction in the fastest fibres of the corticospinal tract has been observed in children with established cerebral palsy who have spastic quadriplegia (Eyre et al., 1989). An 18 month longitudinal study has now been performed in 44 babies (aged 31 weeks gestation to 12 months at recruitment), 24 at high risk and 20 at low risk for cerebral palsy, to determine if an abnormal pattern of development of corticospinal transmission might be associated with the onset of spasticity. Studies were performed at approximately 6 monthly intervals. Ethical approval and parental consent were obtained. Surface EMGs were recorded from biceps and triceps brachii, deltoid and pectoralis major muscles. Electromagnetic stimulation was used to estimate the central motor conduction delay to biceps motoneurones and the threshold of response (Eyre et al., 1989). The threshold of the phasic stretch reflex in biceps and its radiation to the other muscles studied were determined using a mechanical vibratory stimulus (Eyre et al., 1991). Neuro-developmental assessments were made at 18 months. The data were compared with cross-sectional data obtained from 200 normal children aged 32 weeks gestation to 2.5 years.
Of the 44 subjects, 3 subjects died with signs of spasticity (aged 3, 3 and 5 months). 11 subjects have spastic involvement of all four limbs and abnormal motor development at 18 months. 30 subjects have normal neuro-developmental assessment at 18 months. Corticospinal conduction in all subjects fell within the normal range. Abnormal development of the phasic stretch reflex with low thresholds and abnormal persistence of radiation were observed in all subjects of the spastic group and in two of those who died. The 30 normal subjects had normal development of the phasic stretch reflex.
Neurophysiological signs of spasticity in babies developing cerebral palsy therefore occur despite normal conduction delays in the fastest fibres of the corticospinal pathway.

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SILENT PERIOD ELICITED BY TRANSCRANIAL MAGNETIC STIMULATION

preliminary results in normal subjects
and hemiparetic stroke patients.

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The action potentials of a muscle voluntarily contracting muscle are transiently suppressed following a CMAP elicited by transcranial magnetic stimulation of motor cortex(TCCS).
The suppression of electrical activity is designated as the silent period(SP).
We verified in 22 normal subjects(13 female,9 male;mean age 26 years)the relation between SP duration elicited by TCCS and stimulation intensity.We used a Novametrix Magstim 200 (coil diameter 14cm)and a Dantec (coil diameter 10cm) magnetic stimulator, using intensity stimulation at 60%-80%-100% of stimulator power. EMG activity was recorded by means of surface electrodes from thenar muscles during voluntary contraction.
We observed:1)a prolongation of SP duration proportional to the stimulus intensity 2) with the same stimulus intensity a SP prolongation proportional to the coil diameter.
We also studied 10 hemiparetic stroke patients(4 female,6 male; mean age 45 years).We observed:1) in all patients presence of CMAP elicited by TCCS 2) in spastic patients absence or strong reduction of SP elicited by TCCS 3) presence of SP elicited by TCCS in patients without spasticity.
All patients presented SP elicited by electrical stimulation of median nerve at the wrist.
Our data suggest that SP elicited by magnetic TCCS is due to intracortical mechanisms.These preliminary data may prove useful in studying the role of intracortical mechanisms of voluntary activity inhibition in the pathogenesis of spasticity in hemiparetic stroke patients.

A STUDY OF PULSED MAGNETIC STIMULATION AND F WAVE IN PATIENTS WITH PARKINSON'S DISEASE

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Introduction

Central motor conduction times (CMCTs) obtained by magnetic stimulation, which reflect the states of the cerebral cortex, descending tracts and spinal anterior horn, are useful for evaluation of the function in the central motor pathways. CMCTs have the possibility of the fluctuation by various states of the central nervous system. F wave response may be changed by parkinsonism. In this report, we studied the fluctuation of the CMCTs and the F wave by parkinsonism in patients with Parkinson's disease.

Subjects and Methods

Twenty-two patients with PD who had obvious lateralities in parkinsonism of upper extremities were examined. Motor evoked potentials (MEPs) by transcranial pulsed magnetic stimulation applied to the head were recorded from bilateral abductor pollicis brevis (APB) and abductor digiti minimi (ADM) muscles. Magnetic stimulation was given to the head when patients gave slight voluntary contraction to the target muscle and did not give voluntary contraction. F responses from APB and ADM were recorded by electrical stimulation to median nerve and ulnar nerve at the wrist. Paired t-test was done between both sides.

Results

1) CMCTs in the dominant side of parkinsonism were significantly shorter than those in the non-dominant side without voluntary contraction. On the other hand, CMCTs during slight voluntary contraction had no significant difference between dominant and non-dominant side of parkinsonism. 2) The differences in CMCTs between slight voluntary contraction and no voluntary contraction were smaller in the dominant side than in the non-dominant side. 3) Amplitudes of MEPs tended to be higher in the dominant side than in the non-dominant side. 4) F wave amplitudes in the dominant side were higher than in the non-dominant side. 5) Cycle of tremor in the discharge of the surface electromyogram was usually changed by transcranial magnetic stimulation, but magnetic stimulation just after period of discharge of tremor did not change the cycle of tremor and produced MEPs with low amplitudes.

Conclusion

Parkinson's disease showed reduced CMCTs and high F wave amplitudes, which may reflect the increased excitability of the anterior horn in Parkinson's disease.

THE EFFECT OF TRANSCRANIAL MAGNETIC STIMULATION ON THE H REFLEX IN RELAXED AND PREINNERVATED MUSCLE

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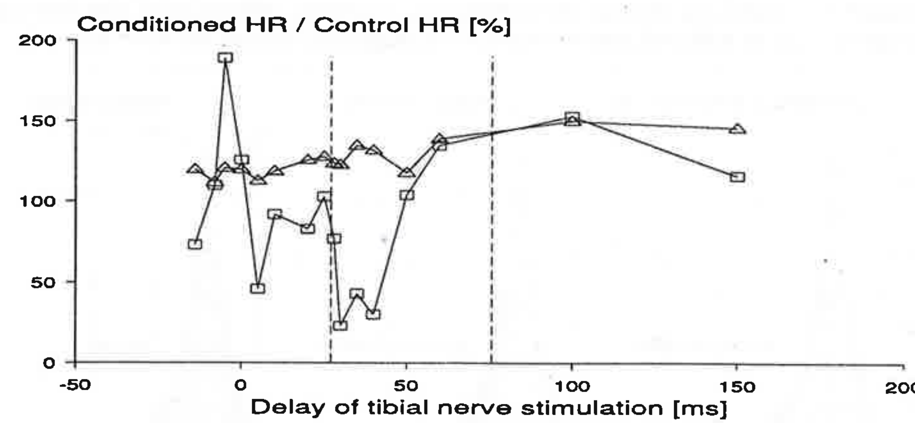
Spinal cord excitability following transcranial magnetic stimulation (TMS) was measured during either muscle relaxation or about 30% maximum force level using soleus H reflex (HR) testing in 10 healthy subjects. TMS at 5% above motor threshold was used as the conditioning stimulus and electrical tibial nerve stimulation at variable delays relative to TMS as the test stimuli. Nerve stimulation intensity was kept below motor fiber threshold and was adjusted to about 30% maximum HR amplitude. Soleus activity was monitored by surface EMG and by force transducers.

As a consistent finding in all subjects we observed a depression of alpha motoneuron excitability during voluntary soleus preactivation, starting on average at a conditioning to test stimulus interval of at least 22 msec (cf. squares in Fig.). The onset of HR depression coincided roughly with the mean onset of the postexcitatory inhibition (PI, 28 msec) occurring subsequent to the motor potential evoked by TMS. However, the duration of HR depression was significantly shorter than the duration of PI (40 vs. 57 msec), thus indicating that the later parts of PI cannot be accounted for by decreased spinal excitability, but are likely due to inhibitory phenomena originating upstream in the motor pathways. In the relaxed soleus muscle, TMS did not result in HR depression at any interstimulus time interval (cf. triangles in Fig.). We were able to demonstrate in an additional experiment that the degree of HR depression covaried with the level of voluntary soleus preinnervation. From this, we propose the activation of recurrent inhibition as a parsimonious explanation of these contrasting results (relaxed vs. preinnervated), since with increasing muscle contraction an increasing amount of motoneurons was activated synchronously by TMS, hence providing a more and more powerful input into the homonymous Renshaw cell pool.

We conclude that the early parts of PI are superimposed upon, or even due to segmental mechanisms (most likely recurrent inhibition), while the later parts are of supraspinal (probably cortical) origin.



Mean H reflex amplitudes of one representative subject, expressed as percentages of H reflex control level (without magnetic brain stimulation, 5 trials per condition), are plotted against the delay of tibial nerve stimulation relative to magnetic brain stimulation. Triangles: Relaxed soleus muscle; squares: Preinnervated at about 30% maximum force; vertical dashed lines: Onset and termination of the postexcitatory inhibition.

MAGNETIC AND ELECTRIC FIELDS PRODUCED BY REAL COILS IN REALISTICALLY SHAPED MODELS

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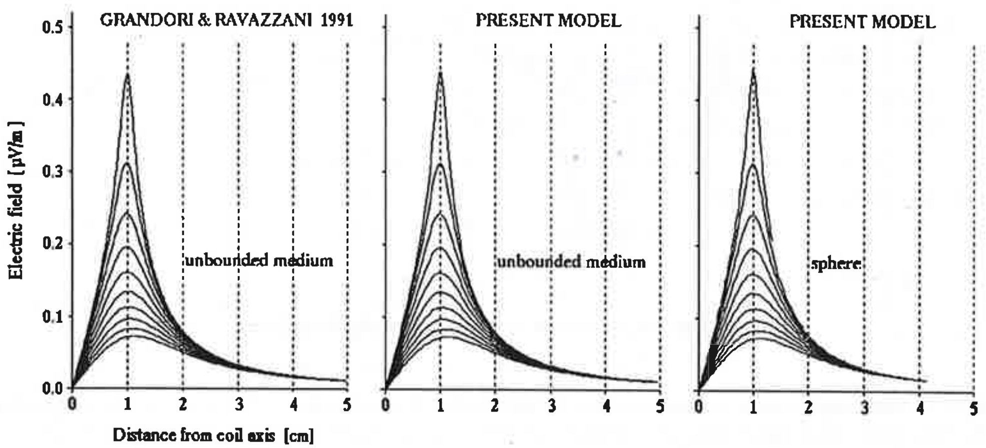
We propose a new model for the computation of the intracranial fields produced by magnetic stimulation. The model is based on the theory of the electromagnetic field (reciprocity theorem) and can be used also in bounded media of various conductivities (with spherical symmetries).
Present results will be first compared to those obtained from a much simpler approach proposed in a previous paper (Grandori and Ravazzani, *IEEE Trans. BME*, 1991, vol. 38, pp. 180-191). In our previous study, only magnetic fields *normal* to the outer surface of the conductor could be studied.
Input into the new model is via the time waveform of the current in the coil; the geometry of both the head tissues (scalp, skull and cerebral tissues) and of the coil are taken into account, to reproduce the influence of these variables upon the intracranial electric fields.
The overall structure of the model has been first validated in several test examples, with stimulating coils placed in typical locations.
The figure below shows an example of the magnitude of the electric field produced by the new model (unbounded medium and spherical homogeneous conductor, in the panel at the centre and at the right, respectively). Left: results obtained from the previous model. A circular coil (radius of 1 cm) was energized with an input current; the horizontal axis represents the distance from the coil axis. Data are computed along an axis parallel to the coil plane. For each panel, field magnitudes are given at different distances from the coil: each curve describes the field at distances from the coil plane ranging from 0.1 to 1 cm (in steps of 0.1 cm).
It will be shown how only the new model can be used to predict the "optimal" location and orientation of the coil(s) to maximise radial and tangential components of the intracranial field in a given cortical area.
Based on the relative predominance of the different components as a function of the stimulation parameters (coil geometry, location and orientation), some basic understanding about the excitation process can be achieved, also for multicoil montages and unusual coil locations.



EVALUATION OF CORTICOSPINAL PATHWAY INVOLVEMENT IN MULTIPLE SCLEROSIS USING MOTOR EVOKED POTENTIALS FROM UPPER AND LOWER EXTREMITIES

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Motor Evoked Potentials (MEPs) in 20 normal subjects and in 22 patients with definite and probable Multiple Sclerosis (MS), on the basis of Poser's Criteria, were investigated using Magnetic Stimulation of the brain. The aim was to analyze electrophysiological insight from lower compared to upper limb recordings.
Responses were recorded monolaterally on the left side from abductor digiti minimi (ADM) and tibialis anterior (TA) muscles, during facilitating contraction of the target muscle. Spinal supramaximal magnetic stimulation of motor roots over C5-C6 and D11-D12 intervertebral space was also performed.
We considered several parameters: i) Total Motor Conduction Time (TMCT); ii) Central Motor Conduction Time (CMCT), obtained from the TMCT by subtracting Peripheral Motor Conduction Time (PMCT), determined both directly (motor roots stimulation) and indirectly (F-wave); iii) the amplitude of MEPs, expressed as percentage of maximum peripheral M wave size.
From upper limbs MEPs in 3/20 (15%) were delayed; direct CMCT in 13/20 (65%) and indirect CMCT in 11/17 (64,7%) were prolonged. From lower limbs in 1 case MEPs were absent and in 12/22 (54,5%) delayed; direct CMCT in 13/18 (72,2%) and indirect CMCT (in 2 patients were absent) in 14/18 (77,7%) were outside normal upper limits. MEPs amplitude was abnormal in 14/17 (82,3%) from upper and in 15/18 (83,3%) from lower limbs. Responses showed a good correlation with involvement of the motor tract and have also been found abnormal in patients whose conventional EPs and MRI were normal.
In conclusion, these results show that the incidence of electrophysiological MEP abnormalities is higher recordings from lower extremities, probably because they explore longer pathways and may diagnose lower spinal cord lesions.

FACILITATION OF MAGNETIC EVOKED RESPONSES WITH CONTRA- AND IPSILATERAL ACTIVATION

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In most magnetic evoked respons studies preactivation of the target muscle has been used to facilitate the motor evoked response. An obvious advantages of contracting the *contralateral* muscle is the absence of EMG contamination of the baseline. In this study differences of central motor conduction obtained with slight activation of target muscle (ipsilateral:IL) and strong contraction of the contralateral (CL) muscle was studied in 18 controls. Responses were recorded at the abductor pollicis brevis (APB) and tibial anterior (TA).
Magnetic stimulation was done with a Cadwell MES-10 stimulator. The main results can be summarised as follows: Ipsilateral activation (as compared with contralateral activation) results in a faster respons to the hand muscle (mean: 1.7 ms), the amplitude increased only slightly. To the leg muscle a nearly twofold amplitude increase was found with IL activation; however, latency didn't change (figure 1A and B).
For both facilitation procedures a clear correlation was found between left and right (normalised) amplitude, both for APB 2) and TA; except for TA respons with IL activation (figure 2). This points to a different mechanism of enhancement for IL activation of TA muscle; probably a rise in excitability of spinal motoneurones is largely responsible for this increase in amplitude. Thus, the normalised amplitude predicts the amplitude on the other side and can be used as a criterium in case of a unilateral low respons.
In clinical practice contralateral activation as a method of facilitation seems appropriate in most cases: lack of EMG contamination of baseline makes reading of latency onset easier, in addition, normal values can be applied in case of severe hemiparesis. Only in case of a low TA respons IL activation can give better responses.

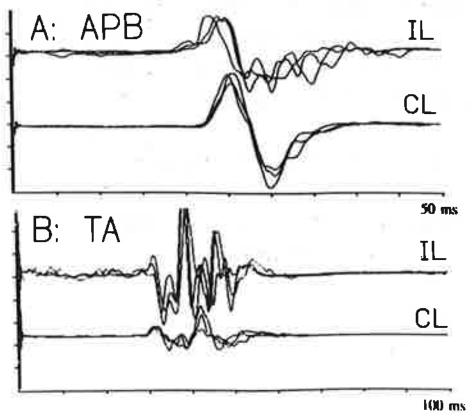

Figure 1

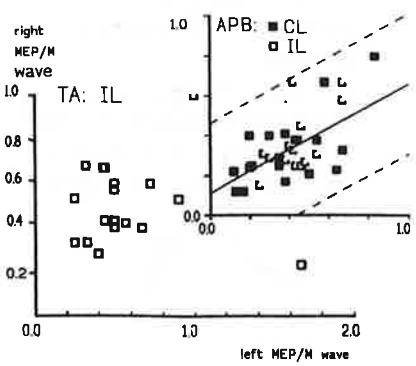

Figure 2

INVESTIGATION OF CENTRAL MOTOR PATHWAYS IN STROKE PATIENTS WITH ELECTROMAGNETIC STIMULATION AS A INDICATION OF OUTCOME

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In a longitudinal study central motor conduction time (CMCT) was measured to muscles of the upper limb in 118 first-ever stroke patients to determine the value of electromagnetic brain stimulation in predicting functional outcome at 12 months. Estimates of CMCT, obtained within 12-72 hours of symptoms, were shown to be predictive of outcome. Observations on CT scans performed on 107 of the same patients are compared in this paper with the CMCT values, mortality and functional assessments at 12 months, including motricity index for muscle strength, nine hole peg test for manual dexterity, Barthel score for activities of daily living and the modified Rankin scale for functional outcome.

The absence of recordable responses following electromagnetic brain stimulation correlated with poor outcome at 12 months and increased mortality. Patients with cerebral infarction were the most likely to demonstrate abnormal CMCT and showed the least functional recovery. CT scans were less reliable than measures of CMCT in predicting outcome.

Clinical assessments, CMCT and CT scans all have some predictive value. The main advantage of CMCT is that the measurements can be performed at the bedside within hours after stroke to provide prognostic information regarding mortality and recovery.

CLINICAL USEFULNESS OF MOTOR EVOKED POTENTIALS BY TRANSCRANIAL MAGNETIC STIMULATION

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Transcranial magnetic stimulation is now being employed in examination of the integrity of descending motor pathways. In clinical application to patients having pathology in the spinal cord, it is sometimes difficult to evoke the response from the muscle involved. Therefore, the target muscle contraction which is well known to facilitate the Motor Evoked Potentials (MEPs) has been used widely. However, the latency of the MEPs is reduced remarkably in this maneuver.
The purpose of this study is to elucidate the usefulness of the remote muscle contraction for normal persons and patients by recording the MEPs from the upper and lower extremities. We have examined the motor evoked potentials (MEPs) in 35 normal subjects and 126 patients with spinal cord disorders leading quadriparesis or paraparesis. The subjects rested comfortably in a sitting position and the stimulator coil was placed at the vertex and then the MEPs were evoked by transcranial motor cortex stimulation (Cadwell MES-10) during relaxed position and the voluntary contraction of the target muscle. MEPs of the Tibialis anterior, lateral Gastrocnemius and thenar muscle were recorded with the conventional surface electrode while the subjects assumed the target muscle and/or the remote muscle contraction. Special consideration was focused on the silent period after the MEPs during the maximum voluntary contraction of the target muscle.
The results of the 20 normal subjects showed that remote muscle contraction enhanced the amplitude of MEPs in both side of the leg muscles. However, the augmentation in the upper extremity was not remarkable. The presence or absence of MEPs in muscles below the zone of injury largely corresponds to the clinical and other investigative features of the paralysis. Frequently MEPs could only be identified during reinforcement maneuvers. During the voluntary contraction of the target muscle, the consistent silent period was observed after the MEPs. The duration of the silent period was changed by stimulating intensity and location of the coil. Furthermore, in spastic patients, this duration was remarkably shortened and the silent period was not recognized in severe case, especially in thenar muscle.

THE ANALYSIS OF SINGLE MOTOR UNIT POTENTIALS BY CORTICAL PULSED MAGNETIC STIMULATION

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Using the newly developed twin coil magnetic stimulator limited areas of cortex or spinal roots can be stimulated in man. In the analysis of motor evoked potential (MEP), the sites of stimulation by cortical magnetic stimulation have been debated. Day et al reported that a few interneurons connected to the pyramidal cells were evoked first by cortical magnetic stimulation. In this study, a patient with few motor units in small hand muscles was studied to analyse MEPs activted by magnetic stimulation with twin coil. An M wave with the amplitude of 204 μV was recorded from the right APB by C_8 root magnetic stimulation and an M wave with the amplitude of 106 μV was recorded from the right APB by T_1 root magnetic stimulation. The latencies of these M waves were both 18.8 ms. The shape and amplitude of both of these M waves were constant in spite of changing the strength of the stimulation.

An M wave with the amplitude of 560 μV and latency of 20.6 ms was recorded from the left ADM by C_8 root stimulation. An M wave and F wave were recorded from the left ADM by left ulnar nerve electrical stimulation at wrist and they were the similar amplitudes of 560 μV. MEP recordings from the right APB by left cortex stiumulation showed that the largest evoked area was on the line 1 cm in front of Cz - ear line, where the MEP was evoked in 1cm, 2cm, 3cm, 4cm, 5cm and 6cm latency from the middle. The shapes of these 3 MEPs were similar and the latency of the MEP by stimulation of the site of 2cm was delayed 0.7 ms compared with the latency stimulated at the two other areas. MEPs with two negative peak were recorded. The shape of MEP belonging to the shorter peak of the two negative peaks, was almost similar to the shape of the MEP evoked by the other area of skull, and the shape of this MEP was almost the same as the M wave evoked by C_8 root stimulation. The shape of the MEP belonging to the longer peak of the two negative peaks was very similar to the M wave evoked by T_1 root stimulation.

The mapping of CMCT-mag showed that the shortest CMCT was 6.8 ms recorded from several points of stimulation, the next shorter CMCT was 7.2 ms from only one point, the next shorter CMCT was 7.5 ms from several points, the next was 8.2 ms from several points, the next was 9.0 ms from several points, and the longest one was 9.7 ms from one point. The difference between each of these was almost constant at 0.7 - 0.8 ms and its multiples except for one point. MEP recordings from the left ADM by right cortex stimulation, showed that MEPs with an amplitude of 550 μV were recorded by stimulation of 4cm, 5cm and 6cm lateral from midline on the Cz - ear horizontal line. The MEPs with the same amplitude were recorded by stimulation of 5cm and 6cm lateral from midline 2cm in front of the Cz - ear horizontal line. The latencies of the MEP were 28.5 ms and 30.1 ms. The CMCTs were 7.9 ms and 9.5 ms. The difference of these CMCT was 1.6 ms.

By mapping of MEPs as the site of cortical stimulation was changed, the latency of the MEPs increased by 0.7-0.8 ms suggesting one synaptic delay. It is suggested that magnetic stimulation of human cortex indirectly activate pyramidal cells via interneurons.

DOES THE TRANSCRANIAL MAGNETIC PULSE STIMULATION AFFECT THE CARDIAC PACEMAKER FUNCTION?

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1. INTRODUCTION

It is said that the transcranial magnetic pulse stimulation (TMPS) should be avoided in the case of patients with epilepsy, severe psychiatric disorders, and wearing metals of electromagnetic devices[1,2]. Some patients who need the examination of neurological functions have severe hemo-dynamic problems, and with a cardiac pacemaker (CPM). We tried to evaluate effects of TMPS on the function of CPM.

2. MATERIALS AND METHODS

Three Mongrel dogs of about 10kgBW were used, Multiprogrammable pulse generator CPM unit (Siemens-Elema AB, Sweden: DIALOG 728) was implanted underneath the chest skin of dog, the tip of catheter electrode was fixed inside the right ventricle via the external jugular vein, and ECG and BP were monitored during the examination. The magnetic pulse stimulator with 800V of 1.5mF (Nihon Kohden, Japan), stimulation coil of 85mmϕ, and evoked potential monitor were used. Acetylcholine of 10mg/kgBW was injected, i. v., for fully effective pacing.

3. RESULTS

The output of multiprogrammable pulse generator CPM unit was suppressed by exposing to TMPS when it was in demand mode and TMPS was applied within the CPM sensitive period of intracardiogram R wave. When CPM unit was in fixed-rate mode TMPS made stimulus artifact on ECG in any time without effective cardiac output.

4. DISCUSSION

Electromagnetic interference from TMPS on CPM was evaluated in this study. Change of mode and depression of CPM output could be occurred by TMPS. When TMPS is applied to patients with CPM, it is necessary to have a special attention, including the changing mode of CPM. Synchronized (triggered type) TMPS should be developed for the case of patients with CPM, and also for cases with severe hemo-dynamic problems.

5. ACKNOWLEDGEMENT

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CNS CHANGE AFTER OPERATION OF NERVE SUTURE BETWEEN INTERCOSTAL NERVE AND MUSCULOCUTANEOUS NERVE IN BRACHIAL PLEXUS INJURY

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From the study of amputation, the organizational patterns observed in the cereberal cortex were not static, but could change under certain condition.

To evaluate motor reorganization in motor pathways after nerve suture between intercostal nerve and musculocutaneous nerve following the complete brachial plexus palsy, we studied motor evoked potentials (MEP) by transcranial magnetic stimulation.

In complete brachial plexus palsy, the operation of nerve suture with intercostal nerve to musculocutaneous nerve is successful in restoring the function of biceps brachii muscle (BBM). After operation the biofeedback technique was used for motor reeducation. The mechanism of functional changes in the central nervous system (CNS) was not clear. We analysed the CNS change during recovery of motor function of BBM neurophysiologically. Surface EMG of BBM and intercostal muscle (ICM), nose and chest respiration sensor were used for recording during respiration and during magnetic stimulation by using the twin coil. The subjects were all male, mean age 18.5 years, with complete brachial plexus palsy. T_4 root stimulation evoked motor potential (MEP) in BBM of operated side (op-BBM). Initially during deep respiration the muscle discharges in op-BBM were observed. Muscle discharges during voluntary movement were connected to the respiration more in initial stage, and gradually this connection was decreased. The MEP latency of op-BBM by cortical magnetic stimulation became shorter gradually. The cortex area, where MEP in op-BBM were evoked, moved laterally gradually.

This study suggested that the motor reorganization in motor cortex and pathways appeared after nerve suture between intercostal nerve and musculocutaneous nerve following the complete brachial plexus palsy.

SOME EFFECTS OF THE TRANSCRANIAL ELECTRICAL STIMULATION ON THE ALPHA ACTIVITY

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The effects of transcranial electrical stimulation on the brain electrical activity were studied in healthy subjects with dominating alpha activity in EEG. Five electrical stimuli with intensity sufficient to elicite motor action potential at m. abd. poll. br. (anode over the motor area, palathal cathode) were delivered in two series with frequencies of 2Hz and 10 Hz. Topographical analysis of the following EEG parameters defined for linear power density spectra (band 8-13 Hz) was made: 1. peak frequency; 2. mean frequency; 3. power; 4. percentage power; 5. mobility; 6. complexity. The prestimulus 1 minute epoch was compared with T-test to each of the 5 successive minutes after the stimulations.

From the results the following conclusions can be drown out: 1. Most of the parameters differed significantly before and after the stimulations. The prominent effects of the stimulations were in the mean frequency and mobility (up to 10 SD) in all trials. 2. The changes persisted in the analysed time interval of 5 min after each stimulation. Some of them attenuated with the time, but for other there was a tendency for increase. 3. The poststimulus changes were related to the background characteristics of the alpha activity. 4. From the two stimulations more prominent and enduring effects were achieved with higher frequency pattern.

In conclusion, the shortlasting and localized transcranial electrical stimulation affects definite parameters of the EEG. Some of the effects are quite distant from the stimulated area, which could be explained with cortico-cortical mechanisms or with the engagement of deaper structures.

DIFFERENT EFFECTS OF TRANSCRANIAL MAGNETIC STIMULATION ON SIMPLE AND CHOICE REACTION-TIME TASKS

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It has been shown that an appropriately timed transcranial cortical stimulation (TCS), besides evoking contralateral motor responses, can delay the onset of a voluntary movement *(Day et al., 1989)*.

In a group of normal right-handed volunteers we investigated the effects of magnetic TCS (Novametrix Magstim 200: 9 cm. diameter coil positioned flat on the vertex with clockwise current flow, threshold and supra-threshold output intensity) on the performance of a "simple" *(Tartaglione et al., 1986)* and "choice" *(Tartaglione et al., 1991)* visuo-motor reaction time task (releasing a reaction key and pressing a response key with the right hand). Changes of reaction time (RT) and movement time (MT), induced by randomly distributed cortical stimuli, were measured.

Magnetic TCS, preceding the motor task execution by approximately 100 msec., significantly increased the "simple" RT at supra-threshold intensity. TCS did not delay the RT in the "choice" paradigm, whilst induced a marked increase of MT, at both threshold and supra-threshold levels. The discrepancies in the two paradigms cannot be attributed to the process of choice, since similar results were observed comparing "multiple-choice" and "single-choice" tasks.

This study confirms that magnetic TCS can interfere with the execution of voluntary movements by inducing an inhibitory effect. Such effect may be apparently exerted on different parts of the motor programme, in relation to the complexity of the motor task, being more evident on more complex movements.

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SOME NEUROPHYSIOLOGICAL ASPECTS OF MUSCLE CRAMPS

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INTRODUCTION

Muscular cramps, clinically defined as painful shortening of the muscle (1), are supposed to take origin in the peripheral nervous system (2) even though some event that characterize them is still debated (3, 4, 5, 6). In a previous report we studied the effect of muscle stretching stopping cramps and it seems to be unrelated to the spinal monosynaptic or oligosynaptic connections described on basic features of the stretch reflex. Using the experimental model proposed by Lambert in 1969 (7), we verified that muscular cramps are elicitable by a high frequency electrical stimulation of the peripheral nerve. We found that stretching is able to break the cramp even if the connection between muscle and spinal cord is functionally interrupted by the anaesthetic block of the peripheral nerve. Muscular cramps are still evoking even after anaesthetic block of the nerve and stretching is still stopping them suggesting the influence of local factors interfering.

Cramps are commonly found in paraphysiologic conditions as pregnancy and in pathologic conditions as renal dyalisis and water intoxication (6) suggesting that local aedema can be an important factor influencing cramps presentation. Moreover it seems that the ordinary cramps can be more frequent in the cold water for the swimmers.

In the present work we studied an experimental model in order to evoke muscular cramps by giving trains of electrical pulses using several stimulation frequencies. Furthermore we used this model in order to analyze the influence of changes in local temperature and local aedema on cramps. On the same time we monitored the muscular contraction time by recording the myogram.

SUBJECT AND METHODS

5 voluntary healthy subjects, 4 males and 1 females, aged between 35 and 63 years (mean 44.2+/-12.7), were studied with the approval of the Ethical Committee.

Cramps experimentally induced were evoked, according to Lambert (7), by stimulating the tibialis posterior nerve at the ankle with trains of electrical pulses. Stimulus frequencies were ranging from 1 to 50 Hz. The initial stimulation frequency employed was 1 Hz (100 stimuli). Subsequently it was progressively increased to achieve the minimal frequency capable of inducing a cramp using no more than 40 stimuli. This frequency value was called "threshold frequency". Recordings were performed from flexor hallucis brevis muscle by surface electrodes, belly-tendon displacement, using B.A.S.I.S. Myograph.

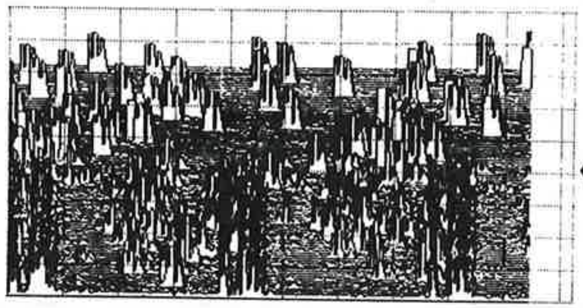

FIG. 1
"Threshold frequency" determination in one subject studied.
At high stimulus frequency an involuntary activity fallowing. M. and F waves appeared, persisting after the stop of the stimulation.

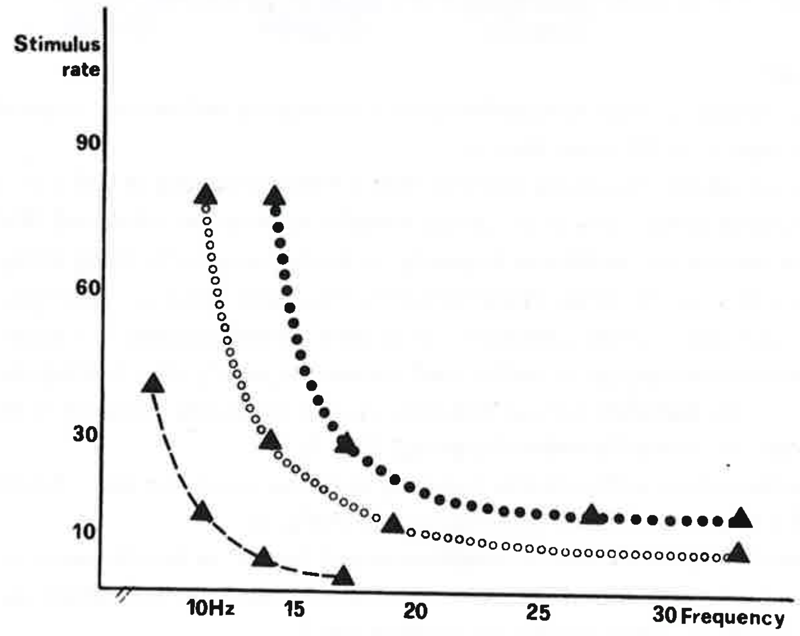

FIG. 2
Inverse correlation between stimulus frequency and duration of the train when the stimulus frequency is higher than "threshold frequency" in the 3 subjects studies.

Minimal stimulus frequency able to induce a true cramp, defined as self maintaining involuntary muscular activity after the end of stimulation with a clear shortening of the muscle and consequent fixed flexion of the hallucis, was called "threshold frequency".
All experiments, usually, were performed in a warm place and the temperature of the skin overhanging studied muscles was monitored at different steps of the experimental sessions with an electronic thermometer.
An infrared lamp placed close to the foot was employed heating the skin up to 38 degrees temperature, ice water was used cooling it up to 20 degrees.
The angular displacement of the big toe induced by the muscle twich, indirectly representing the length (myogram) of the flexor hallucis brevis, was recorded by using a position transducer connected to the first falanx of the big toe. The transducer was connected to the preamplifier of the electromyograph and the twich was induced by stimulating the tibialis posterior nerve at the ankle. The shortening time ratio was expressed dividing the amplitude by the duration of the displacement. The absolute length and duration of the rising phase of the displacement were also measured. Vertical calibration was 15 (0.5 mV).
Venous obstruction was obtained by placing a cuff of a sfingomanometer at calf level and setting it for a pressure higher than the minimum and lower than the maximum blood pressure for different time intervals.
T-Student test for paired data was employed for statistical analysis.

3. Results

All experiments, usually, were performed in a warm place and the skin temperature of subjects ranged from 36 to 36.5 degrees.
The initial stimulus frequency was 1 Hz with a maximal duration of 100 s for each train. The stimulation was given to the tibialis posterior nerve at the ankle level. With the progressive increase in the stimulus frequency, an involuntary activity, rising during the silent period was recorded. It was characterized by bi-triphasic spikes of motor unit potentials and disappeared when the stimulation was stopped. Further increases in stimulus frequency triggered true cramp represented by a self-maintaining activity after interruption of the stimulation. The minimum stimulus frequency capable of inducing cramp with a maximum of 40 stimuli was defined "threshold frequency" (Fig. 1).
On further increasing the stimulus frequency an inverse correlation was found between the duration of the train and the frequency employed (Fig. 2).
"Threshold frequencies" at basal conditions ranged from 8.3 to 16.6 Hz (mean 11+/-3.6 SD) and were not significantly different from "threshold frequencies" determined one week before ($p > 0.5$). Individual findings are shown in Tab. 1.

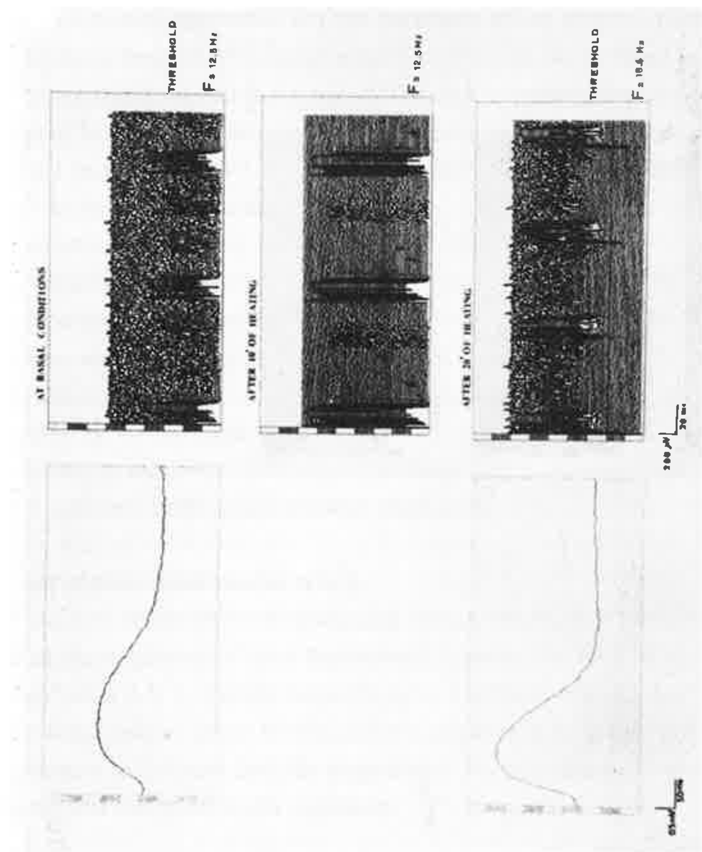

FIG. 3
Myogram (on the left of the picture) and "threshold frequency" heating the foot. At basal conditions "threshold frequency" was 12.5. Heating the foot it changed in 14 Hz.

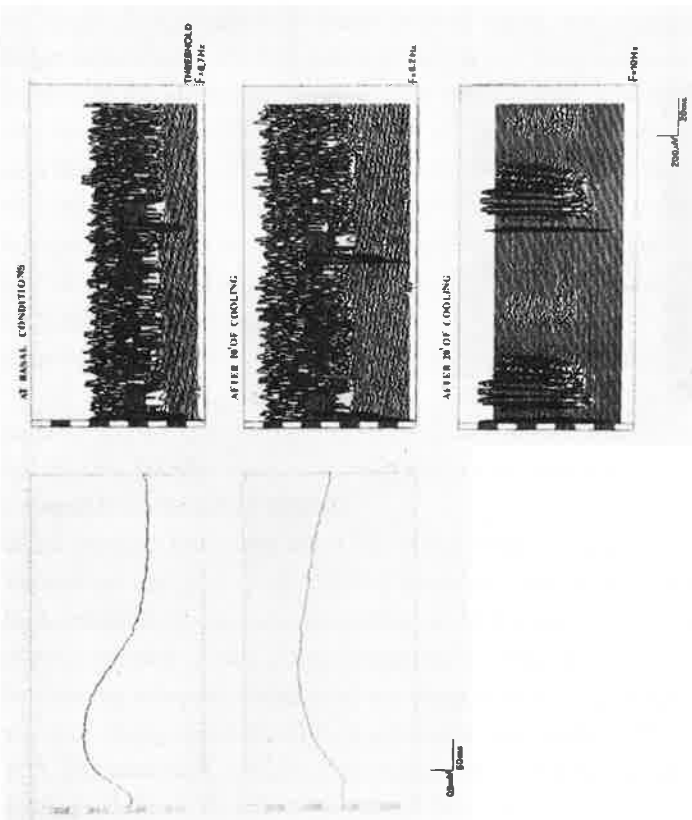

FIG. 4
Myogram (on the left of the picture) and "threshold frequency" cooling the foot.

3.1 Effect of temperature on muscle cramp

"Threshold frequencies" were assessed after cooling the muscle in three subjects during the same experimental session. With decreased skin temperature, "threshold frequencies" were slightly different compared to the values determined in basal conditions (Fig. 3). Individual findings are reported in Table 2. The myograms on the contrary, recorded in each condition, were slightly different compared to values determined in basal conditions.

In one subject the foot was kept in ice water for 10 min. The excitability of the nerve was reduced to such an extent that the M wave changed in amplitude and duration (Fig. 3). At that point it proved impossible to elicit the cramp at any stimulus frequency.

The "threshold frequency" in basal conditions was also determined later in the same experimental session in the same 5 subjects. The foot was then heated until the skin temperature reached 36,5C, after which the "threshold frequency" was again determined (Fig. 4). The individual results are given in Table 3. "Threshold frequencies" compared for paired data showed no statistical differences ($p > 0.5$).

Mean duration of the twich rising fase was at 37,5 104+/-9.2 ms.

Rising temperature significantly decreased twich duration of 83% ($p < 0.01$) while decreasing temperature significantly increased the twich duration of 162%.

The mean angular displacement in the basal condition was 34.2+/-6.4 degrees and rising temperature produced an significant increase of 184% ($p < 0.01$) while decreasing temperature produced 96% angular displacement.

The shortening ratio was 220% when the temperature increased and 60% when the temperature decreased. Both variations were significant.

3.2 Effect of obstructed venous return

For evaluation of the effect of obstructed venous return, the "threshold frequency" was assessed at the beginning of each experimental session for each subject. The "threshold frequencies" were 6.5, 7, 16.6 Hz respectively in 3 subjects examined at the beginning of each experimental session. After 15 minutes of obstructed venous return the "threshold frequencies" changed in 7, 7 and 16.6 Hz respectively. No significant differences were observed in the myograms recorded in this condition.

4. Discussion

The possibility of evoking muscular cramps by stimulating the peripheral nerve emerged as a result of Lambert's work (1). The aim of this study was to reproduce a reliable experimental model of muscular cramps which would be useful for testing the effective interference of numerous factors. We found a stimulus frequency, which though differing in the individual subjects, was reproducible enough to evoke muscular cramps in the target muscle.

On further increasing the stimulus frequency, an inverse correlation was found in each subject examined between duration of the pulse train and the frequency with which the stimuli were delivered. In other words, stimulation frequency appears to play a decisive role in the genesis of muscular cramps.
On the basis of these characteristics, the model we have elaborated may make it possible to test the effective role of numerous factors in the phenomenology of muscular cramps. To date, in fact, muscular cramps have proved largely difficult to control either by the subject himself or by the examining physician, making any systematic physiological study of such phenomena very unlikely.
In the preliminary testing of this experimental model, we considered a number of local factors which in clinica practice appear to play a certain role in causing muscular cramps. Support for our hypothesis that local factors may be crucial in the modulation of muscular cramps is provided by numerous observation. Cramps are relieved by muscular stretching (10) even after peripheral nerve block, excluding the interference of spinal loops (7). Their origin within the peripheral nervous system was one of the clearcut findings reported by Lambert studies (1), who demonstrated, on the one hand, the impossibility of evoking cramps in curarized muscles and, on the other, the possibility of evoking cramps by stimulating the peripheral nerve distal to a nerve block, thus suggesting a possible distal origin of electrical activity producing cramps. Moreover the fasciculations, which often accompany cramps, still persist after nerve block and nerve section (11, 12). Cramps are common features in paraphysiologic conditions as pregnancy and pathologic conditions as renal dyalisis and water intoxication (8). In the light of this, local mechanical factors may interfere with the peripheral nerve generating spontaneous activity. To analyze the mechanical changes in neuromuscular apparatus we chose to record the myogram in the various different conditions. While the myogram was clearly affected by different temperature and imbibition conditions, testifying to changes in mechanical features, the "threshold frequency" showed no changes.
The stimulus frequency capable of inducing cramp is not meaningfully affected by muscle temperature changes or by changes in tissue imbibition. On the contrary, the myogram is clearly modified by these factors.
The results rule out the possibility that local factors considered in our study may be the main factors bringing about changes in excitability of the peripheral nerve responsible for muscular cramps in physiological or paraphysiological states.

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TAB 1

SUBJECT	YRS	THRESHOLD (Hz)	THRESHOLD (Hz)
1	34	8.3	12.5
2	63	7.6	10
3	52	12.5	8.3
4	37	16.6	16.6
5	35	10	10

TAB 2

SUBJECT	THRESHOLD	THRESHOLD AFTER EATING
1	8.3	10
2	7.6	7.6
3	12.5	10
4	16.6	14
5	10	10

TAB 3

SUBJECT	THRESHOLD	THRESHOLD AFTER COOLING
1	8.3	6.2
2	7.6	10
3	12.5	8.3
4	16.6	16.6
5	10	8.3

THE EMG RESPONSES OF POORLY COORDINATED AND NORMALLY COORDINATED CHILDREN DURING AN ISOKINETIC CONCENTRIC KNEE EXTENSION TASK

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Research related to the neuromuscular processing in children has received little attention with even less attention focusing on the neuromuscular processes of the poorly coordinated (PC) child. It has been established that the PC child displays an inefficient movement pattern however, the causes for such inefficiencies are unknown.

The aim of this study was to identify the effect of coordination level and movement speed on the neuromuscular responses associated with the production of a concentric quadricep knee extension torque. Twenty subjects were classified as poorly coordinated (PC) or normally coordinated (NC) according to results obtained from the McCarron Assessment of Neuromuscular Development (McCarron, 1982). The mean age of subjects was 6 years 11 months and 7 years 2 months for the PC and NC subjects respectively. Subjects were required to perform a simple knee extension-flexion task thus reducing possible confounding influences associated with skill deficiencies. The task was performed on a Biodex dynamometer at isokinetic speeds of 120, 165 and 210 deg/sec. Reciprocal muscular actions were concentric with subjects performing three individual trials at each speed. EMG recordings were obtained from vastus lateralis (VL), vastus medialis (VM) and biceps femoris (BF) of the right lower limb using disposable paediatric Ag/AgCl surface electrodes, bandpass filtered at 3Hz and 1kHz, and stored on computer at a 1kHz sampling rate.

It was hypothesized that PC subjects would display greater co-contraction between VL and BF as represented by the IEMG ratio of VL:BF during the isokinetic period. It was expected that this ratio would be lower in the PC subjects at all movement speeds, with the amount of co-contraction increasing with speed. As a result of this co-contraction, the maximal rate of torque development and the peak torque attained by the PC subjects would also be lower. Variability in a movement pattern has been previously associated with less skilled performances. It was expected that PC subjects would show greater variability in the amount of co-contraction than NC subjects across the three trials.

Preliminary results ($n=12$) indicate that the PC subjects have greater levels of co-contraction than NC subjects at all movement speeds during the isokinetic period ($p < 0.05$). Variability in the neuromuscular pattern was expressed using the coefficient of variation for the VL:BF IEMG ratio. No significant difference was found in the variability of the two groups at any speed ($p > 0.05$). Despite the increased level of co-contraction in the PC subjects, rate of torque development showed no significant difference between the two coordination levels ($p > 0.05$).

The increased level of cocontraction observed in PC subjects may represent a neuromuscular response to increase the stability of the knee joint or it may suggest an inability of the PC subjects to efficiently organize the timing of their neuromuscular patterns. Inappropriate timing of muscle actions may be responsible for decreased peak torques and an inability to maintain the movement speed over an extended period of time; however, instantaneous measures such as maximal rate of torque development appear to be less affected.

STABILITY OF MULTICHANNEL SURFACE EMG RECORDINGS FOR STUDIES OF MOTOR CONTROL IN UPPER MOTOR NEURON DYSFUNCTION

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Under normal conditions, the brain controls movement through multiple connections to spinal motor centers which activate the muscles needed to perform the movement. However, following spinal cord injury (SCI), this connection of upper motor neuron to the spinal motor neurons is damaged in a non-systematic fashion, with consequent alteration of motor control. The brain control of motor function under such conditions must be assessed in more detail than provided by clinical evaluation in order to select the proper therapeutic intervention and to monitor its effects. Previous studies of abnormal motor control and spasticity have focused on the behavior of only one or two muscle groups in one task or paradigm. Multichannel electromyographic recordings of motor unit activity from muscle groups innervated by multiple spinal segments during the execution of various maneuvers can demonstrate the subject's remaining brain influence on spinal motor centers and provide a more comprehensive, objective and quantifiable measure of motor control. To the clinical observer, voluntary and spastic motor activity is thought to be variable to such an extent that objective measures would not be useful. We therefore sought to document the repeatability of such measures in a group of clinically stable SCI subjects.

We chose a group of 32 SCI subjects, 8 female and 24 male, who were without anti-spastic medications or active infections. From each subject we selected two studies at intervals ranging from 6 days to 5.5 years (total of 64 studies). The subjects ranged in age from 8 to 60, and were from 7 months to 33 years post-injury, with injury levels between C1 and T11. All studies were conducted with the patient in a comfortable supine position, and were carried out in the same sequence, usually at the same time of day (morning or afternoon). Surface EMG recordings were made of motor unit activity from quadriceps, adductors, hamstrings, tibialis anterior and triceps surae muscle groups bilaterally. Voluntary and passive single and multijoint movements of the lower limbs and vibration of patellar and achilles tendons were performed. A manual scoring system was employed to assess the pattern and amount of EMG activity recorded during voluntary and passive movements (0-3), and during application of vibration (0-4).

From the group of 32 subjects, 8 subjects showed identical results for all 10 measures on repetition. Sixteen subjects had identical scores for voluntary maneuvers, 15 for passive, and 15 for vibration. The overall reliability index was 0.92, and all measures in repeated studies were significantly correlated ($p < 0.01$), with the exception of response to vibration of the right triceps. The highest correlations were seen in the voluntary maneuvers (0.85 - 0.93). Some of the variability in the response to vibration may be explained by the fact that vibration does not elicit a response in all healthy subjects. Other variability could be understood in terms of an overall shift in segmental excitability, resulting in alteration in the amount of response, including appearance or disappearance, as evident by examination of the overall response - pattern across all maneuvers. The use of this technique to track spontaneous or induced changes in motor control will be discussed.

RECIPROCALE REFLEX ACTIVITY OF MUSCLE TONE ABNORMALITY STUDIED BY ELECTROMYOGRAPHY TOGETHER WITH TONIC VIBRATION REFLEX

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The reciprocal reflex activity in various muscle tone abnormalities was analysed using the H-reflex together with tonic vibration reflex (TVR). The H-reflex was elicited by short interval (5-8 msec.) double volleys while vibratory stimulation (TVR) of 120Hz. was applied to the synergist (gastrocnemius) or antagonist (peroneus) muscles. On the other hand, the frequency depression (FD) curve of the H-reflex was plotted at a higher frequency (1-50Hz.)stimulation.

The patients analysed in this study totaled 50, consisting of 8 normals, 1 with peroneal nerve damage, 22 with spasticity, 6 with rigidity 9 with cerebellar-brainstem lesion, and 4 with spinal cord lesion (upper and lower spinal cord).

In the normal subjects, two successive H-reflexes (H_1 and H_2) were elicited by short interval double volleys, and attenuated by TVR applied to the gastrocnemius as well as the peroneus muscle. The attenustion rates were 68% in gastrocnemius TVR and 80% in peroneus TVR. The FD-curve showed a medium level in lower frequencies and then rose gently higher frequencies. In the peroneal nerve damage patient, although the H_1 and H_2 were attenuated by the gastrocnemius TVR they were not attenuated by the peroneus TVR because of a loss of the reciprocal innervation. The FD-curve showed a normal characteristic configuration because there were no lesion in the upper center. In spasticity, the H-reflexes were not attenuated by gastrocnemius nor by peroneus TVR's at a sufficient rate. The rates were 38% in the former and 46%in the latter. The FD-curve showed a higher level at a lower frequency then showed a steep sloping configuration from medium to higher frequencies. In rigidity, the rates were 26% in gastrocnemius TVR and 51% in peroneus TVR. The FD-curve showed a higher level at a lower frequency, then showed a slighter sloping from medium to higher frequencies. In cerebellar-brainstem lesion, the rates were 25% in gastrocnemius TVR and 49% in peroneus TVR. The FD-curve showed a higher level in all frequencies then showed a gentle sloping configuration at a higher level from medium to higher frequencies. These findings could be considered as a result of the disturbance of the reflex modulation from the upper center.

The combined examination of the H-reflex and TVR seems to be useful diagnostic method and may also help in the determination of a therapeutic approach that is most appropriate.

MOTOR UNITS RECRUITMENT AND FIRING IN PERIPHERAL NEUROPATHIES

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Even if in normal subjects there is a strong relation between the increasing Motor Unit (M.U.) firing rate and the recruitment of new units, the role of these two mechanisms involved in the development of force is not completely clear in the peripheral pathology. Besides the effects of motor involvement, a sensory loss especially of the proprioceptive components, seems to modify the force developed during voluntary muscular contraction through changes of the gamma-loop or in the presynaptic Ia inhibition.
The purpose of the present study was to determine whether both the recruitment and the firing of different M.U.s are modified in peripheral nervous system diseases. Recordings were obtained from patients suffering from different neuropathies. We divided them in 4 groups on the basis of histopathological and electrophysiological findings : motor, sensory, mixed axonal or mixed demyelinating neuropathies. In all the experiments the recruitment and the firing of anterior tibial M.U.s were investigated using high impedance tungsten microelectrodes, permitting a clear identification of the single M.U.s potentials during both minimal and maximal substained isometric contractions. In such a muscle with broad recruitment range we tried to distinguish two extreme types of fatigue-resistant M.U.s, the low and the high threshold ones, in different tasks, with or without visual or auditory feed-back. The frequency of the first recruited low threshold M.U.s was found to be generally higher than in controls. This finding was associated with the severity of strength reduction irrespective of the patholology. In selective motor neuron involvement we found a tendency towards a reduction of M.U.s discharge frequency and endurance differences. We also recorded high firing rates for the high threshold M.U.s in both axonal and demyelinating processes, but the maximum values were not significantly higher than in controls and they seemed to be related to the degree of force reduction. In sensitive neuropathies large and random fluctuations of M.U.s discharges frequency were obtained for both types of M.U.s. This finding was also associated with difficulties in activating high threshold M.U.s. If these data can sometimes to be the risult of Ia afferent involvement, in other cases they are better explaned by selective loss of M.U.s or modifications in their electrophysiological characteristics.

ANALYSIS OF CO-CONTRACTION IN LOWER LIMB ANTAGONISTIC MUSCLES DURING HUMAN WALKING

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Aim of the present study was the development of a computerized method for quantitative assessment of co-contraction in couples of functionally-related muscles during gait.

The method involves preprocessing of the surface EMG signal, recorded by means of an eight channel optic fiber electromyograph. Identification of single step cycles and rejection of artifacts (movement, crosstalk) are performed first, using interactive algorithms. Thereafter, EMG is high-pass filtered (15 Hz), rectified, low-pass filtered (35 Hz), and normalized in time and amplitude. A significance threshold (15% of the maximal locomotor EMG) is finally defined on the signal amplitude.

Quantitative assessment of co-contraction is obtained by means of a fully automatized algorithm, which provides three output parameters: a) Temporal Overlapping, i.e. the time, expressed as a percentage of the whole step cycle duration, during which the two muscles are simultaneously activated above the significance threshold. b) Geometrical Overlapping, i. e. the area of the EMG activity of the two muscles during the common activation period, normalized to the sum of the locomotor EMGs of the same muscles. c) Relative Contribution of each muscle, expressed as a perecntage of the total EMG output during the coactivation period.

The method was tested in ten normal adults, walking at natural speed, for antagonistic muscle pairs acting at the tibio- tarsal joint: Tibialis Ant.- Gastroc. Lat., Tibialis Ant.- Gastroc. Med. and Tibialis Ant.- Soleus. Results will be used as preliminary normative data for clinical application in motor disorders characterized by loss of the normal reciprocal inhibitory pattern during voluntary movements.

A SPECTRAL EMG SYSTEM FOR ASSESSING BACK MUSCLE IMPAIRMENT

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The diagnosis and management of back muscle impairment associated with low back pain (LBP) remains an elusive but critical element to providing successful prevention and treatment of spinal disorders. By developing effective quantifiable measures of muscle performance, LBP rehabilitation can be directed towards attaining physical goals that directly benefit the patient's functional ability. This revised approach to LBP treatment is likely to provide a more effective alternative to treatment approaches solely based upon self-report measurements of pain and disability.

We have developed and are currently testing a unique assessment procedure to fulfill the need for objective measurement of back muscle function. The technique has been implemented in a device called the Back Analysis System (BAS). The BAS measures and analyzes the endurance capacity of lumbar paraspinal muscles via EMG spectral analysis from an array of 6 surface EMG electrodes. When back muscles are deconditioned or inhibited by pain or mechanical deficits, a predictable change in the normal pattern of EMG spectral shifts among the different back muscles occur. An information MAP is created by the initial value, slope and recovery of EMG median frequency from the 6 concurrent channels of EMG signals recorded during the BAS tests to monitor fatigue and recovery. It is the alteration of this information map that distinguishes normal from abnormal back muscle function.

Research efforts to date have identified the median frequency map characteristic of normal back muscle function. Further efforts are underway to identify the influence of age, sex and anthropometric factors such as weight and height to modifying normative data. Clinical trials have also been implemented to standardize a clinical protocol and validate its use for monitoring treatment and predicting outcome. Preliminary studies have focused on chronic LBP patients from sedentary and athletic populations which have been compared to control groups without LBP. Results have identified significant differences in median frequency findings between the LBP and control groups. Differences are muscle-specific as well as force dependent. The discriminating ability of median frequency parameters for LBP was superior to standard clinical measures of mobility and strength. LBP patients were correctly identified with better than 90% accuracy purely on the basis of median frequency parameters. Although BAS results were nearly equal in their discriminating ability, the discriminating variables were different for studies involving sedentary subjects and athletes. Different mechanisms may therefore underlie muscle impairment in patients with LBP disorder.

CONCENTRIC MACRO EMG STUDIES OF NORMAL, MYOPATHIC AND NEUROGENIC MOTOR UNIT ACTION POTENTIALS

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Using a Teflon coated concentric needle, the Concentric Macro electrode described by Jabre in 1987 allows us to record simultaneously the concentric and Macro action potentials of a motor unit.

The study of these potentials makes it possible to evaluate the relationship between the small sample size studied by the concentric electrode (10-15 muscle fibers) and that of the majority of muscle fibers in a given unit, recorded by the Macro surface.

One may ask how representative are these 10-15 fibers, as seen by the concentric electrode, of the motor unit as a whole; and, which of the parameters measured (i.e. duration, area and amplitude) are most representative of that unit.

To answer these questions, we studied 10 men ranging in age from 21 - 61 years without evidence of nerve or muscle disease, and 14 patients, 13 men and 1 woman aged 24 to 73 years old, 7 with myopathic and 7 with neurogenic motor unit changes. The concentric and Macro action potentials of a total of 86 motor units were collected in each group. The duration, area and amplitude of these action potentials were then correlated. The ConMac electrode is particularly well suited for these studies since it allows the selection of only those units which fit particular electrophysiological criteria.

In normals, our study revealed significant correlations between concentric and Macro samples, with the concentric potential's area correlating better with the Macro potential than its amplitude. In the patients with myopathy, we found that the concentric motor unit action potential's area correlates strongly with the ConMac potential, even better than in normals while its amplitude correlates less. In the neurogenic group, we found that both the concentric motor unit action potential's area and amplitude correlated very well with their ConMac counterpart, more so than in normals.

Thus in pathology as in normals, measuring the concentric motor unit action potential's area in addition to its amplitude adds to the diagnostic sensitivity of motor unit potential measurements. These findings are discussed in light of the known dynamic and architectural motor unit changes which take place in the myopathic and neurogenic motor unit.

THE STUDY OF MOTOR UNIT FIRING RATES, RECRUITMENT THRESHOLD AND SIZE USING PRECISION DECOMPOSITION AND MACRO EMG TECHNIQUES

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In the late 60's, Henneman and colleagues described the existence of a size principle which governs the orderly recruitment of motor units. This principle states that motor units are recruited in a relatively fixed order according to size; the smallest units with the smallest input resistance anterior horn cells are recruited first and the largest units, with the largest input resistance anterior horn cells, recruited last.

The description of the size principle spawned a whole new era of research into the firing rate behavior of motor units. Efforts to study the contribution of this dimension to the study of motor control began to surface in the early 70's. Milner-Brown, Stein and Yemm attempted to verify the existence of the size principle in humans in a study of the First Dorsal Interosseous.

We set out to study the relationships between the recruitment order of motor units, their size as measured by macroEMG, their firing rates, and their firing rates behavior. We studied 45 muscle contractions during a trapezoidal 50% maximum voluntary contraction (MVC) ramp contraction. Our investigations were conducted in the First Dorsal Interosseous and Tibialis Anterior muscles of 11 normal subjects ranging in age form 20 to 33 years. Thirty-eight contractions were obtained from FDI and 7 from TA muscles. A total of 165 motor units were analyzed.

Statistical analysis of the data using repeated measures MANOVA and Friedman test showed that the recruitment order of motor units is directly related to the recruitment threshold ($p<0.00005$) and that, with sequential activation, there is a progressive increase in the motor units macroEMG potential area ($p=0.002$). This is in keeping with the size principle of motor units recruitment.

The mean firing rate calculated over the total contraction, and the steady state firing rate calculated over the constant force segment of the contraction, also varied inversely with the order of recruitment ($p=0.006$); and there was a high cross correlation between the firing rates of all the active motor units in a contraction ($p=0.006$)

In this presentation we will review our work with Precision Decomposition and present the results of our investigations using this technique in combination with macroEMG.

This work was supported by the Veterans Administration Rehabilitation R & D Service.

MAPPED SEP (SOMATOSENSORY EVOKED POTENTIALS) TO MECHANICAL STIMULI IN PATIENTS AFFECTED BY CEREBROVASCULAR ACCIDENTS

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Recovery from stroke is different in patients with motor loss alone versus patients with motor loss and alteration of the sensitivity. Prognosis is worse when hemiplegia is associated with hemianesthesia.
It's well-known that SEP's recording in stroke patients gives prognostic information when the patient had an impairment of coscience; SEP's integrity may predict recovery.
Bit-Mapped color imaging of Somatosensory Evoked Potentials (SEP) to mechanical stimuli confirm the usefulness of SEP's recording in patients affected by cerebrovascular accidents.
We recorded with Ag/AgCl disk scalp electrodes SEPs to electrical stimulation of both median nerve at the wrist and digital nerve, and to mechanical stimuli applied on the fingertip of the forefinger (on the side of the palm) from 19 derivations of the international 10-20 system; reference was at the linked earlobes and ground at Fpz.
By using the classic 4-nearest neighbors Shepard's method of interpolation it was possible to obtain color images of the distribution of electrical field.
The latencies of the different waves of SEP were shorter after the electrical stimulation of median nerve at the wrist and longer after mechanical stimulation of the index fingertip.
The field distribution of the different components of SEP was always identical whatever is the stimulation modality employed.
In healthy volunteers the presence of component P22 after mechanical stimulation confirms the hypothesis that this electrical event is dependent on activation of motor cortex by cutaneous afference; the afferences of cutaneous tactile sensations converge on the motor area 4, through thalamocortical pathways independently of connections via parietal areas.
These data give us further insight on the pathophysiology of somatosensory deficits, and the study of these mapped SEP components, generated in area 4, permit to improve the techniques of SEP's recording for diagnostic aims in stroke patients, and can offer further prognostic tool to physiatrists.

DOPAMINERGIC DRUGS DO NOT MODIFY THE BLINK REFLEX RECOVERY CYCLE IN ESSENTIAL BLEPHAROSPASM

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The pathogenesis of essential blepharospasm (EB) is still unknown and the role of dopaminergic system is controversial hyperexcitability of the blink reflex recovery cycle is reported and attributed to a reduction of interneuronal inhibitory effect on the polysynaptic reflex arck.
As far as the neuropharmacological aspect is concerned, different drugs have been tested in EB acting on dopaminergic, cholinergic and gabaergic systems The experience of Marsden et Al. leads the authors to affirm that there is no consistent pharmacologic response to drugs manipulating either the dopamine or acetylcholine systems and the differences among cases reported suggest no uniform pharmacological profile in this disorder In a few patients presenting EB associated with Psrkinson's disease, the dopaminergic drug induced improvement in parkinsonian symptoms was related to a worsening of EB symptoms, notwithstanding that Parkinson's disease is also characterised by a hyperexcitability of blink reflex recovery cycle, which improves using dopaminergic drugs Thus we studied the effect of different dopaminergic drugs on the excitability of the blink reflex recovery cycle in 5 EB patients The recovery cycle of the R2 component of the- blink reflex was evaluated before and after the administration of levodopa (500 mg. po), lysuride (1 mg po), apomorphine (10 and 50 ug/Kg sc) To avoid the peripheral effects of these drug, the patients were pretreated with domperidone (60 mg po tid) None of these drugs was able to unequivocally modify the excitability of the blink reflex recovery cycle We conclude that in EB the brainstem interneuronal hypersctivity which underlies the increased excitability of the blink reflex recovery cycle, differently from what has been found in parkinsonian subjects, is not related to a decrease in the inhibitory dopaminergic drive from the basal ganglia

TRH-T IN THE TREATMENT OF THE UPPER MOTONEURON SYNDROME (u.m.s.): EVALUATION OF FORGE AND EMG FATIQUE PHENOMENA

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One promising attempt for treating u.m.s. syndrome might be to interfere with residual activity of the pyramidal tract or on others spinal pathways acting on motoneurons. Terminals of descending fibres containing Thyrotropin Releasing Hormon (TRH) have been visualized in the human spinal cord around the motoneurone. Furthermore TRH receptors have also been identifiable in a similar distribution as that of TRH. These factors suggest that TRH could play a functional role in the control of motoneurons activity, probably facilitating cholinergic transmission. A slowly developing sustained excitation of frog and rat motoneurons after TRH-T has been shown by Nistri and Lucy (1988). A clinical trial by Delwaide (1988) on patients affected by chronic post stroke u.m.s. revealed both spasticity reduction and muscle strength increase. According to these data we verified in patients affected by u.m.s. the effect of acute TRH-T administration (4 mg protirelin tartrate) on motor recruitment of elbow flexor synergist muscles. EMG activity was recorded from biceps brachii (BB) and brachioradialis (BR) by surface electrodes. 500 msec epochs EMG signal were sampled and for each of them power spectrum median frequency (Fm) and corrected root mean square (cRMS) according to Kranz et al. (1985) were calculated. Six patients 2O-70 yrs. affected by hemiparesis as a consequence of ictus cerebri, were included in the study. The interval between stroke event and study ranged from 2 months to 5 years. Each subject performed an isometric elbow flexion at 50% of maximal voluntary contraction, for 60" receiving visual feedback of force, after that maximal force have been assessed during short 5 contractions. After TRH-T administration maximal strength l-reased in all patients, but consistently in three cases.
In basal condition Fm values during the exercise showed a decrement in three cases in BR. No decrement appeared for BB muscle. After TRH-T administration the Fm decrement in BR disappeared. Regarding cRMS values in basal condition, in BR some oscillations in one case and decrement in two, appeared. In BB oscillations in one and decrement in one were observed. After TRH-T a slight final increment was observed in three cases in BR and in two cases in BB, while a decrement in BB was observed parallely to the increment in the BR in two cases. In no case negative cardiocirculatory effects were observed. These data suggest the existence of some favourable effects of the acute administration of TRH-T on strength and fatigue resistance in our subjects with u.m.s. The control of both byomechanical and EMG parameters and of synergist muscles allows to better interpretate these processes, which nevertheless have to be furtherly verified.

INFLUENCE OF MUSCLE LENGTH ON FATIGUE IN MUSCULAR DYSTROPHIES

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Introduction: Muscle length is a relevant factor in conditioning functional characteristics of the contractile tissue, both in basal conditions and in fatigue experimental protocols. Fitch and Mc Comas (1) found that less fatigue is observed in normals when muscle contracts at optimal length, compared to shortened length, suggesting that contractile and, perhaps, excitation-contraction mechanisms can be involved in explaining this difference. On the basis of these results and the implications that they can have with respect to rehabilitation in muscular dystrophies (MD), the influence of muscle length on fatigue has been assessed in patients affected by diffent forms of MD.

Methods: The effects of fatigue in foot dorsiflexor muscles, using tetanizing deep peroneal stimulation, before and at different times after a contraction protocol in isometric conditions, were investigated in 4 patients with Becker, 3 with FSH MD and 5 with Miotonic Dystrophy (MyD), compared to 6 age-matched controls. The fatiguing protocol was performed by means of 15' intermittent maximal voluntary contractions, 3 sec on-2 sec off. Stimulation of peroneal nerve was performed using supramaximal, 0.1 msec duration, 20 Hz, 500 msec tetanizing sequences. Force was recorded in isometric condition using a strain-gauge and EMG by means of silver chloride superficial cup electrodes. Two different muscle lengths were studied, corresponding to angles of the ankle joint of 115° (optimal length: Lo) and 85° (shorten length: Ls). Following parameters were evaluated in basal conditions and at times of 30 sec, 1, 2, 3, 5, 10 and 15 min after the end of the fatiguing protocol: peak to peak M-wave amplitute and duration, contraction- and half-relaxation time, tetanus-peak force.

Results and Discussion: No statistical differences were observed in M-wave parameters after the fatigue protocol. In both length conditions the decrement of tetanus-peak force was greater in MD compared to the other two groups. Furthermore, while in normals and MyD patients the decrement of tetanic force was less pronounced at Ls with respect to Ls, in MD patients the fatigue was at comparable levels at the two lengths.
It is concluded that the dystrophic process appears to minimize the influence of muscle length on fatigue, probably because of the greater susceptibility of dystrophic muscle to the damage induced by the contraction itself (2).

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THE PLANTAR FASCIITIS OF THE FOOT: CLINICAL AND BIOMECHANICAL EVALUATION

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Plantar fasciitis of the foot is an inflammatory condition characterized by pain in the medial, central and lateral sectors of the sole accompanied by stiffness. The aim of this study is to examine this condition from a clinical and estesiological point of view and in relation to the objective findings of a biomechanical evaluation.

22 subjects aged 19-51 who practised sport regularly, were examined; all presented with pain in the plantar fascia of the foot. Each subject underwent a clinical and estesiological examination with measurement of pain tresholds to electrical stimulation of the skin, subcutaneous tissue and muscle in the trigger point zone and in the corresponding area controlateral and in addition an evaluation of biomechanical parameters related to function by means of an analysis of the foot-ground reaction and peak force revealed by isokinetic dynamometer. The subjects were divided into 3 groups on the basis of the TrP site and area of referred pain, as follows: Group A (12 cases) had fascial pain in the calcaneal inserction. The objective examination showed an active TrP causing referred pain to pressure in the median part of the central aponeurosis and palbable band. Group B (8 cases) had pain localysed in the middle third of the medial part of the fasciae. The objective examination showed an active TrP in this area which causes pain to pressure radiating disto-proximally. Group C (2 cases) had pain localised in the middle third of the lateral part of the fasciae. The objective examination showed an active TrP causing pain to pressure radiating disto-proximally to the attachmant of the peroneus muscle at the base of the 5° metatarsus.

The estesiological evaluation of these groups showed a lowered pain treshold in the skin and muscle compared with the controlateral areas. The isokinetic examination revealed in these groups a significant decrease in the peak force in concentric and eccentric contraction. The foot-ground reactions were abnormal in the morphology of the butterfly diagram and in the spatial and temporal components.

ELEKTROKINESIOLOGIC (EK) MEASUREMENTS OF TIZANIDIN EFFECT ON SPINAL FUNCTION

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Introduction
Elektrokinesiologic (EK) assessment of lumbar spine is a method of measuring spinal mobility and EMG activity of paravertebral muscles [1]. Computerized protocol is programmed for immediate access to the measuring set-up allowing analysis of both EMG pattern and kinematic data of spinal movements.

Tizanidine, an imidazoline derivative and centrally acting muscle relaxant, has been used as a muscle relaxant in patients with chronic muscle spasm of neurological origin and in patients with acute low back pain. [2].

This report is based on our hypothesis that tizanidine may be used in chronic low back pain (LBP) when muscle spasm is evident.

Methods
Twelve chronic LBP patients (7 men and 5 women), mean age 39.6 years (range 27-52), and mean weight 81 kg (59-118) were included.

EK measures were compared at three successive testings: at baseline, two hours after first intake of tizanidine (2mg) and after two weeks of treatment with tizanidine (3x2mg).

Elektrokinesiologic assessment used an MS6 EMG system with integrators, POLGON Goniometer PG6 and computer (MDS Hero). EMG activity was normalized and expressed as a percentage of maximal voluntary contraction (%MVC). EMG measures for peak flexion (F), averaged values for both F and extension (E), and kinematic measures: range of movement (ROM), duration and speed of movement, were recorded.

All statistical analyses were conducted with probability of 0.05. Friedman two-way analysis of variance with ranks was used, and group mean measures were compared by Wilcoxon rank test.

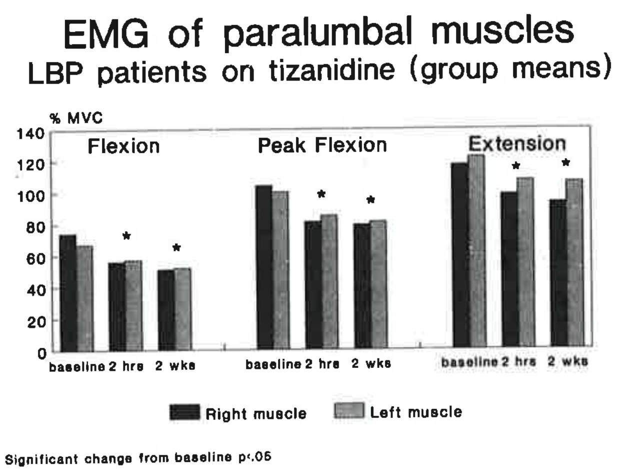

Figure 1

Results
Figure 1 presents the means of paralumbar EMG during F, peak F and E at the baseline and two follow-ups. EMG values significantly decreased from baseline levels at 2-hours and 2-weeks assessments ($p<0.05$). The group means of ROM did not change significantly. Group means for duration and speed of F-E movement improved significantly at 2-weeks assessments.

Conclusion
EMG of paralumbar muscles decreased significantly two hours after tizanidine intake, which was even more pronounced after two weeks. Duration and speed of movement were significantly improved after two weeks of tizanidine treatment. ROM did not change at follow-ups. EMG response following a single dose of tizanidine should be considered a prognostic sign for the improvement of spinal function.

It appears that tizanidine may be one choice for the beneficial therapy of chronic LBP; chronic LBP patients have improved spinal function regarding changes in EMG of paralumbar muscles and both speed and duration of body movement.

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ELECTROPHYSIOLOGICAL EFFECTS OF 3, 4-DIAMINOPYRIDINE IN PERIPHERAL NEUROPATHIES

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Introduction

3,4-Diaminopyridine (3,4-DAP) is known to improve an action potential propagation as a potassium channel blocker in experimentally demyelinated nerve. In this report we administered oral 3,4-DAP to the patients with various peripheral neuropathies and studied electrophysiological effects by monitoring nerve action potentials, before and after the administration of 3,4-DAP.

Subjects and methods

After informed consent was obtained, 14 patients with peripheral neuropathies including hereditary motor sensory neuropathy type I, II, and III, polyradiculoneuropathy, recurrent inflammatory radiculoneuropathy, and diabetic neuropathy were administered oral 3,4-DAP and monitored NCV in weak muscle of each patient before and after the administration of 3,4-DAP. The dosage of 3,4-DAP was 30mg, 45mg and 60mg. Monitoring of NCV was done every about 30 minutes. The clinical testing of 3,4-DAP was accepted by the Ethics Committee of the University Hospital.

Results

1) M wave amplitudes increased gradually after the administration of 3,4-DAP in some cases of peripheral neuropathies such as polyradiculoneuropathy and recurrent inflammatory radiculoneuropathy. 2) When M wave had two peaks, faster peak of them tended to be higher after the administration of 3,4-DAP as compared with before it. 3) Electrophysiological effects of 3,4-DAP tended to appear in lower dosage such as 30mg and 45mg. In case of 45mg or 60mg 3,4-DAP, adverse effects such as perioral or distal paresthesia and abdominal discomfort sometimes came out.

Conclusion

In some patients with peripheral neuropathies, excitability of peripheral nerves was changed to be more active by oral administration of 3,4-DAP and it might renovate the function of peripheral neuropathies.

MUSCLE STRENGHT'S DEFICIT IN ADOLESCENTS WITH DOWN'S SYNDROME (DS): A SIGNAL OF PRECOCIOUS AGEING OF THE NEUROMUSCULAR SYSTEM?

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Muscle strength is a crucial factor in the development and functioning of motor behavior, and it is probably the most age-related between the motor parameters. Despite its functional importance, little informations are reported about the characteristics of muscle strength during the development of subjects with DS. Therefore the purpose of this study was to analyse the isokinetic strength of knee extensor muscles of twenty five children and adolescents with DS (14 males and 11 females, aged 11-18 years).

The control groups, matched for age, weight and height, consisted of 40 normal subjects (22 males and 18 females), 20 subjects with mental retardation of unknown origin (11 males and 9 females), 10 subjects with Fragile X syndrome (all males). Muscle strength of knee extensor muscles of both limbs, was evaluated during two sessions, on two different days, by an isokinetic dynamometer at the speed of 30°/sec. Integrated EMG was recorded from m.m. ventralis lateralis. Knee extensor muscles of one limb were considered dominant when stronger of 20 % or more than the controlateral ones.

All DS subjects were markedly and significantly weaker than normals at all ages, but by the age of 13-14 years they failed to show the physiological increase of muscle strength observed during adolescence. Patients with mental retardation or with Fragile X syndrome were significantly stronger than DS patients and showed values of peak torque similar to the normals. Muscle strength dominance was observed more frequently in DS patients (11/25) than in normals (6/40) (chi square = 5.28, p <.05). Knee extensor muscles of the left side were the dominant in most of DS subjects (9/11).

Our conclusions are twofold: 1. subjects with DS are weaker than normals during late childhood and adolescence, probably because of a basic defect in the development of muscles and/or of a dysfunction of the motor control (reduction in the number of pyramidal neurons located in the motor cortex ? Marin-Padilla, 1976); 2. the lack of the physiological increase of muscle strength by the age of 13-14 years, in the face of normal sex hormon levels (Pueschel, 1990), suggests that strength's deficit in adolescents with DS could be possibly the expression of a precocious ageing and deterioration of the neuromuscular junction, as showed by other authors in transgenic mice carrying an extra human CuZnSOD gene (Avraham et al., 1991) or in the tongue muscles of DS subjects (Yarom et al., 1987).

CO-CONTRACTION IN CEREBRAL PALSY DURING GAIT

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Introduction

Gait Analysis data are being used as an adjunct to clinical decision making. Two factors are important: reliability and repeatibility, both intra and inter subjects. We have particularly studied, regarding EMG data, the co-contraction areas of two antagonist muscles referred to normal.

Methods

This study was conducted with surface EMG electrodes positioned on Triceps Surae and Tibialis Anterior; the recorded EMG was related with basographic data. The sample consisted of 18 subjects, aged between 3-8 years, with Cerebral Palsy, Gait data were compared between different test days; a trial consisted in a walk of 15 meters, every subject performed two trials. EMG was normalized to maximal EMG related to every child.

Results

Tab.A (see) represents the increasing of percentage of co-contraction in children with C.P. referred to age. In Tab.B (see) we demonstrate that the greater intra-subject variation of the cocontraction area is in C.P. children, compared to normal peers. In the last Tab.C. (see), we have divided the co-contraction patterns of the gait cycle in three tipologies:

<u>Increasing area</u>: determined for homologous recruitment of both muscles.
<u>Shaded area</u>: determined for mixed recruitment and decruitment of muscles.
<u>Reversal area</u>: determined for homologous decruitment of both muscles.

tab. A

age	normal	pathol.
3y	58%	85%
4y	63%	73%
5y	50%	81%
6y	50%	89%
7y	41%	73%
8y	36%	86%

tab. B

normal	CV	pathol.	CV
50%	12.1%	81%	21.3%
56%	16.7%	70%	12.5%
41%	9.8%	77%	28.1%
45%	9.4%	80%	32.7%
35%	8.1%	66%	27.4%
33%	8.3%	86%	49.9%

tab. C

gait	increasing		shaded		reversal	
%	nor.	path.	nor.	path.	nor.	path.
0	100%	27.2%	0	72.7%	0	0
10	0	36.3%	100%	72.7%	0	18.1%
20	0	45.4%	100%	63.6%	0	0
30	0	36.3%	100%	72.7%	0	0
40	0	36.3%	100%	81.8%	0	27.2%
50	0	27.2%	100%	54.5%	0	45.4%
60	0	27.2%	100%	36.3%	0	45.4%
70	0	0	0	63.6%	100%	36.3%
80	0	18.1%	100%	54.5%	0	27.2%
90	100%	9.1%	0	63.6%	0	27.2%

Conclusion

Results indicated that percented cocontraction normally decreases during the developmental age. This pattern was not observed in the pathological sample. The patterns of C.P. children, characterized by an increased percented cocontraction, suggest that the original control mechanism of infants is not transformed in children with C.P. and indicates inefficient sensory feedback. Coactivation of all leg muscles predominates during the stance phase of gait. At this stage children with C.P. demonstrate greater intra-subject variation in cocontraction components compared to their non handicapped peers. This may signify that handicapped children do not possess the same degree of automatic control as non handicapped children.

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SPECTRAL EMG ASSESSMENT OF MUSCLE ADAPTATION TO SPACEFLIGHT

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Exposure to microgravity during spaceflight can result in a profound impairment of muscle function. Numerous studies in animal and man during microgravity and land-based models of microgravity have described alterations in the force, velocity and endurance characteristics of muscle. Most of the techniques used were invasive or not amenable to repetitive measurement of the same muscle. Surface electromyography (EMG) combined with spectral analysis does not suffer from these limitations and has been used successfully to monitor the functional capacity of muscle, but not in association with prolonged spaceflight. We hypothesized that this technique would be useful in this context because EMG spectral parameters are influenced by muscle cross-sectional area and enzyme profile, both of which are typically modified by exposure to microgravity.

Four astronauts (2 Male; 2 Female) were recruited for this study as a part of the 9-day NASA Spacelab Life Science 1 (SLS-01) shuttle mission. Subjects were tested at approximately 180, 75, 45, and 15 days prior to launch and again within hours following landing and days 2, 4, and 6 post-flight. Testing consisted of concurrent surface EMG measurement of median frequency from the tibialis anterior, medial gastrocnemius and soleus muscles during 30 s sustained isometric contractions in dorsiflexion and plantarflexion. All contractions were sustained at 40% and 80% of the pre-flight maximal voluntary contraction (MVC).

Results indicated that static strength, as measured by the MVC, was reduced post-flight by 10%-40% in plantarflexion but not in dorsiflexion. Strength decrements recovered by post-flight day 6. All subjects, except one, demonstrated a post-flight reduction in the initial median frequency (IMF) of 10%-20%. This reduction was more evident in the soleus than the gastrocnemius or tibialis anterior muscles. With few exceptions, post-flight IMF recovered in 4-6 days. The rate of MF decay during a sustained contraction was quantified by the slope of a least-squares regression line fitted to the data (MF slope). MF slope data was more variable than IMF data. In most cases, however, the MF slope was more negative post-flight compared to pre-flight. This effect was most evident in the soleus muscle, followed by the gastrocnemius and tibialis anterior.

Changes in IMF and MF slope in this study are consistent with previous reports describing the presence of muscle atrophy and enzyme shifts towards the metabolic profile of type II fibers. Similarly, these effects were most prominent in muscles with a predominant postural function. Further studies are needed to validate that changes in EMG spectral parameters are the result of the physiological adaptations which occur in muscle exposed to spaceflight.

NON-INVASIVE ASSESSMENT OF NEURAL RESPIRATORY DRIVE IN MAN: SURFACE ELECTROMYOGRAPHIC ACTIVITY OF THE DIAPHRAGM

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INTRODUCTION: Previous studies have demonstrated the usefulness of recording electromyographic activity of the diaphragm (EMGd) in assessing neural respiratory drive in animals and in human beings. EMGd may be recorded by surface electrodes; this procedure has been proposed as an useful, non invasive measurement of neural respiratory drive for clinical purposes. In the present studies we have evaluated surface EMGd (EMGds) in 10 normal subjects and in 40 patients with chronic obstructive pulmonary disease (COPD) during breathing at rest and during stimulated breathing with carbon dioxide.

METHODS: EMGd was recorded by surface and esophageal (15 subjects) electrodes. Raw EMG signal was amplified, and filtered below 100 Hz and above 1000 Hz to remove as much of the ECG as possible, without significantly filtering EMG. The signal was then rectified, and integrated over the time (time constant 100 ms) to provide a measurement of change in average electrical activity as a function of time (moving time average), as previously described (Gorini et al., Am Rev Respir Dis 1990; 142: 289-294). EMG activity was quantified both as peak of activity and as rate of rise of activity (peak/inspiratory time). We also measured the maximal inspiratory pressure (MIP) and the pattern of breathing.

RESULTS: The reproducibility of EMGds response slope to CO_2 stimulation was good and in no case was the response obtained in one study twofold greater or less than that obtained in any other study. The interindividual variability of EMGds was not dissimilar to that observed using esophageal lead (EMGde). Furthermore, EMGds/EMGde ratio was not modified by changes in lung volume from 60 to 90% of total lung capacity. Compared with an age and sex-matched normal control group, COPD patients exhibited lower MIP and significantly greater EMGds. In patients with chronic hypercapnia MIP was significantly reduced whereas EMGds was significantly increased compared to patients with normocapnia. Hypercapnics exhibited a more rapid and shallower breathing than normocapnics. In the patients as a whole, multiple linear regression analysis showed a significant indirect relationship of EMGds with forced expiratory volume in 1 s (FEV1), arterial oxygen tension, and respiratory frequency.

CONCLUSIONS: 1) EMGds measurement seems to be reproducible and not affected by lung volume changes; 2) COPD patients have a greater neural drive to the respiratory muscles than normal subjects; 3) both chemical and mechanical afferent informations seem to be involved in this increased neural drive.

AN ANALYSIS OF THE "LATENCY TIME JITTER" IN CHOICE VERBAL REACTIONS AND THE AGING EFFECT

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Since 1975 electromyography has been associated with kinesiology in the study of speech production and control. This study implies a polygraphic recording including respiration, glottal contractions, labial and jaw movements.

The investigation of the jaw movements was primarily carried out for mastication and was intended to identify the distribution of innervation between antagonistic muscles in relationship with loading. Proprioceptive and exteroceptive, myotatic and inhibitory reflexes have been also evaluated in these muscles. It became evident that the effects and purpose of these reflexes in jaw movements are different for the two functions, speaking and mastication, and several studies have shown the relative autonomy of the articulatory movements of speech [1, 2] .

Our interest has been focused mainly on verbal choice reactions with the aim of analyzing the duration of the related central processes in reaction times (RT) and fractionated RTS. The following questions were faced: 1) the nature of the intraindividual variability in the RT; 2) the analysis of the early preparatory innervation as a process of timing; 3) the effect of aging.

Single letters, mono- and poly-sillabic words, were the targets to be pronounced with minimal and maximal preperiods for the study of the three physiological processes which are possibily involved in age-related deterioration: a) modular slowing; b) information loss; c) decay in ST (short-term memory) buffer.

Parts of this program has been now fulfilled and the preliminary results will be briefly reported.

Methodology

In the basic experiment the examiner presents on a P.C. screen some images or written words and asks the subject to pronounce them as soon as possible. The subject is confortably sitting in front of the screen with a dynamic microphone placed 15 cm in front of the mouth; small surface electrodes (Dantec 13L20) are placed over the oral and mandibular muscles. Both the microphone and the electrodes are connected to the preamplifiers of an electromyographer (Medelec - Mystro Plus) for recording respectively the acousticogram (ACG) and the electromyograms (EMGs). The muscles investigated were the orbicularis oris superior (OOS), the inferior (OOI) and the ventral bell of the digastric (Dig). The phonemes to be pronounced were the consonant /b/, the couples of words /clan/ - /cane/ and /mare/ - /muro/ and the non-words /ubavek/ - /ibavek/.

The subject was asked to pronounce the word immediately or with an interval or preperiod (t) between the presentation of the word and a "go" signal; thus t could be 0 (immediate response) or was chosen among 0.05, 0.1, 0.5, 1.5, 4 and 10 s.

A series of 12 reactions was requested for each word and t value. The interval between two successive reaction was 3 $\pm$ 1 s. Thirty-four normal Italian speaking subjects, 16 to 78 years old, were investigated. One spastic disarthric patient affected by cerebrovascular disease (56 years old) and two ALS patients (53 and 68 y. old) underwent the same examination.

Findings

1. Normal population.

The RTS mean values of a sery of successive reactions are shown in Fig. 1. Inside each single series of reactions, a trend towards shortening of the individual RTs was found. This diminution however is not regularly progressive, but it shows marked fluctuations (jitter). These are measured as standard deviation of the mean values of 12 reactions and can be taken into consideration as an index of the RTs jitter.

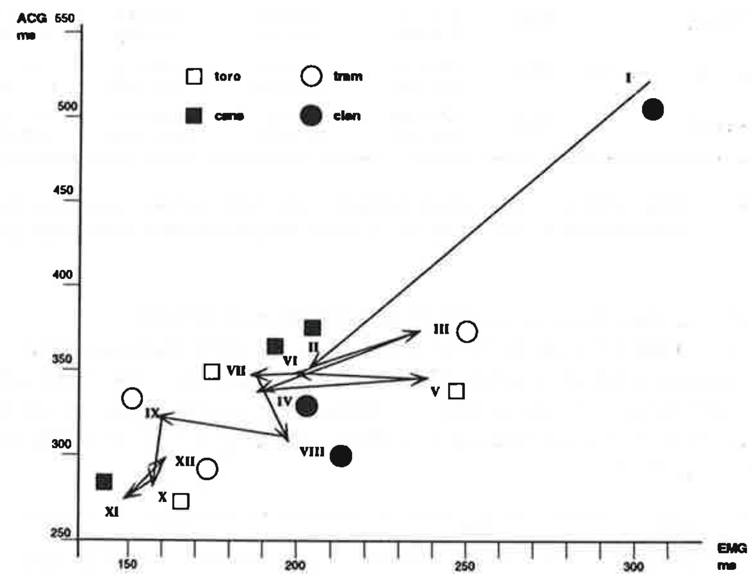

Fig. 1 EMG and ACG choice verbal reaction times in 12 successive trials in a normal subject (male, 35 y.o.).

1.1 Basic values.

The mean RT of both ACG and EMG responses in random choice reactions for couples of words showed a statistically significant increase in people aged more than 55 years (Tab. 1). This difference was evident only for the direct responses (t=0) whereas no significant change in comparison with these of the younger subjects appeared when a preperiod (t) of 3 s was interposed between the graphema and the "go" signal (Fig. 2, 3 and 4).

AGE (n. subjects)			16-25 (8)	26-35 (10)	36-60 (7)	61-78 (8)
Mean age			21.6	29	54.3	67.7
Choice Verbal Reactions	/clan/	EMG	232±44 (158-289)	244±57 (172-365)	267±74 (192-396)	313±50 (251-406)
		ACG	497±39 (417-550)	449±64 (383-541)	551±64 (480-673)	585±54 (472-641)
	/cane/	EMG	225±36 (156-275)	221±31 (185-265)	253±61 (200-357)	299±39 (240-354)
		ACG	470±36 (410-526)	437±61 (354-529)	530±41 (456-590)	553±52 (441-640)
Verbal Reactions with 3 s preperiod	/clan/	EMG	171±27 (131-202)	158±42 (123-256)	169±23 (140-199)	184±26 (144-217)
		ACG	377±47 (315-440)	332±66 (266-459)	427±56 (322-479)	381±49 (308-455)
	/cane/	EMG	164±32 (142-288)	160±33 (109-225)	169±13 (141-180)	177±20 (152-211)
		ACG	376±48 (309-449)	335±62 (252-434)	436±46 (341-474)	385±45 (329-462)

Table I Mean EMG and ACG verbal reaction times (with standard deviation, minimum and maximal values in ms) of 42 healty subjects divided in four age groups.

1.2 The standard deviations (s.d.) of the EMG RT and ACG RT.

The s.d. of the RT mean values of /cane/ EMG and ACG responses did not change with the increase in the mean values RT. As it is reported in the Table 2, the s.d. of the ACG RT increases progressively from the young to the middle age group (36-60 y.o.) ranging between 60 and 66 whereas it remains constant (s.d.: 61 to 66) in the older group (61-78 y.o.).

AGE	EMG		ACG	
years	/cane/	/clan/	/cane/	/clan/
16 - 25	37	51	43	71
26 - 35	44	59	60	60
36 - 60	62	60	66	61
61 - 78	61	78	66	80

Table 2 Intraindividual variations of RT values during choice verbal rections in four age groups of normal subjects.

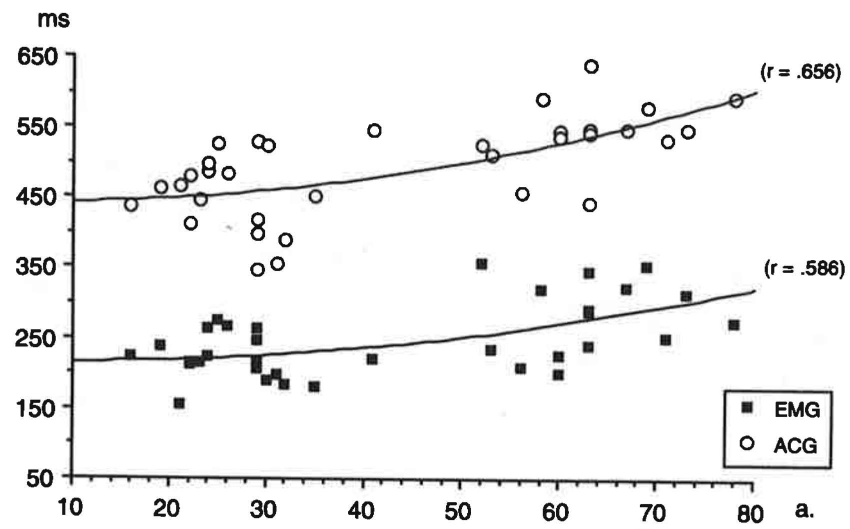

Fig. 2 Mean EMG and ACG choice reaction times for the word /cane/ (dog) in 34 healthy subjects aged between 16 and 78 years.

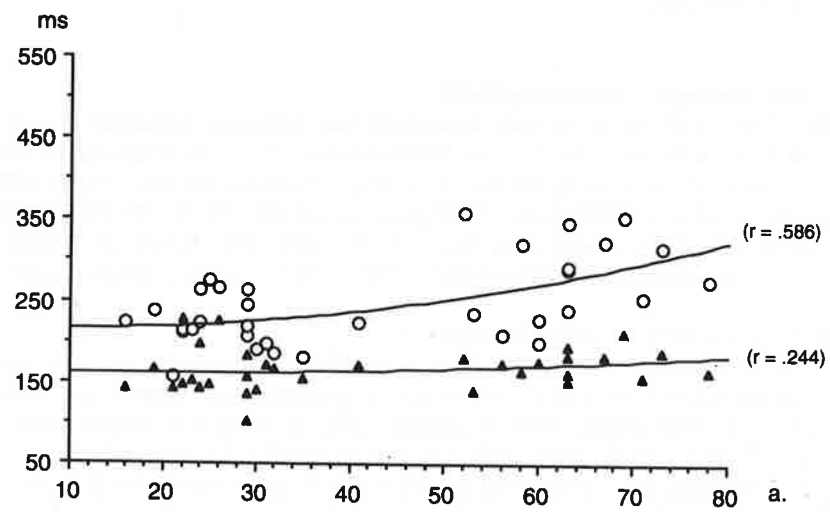

Fig. 3 Mean EMG choice (circles) and with a preperiod of 3 s (filled dots) reaction times for the word /cane/ (dog) in 34 healthy subjects aged between 16 and 78 years.

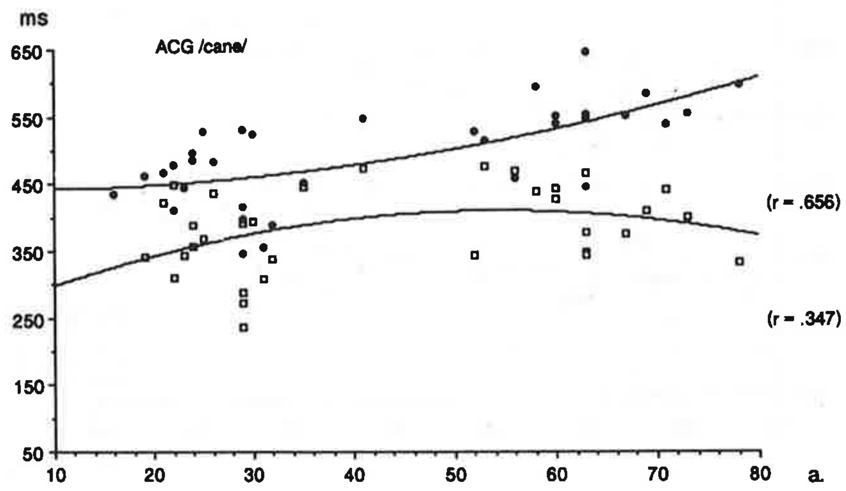

Fig. 4 Mean ACG choice (filled dots) and with a preperiod of 3 s (squares) reaction times for the word /cane/ (dog) in 34 healthy subjects aged between 16 and 78 years.

1.3 Preparatory and articulatory EMGs.

The duration of the preparatory innervation has been also calculated. It can be measured between the initial point of the EMG and the point in the EMG that preceeds the onset point of the ACG by 65 ms. This delay represents the time which elapses between the onset of OO articulatory EMG response and the onset of the corresponding muscular twitch which strictly coincides with the ACG [3]. Values of latency of preparatory innervation (EMG) are reported in Tab. 1 for the different groups of age.

1.4 Changes in RTs at different values of t.

The preperiods (t) produce a diminition of the RT (calculated from the "go" signal) when t ranges between 0.5 and 3 s. In the group of young subjects one could observe for some out of them a rather constant reduction (Fig. 5) whereas in several others the maximal reduction was observed with a preperiod (t) of 1.5 s (Fig. 6). In the old subjects the maximal RT reduction was present with a 4 s preperiod (t=4s) (Fig. 6).

The greatest differences between direct and with preperiod responses in the elder subjects are evident for extreme values of t: 0.05 s with a 145 ms increase, and for t: 10 s with a 20 ms increase.

2. Pathological cases.

2.1 The spastic dysarthric patient showed a mean RT of ACG of 580 ms with a st. d. of 43. The pattern of EMG and the shape of ACG of the successive reactions showed a greater regularity than in normal subjects (Fig. 7).

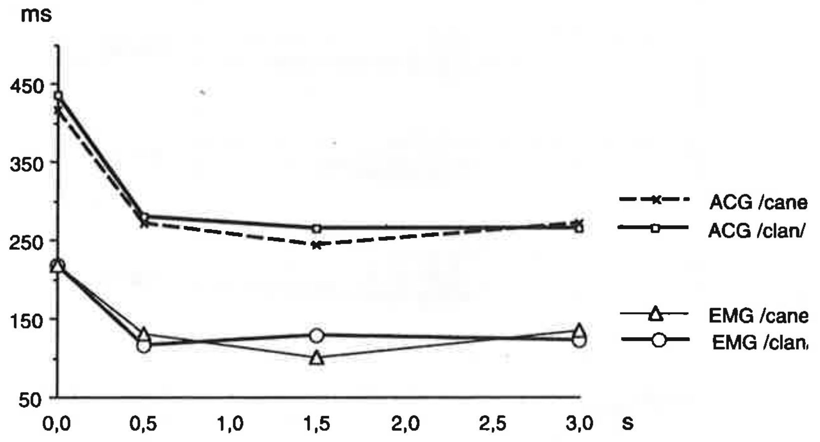

Fig 5 Mean EMG and ACG verbal RT with different preperiods in a normal subject (m, 29 y.o.)

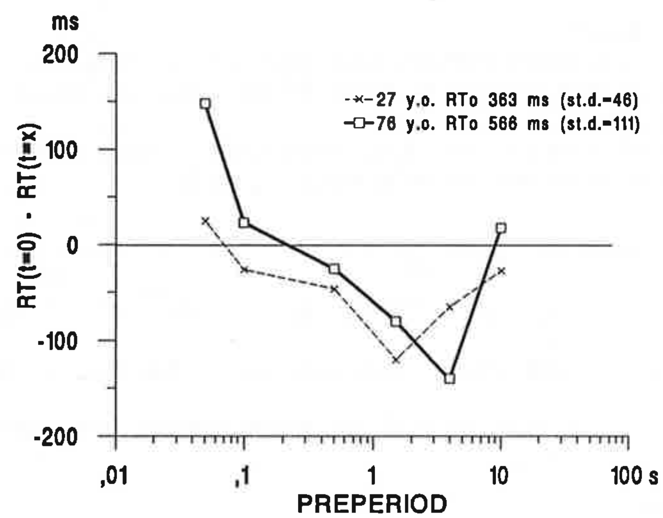

Fig 6 Difference between verbal RT with t=0 and different preperiods (t=x) in two normal subject (27 and 76 y.o.)

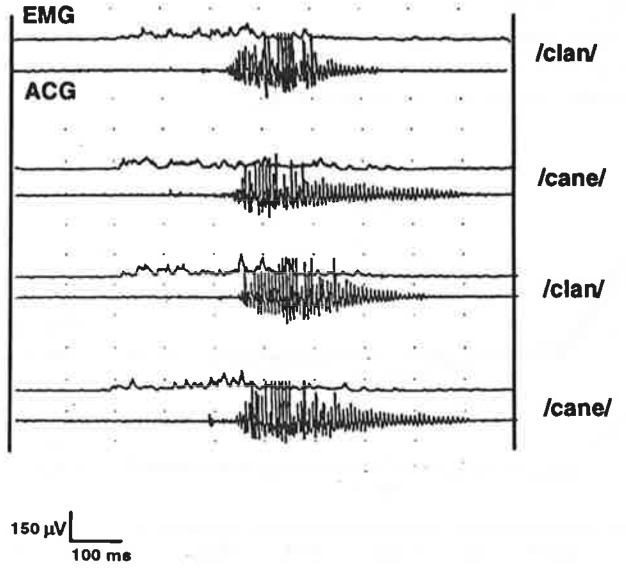

Fig. 7 EMG and ACG recordings of verbal reactions with a preperiod of 3 s in a patient affected by spastic dysarthria (male, 59 y.o.).

2.2 ALS patients.

In one ALS patient (48 years old, male) suffering from early dyshartric symptoms a slight increase in both EMG and ACG mean RT was found, while the s.d. values were reduced (Tab. 3).

In a second patient affected by MND (56 years old, female) mean RT values (EMG and ACG) were greatly increased and so was st.d. (Tab. 3).

ALS patients	EMG /clan/	EMG /cane/	ACG /clan/	ACG /cane/
48 y.o., m.	248 ± 20	269 ± 28	545 ± 51	626 ± 51
56 y.o., f.	379 ± 43	346 ± 59	798 ± 108	725 ± 61

Table 3 Mean choice EMG and ACG RT with standard deviation in two ALS patients.

Discussion

A central issue in the study of cerebral processes related to speaking is represented by the intraindividual variability in the utterance of the same phoneme, which involves the ACG and indipendently the RTs. It has been claimed, on the ground of several data and theoretical considerations [4], that the stochastic behavior depends on a balance

between opposite biological phenomena responsible for the Rt jitter. Our findings proves that it is independent on fatigue phenomena (Fig. 1) as well as on biological deterioration occurring in aging (Tab. 1). This kind of jitter implies a dualistic process; the oscillations and the corresponding s.d. may be interpreted as the result of two series of impulses, the ones acting as central invariant commands and the other as feedbacks assuring the correct variations for changing peripheral conditions [5].

The results on the behavior of such jitter in peripheral and central nervous system impairments throw new light on this interpretation. In the hemiplegic patient the ACG RT did not show an increase both in mean value and standard deviation. However a higher regularity was found in the whole pattern of the EMG and the shape of ACG. In one of the two ALS patients were present more constant values of the preparatory innervation RT than in normal subjects.

Eventually our research on RT with different preperiods shows that the greatest acceleration induced by the preperiod requires a longer preperiod in the older group. This shift implies that the aging effect is exerted on both modularity processes and information loss [6].

From a more general point of view the latency time jitter in choice reactions, far of representing a collateral negative phenomena, appears to be a primary event in the stochastic way of functioning of the central motor system. Previous investigations [3] did not show a relationship between the jitter of single M.U. firing frequency and the jitter of verbal choice RT, so that we could hardly explain the RT jitter with fluctuations of elementary biophysical processes. The fluctuations seem rather depend on intimate mechanisms of the whole production and control system which could be reproduced by a dualistic model of excitatory and inhibitory processes. If we combine this general concept with the previously mentioned more detailed analysis we could propose a more realistic interpretation.

In fact the control system of purposeful sequential actions seems to require an integration between invariant descending patterns of command and ascending feedbacks of adaptative servomechanisms which give rise to excitatory and inibitory processes responsible for the fluctuations observed at the output.

During the age-related physiologic-deterioration the slowing of processes resulting mainly from information loss [6] does not interfere with this balanced fluctuations until a critical point is reached: at this point the range of adaptations is reduced and a more stereotyped performance takes place. In fact at this latest phases of aging the adjustement is more coarse with a narrower role of feedbacks.

An abnormally increase in their stereotyped fluctuations can be found to occur in the lesions of the cortico-spinal system where a more marked diminution of fluctuations has been shown.

These data specifically concern speech production and control which seems more accessible to an experimental clinical investigation [3]. Since however the brain processes involved in other purposeful complex sequential actions have many common properties with those of speech production and control, the impact of these investigation for understanding the brain motor organization and its impairments deserves to be taken into consideration for both basic and applied neurology of other kind of motor performances like gait and manipulations.

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ASPECTS OF AN OSCILLATION THEORY OF SPEECH MOTOR CONTROL - EVIDENCE FROM KINEMATIC STUDIES WITH ELECTROMAGNETIC ARTICULOGRAPHY

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In recent studies making use of electromagnetic articulography, we have demonstrated that individual articulators may be characterized in terms of sinusoidal oscillatory movements. E.G. when German speaking subjects repeat the "ANNA"/ana/ the velum opens and closes the nasopharyngeal port in a rhythmic cyclic manner. Moreover we have shown that speech disorder associated with cerebellar disease interrupt these cyclic patterns in quantifiable ways. In this paper, we shall propose additional types of speech stimuli which may be useful in determining this. Paradigmatically, 3 types of oscillatory, coupled movement patterns are described: lingual vs labial, coupled labial-mandibular vs laryngeal oscillations.

Coupled lingual vs velar oscillations

In a simple tzo-articular model involving tongue tip and velum, alveolar non-nasal (e.g. ((n))) consonants may be generated by phase modulation of the oscillatory movements of the two articulators. When repetitive CV_nCV_n ... sequences (C=d, V_n=nasal vowel) are produced, tongue tip maxima (i.e. forming alveolar closure) coincides with velum maxima (i.e. forming closure of nasopharyngeal closure). This means that tongue tip oscillations are in phase with velum oscillations. When velum oscillatory movements are phase modulated by a shift of pi with reference to tongue tip oscillations, this results in production of CV_{nn} CV_{nn} " sequences with C= ((n)) and V_{nn} = non-nasal vowel.

Lingual vs labial oscillations

Consider the dynamic systems of tongue and lips in the production of the German vowels /u/ and /i/. When tongue and tip anterior movements occur in phase, the articulators are in position to produce the vowel /u/. If this phase relationship is reversed, i.e. tongue is fronted and the lips are retracted, the articulators are in position to produce the vowel /i/. If this phase relationship is synchronized so that the anterior position of tongue occurs when the lips are protruded the vowel /y/ (high front rounded) will result. Similarly, if the lips retract when the tongue is back (i.e. synchronized phase for posterior movement), an unrounded back vowel e.g. /y/ may result.
If one observes patients repeating sequences of vowels with defined phase differerences (e.g. /u/i/u/i) kinematic data for subjects speech errors may be very informative in determining phase coordination difficulties.

Coupled labial-mandibular vs laryngeal oscillations

A well explored characteristic of aphasic speechproduction is voice onset time (VOT). An insight which might be gained from an oscillatory explanation is that there may be a broad cyclic continuum in labial and laryngeal phasing for both voice onset and offset. For example, a subject in our laboratory was asked to produce repetitive tokens of /pa/. After being given a large dose of alcohol (blood level .07%), this subject often erroneously shifted into /ap/ /ap/ /ap/ productions. This type of speech error could be described as a loss of phase shift control. That is this subject does not produce lip and jaw movements prior to glotal pulsing, but shifts to a pattern of laryngeal activity, followed by lip and jaw movements prior to glotal pulsing, but shifts to a pattern of laryngeal activity, followed by lip and jaw closing. We speculate continuous phase shifting of lip/jaw and larynx leas to a continuum of sounds e.g. fro, /pa/to/ba/to/ab/to/ap/. Characteristic speech errors along this continuum might be expected to obtain with phasing difficulties in repetitive speech.

EXTEROCEPTIVE INTERFERENCE DURING VERBAL REACTIONS

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The oromandibular coordination of speech has been investigated during simultaneous exteroceptive stimulation of different kind. It has been assumed in previous papers that a gate control of interfering exteroceptive stimuli occurs in the mandibular muscles during speaking, but in spite of several detailed investigations, the specificity of the related process is still matter of debate. Jaw muscles activity (EMG) and acousticograms (ACG) were recorded in seven healthy subjects (22-35 years, 27.5 mean) during verbal and non-verbal movements. EMG activity was recorded from both masseter and digastric muscles by means of surface electrodes; ACG signals were recorded by means of a dynamic microphone placed 15 cm in front of the mouth. Electrical (exteroceptive) stimulation was applied contralaterally to the side of EMG recordings, 2 cm by the side of the mouth. Stimuli of 20 mA and 0.05 ms were delivered in three different conditions:
1) during tonic muscle voluntary contraction of about 25% of maximal;
2) during a simple reaction task of chewing (opening and closing alternate movements at about 2.5 Hz) 100, 200 and 300 ms after the appearance of the "go" signal (an asterisk was shown on a computer display);
3) during a simple verbal reaction task (utterance of the non-word /sas/) 100, 150, 200, 250 and 300 ms after the "go" signal.
Findings
First condition. Exteroceptive suppression of EMG activity was found to regularly occur in masseter muscle at short (ES1) and long (ES2; latency 40-50 ms and duration 30±10 ms) intervals after the stimulus.
Second condition. When the exteroceptive stimulation was applied 100 ms after the "go" signal no exteroceptive suppression was found to occur neither in masseter or digastric muscles.
Third condition. During verbal reactions no suppression effect on EMG activity was observed with exteroceptive stimulation even at 200 and 300 ms intervals after the "go" signal.
However the stimulation could negatively interfere: a delay of the ACG latency with respect to the EMG preparatory activity occurred in comparison with the latencies in normal conditions.

SPEECH MOTOR CONTROL - COMPUTERISED ANALYSIS OF ACOUSTICOGRAPHIC AND ELECTROMYOGRAPHIC DATA

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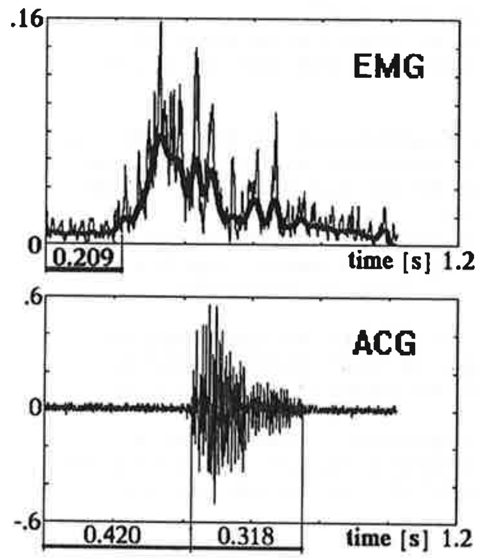
Techniques for simultaneous recording of voice (or Acousticogram, ACG) and oromandibular electromyographic signals are described in a companion paper. Present paper describes some preliminary results obtained from a number of semi-automatic procedures to extract from ACG and EMG signals the basic features which were shown to be of a crucial importance for the diagnosis of neuromuscular and cerebral disorders. Several algorithms are presented for the automatic recognition of absolute and relative latencies of EMG channels; basic manipulations include digital low pass filtering, identification (with linear algorithms), computation of slopes, determination of response onset time. An example of this analysis is shown in the figure below.

ACG signals are analysed with short-time spectral techniques (dynamic spectrograms and formants extraction). In addition, onset time and total duration of the most significant epoch are determined with the same techniques as for the EMG signals.

Strategies and numerical results obtained from these computerized methods are evaluated in relation to the performances of experienced observers, for both normal and pathological subjects.

These data are presently used to develop a first approximation model of the speech motor control.



Top: *analysis of EMG signals. The original EMG signal (thin line) is superimposed to the filtered signal (thick line). The thick horizontal bar gives the onset time as recognized by the automatic procedure (0.209 s in this example). Data in mV; duration of the whole time window is 1.2s.*

Bottom: *analysis of ACG signals. Onset time (0.420s) and overall duration (0.318s) of the ACG signal, during the same test session. Data in mV.*

A NON-INVASIVE METHOD FOR ANALYSING LIPS MOVEMENTS DURING SPEECH PRODUCTION

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A new methodology for studying the motor co-ordination of the speech articulators will be presented. The study is focused on the external (visible) articulators, i.e. the lips and the jaw. The kinematic characteristics of the articulators are determined by means of the ELITE automatic motion analysis system. The characteristics of this system become of paramount importance in the present application, because of the very low masses of the articulators and because of the small field of measurement. ELITE fits the requirements of the study in terms of small passive markers (2 mm plastic hemispheres coated with retro-reflective paper) and flexibility to suit the small field of view (20 cm x 20 cm). Eight markers were fixed to the subject: two on the central points of the upper and lower lips, to the corners of the mouth, in the middle of the chin, on the tip of the nose and on ear lobes. The choice of body landmarks has been dictated by the parameters that we wanted to analyse: lip rounding and opening, jaw opening and protrusion of upper and lower lips. These parameters have focal importance in the studies on the motor strategies used for the speech production. Several theories have been proposed in this field and the presented experimental method allows to check their validity. Another application of the approach is for the speech recognition from pure kinematic data. Several results on both fields will be presented in order to show the reliability and accuracy of the method.

A NEW NON INVASIVE SYSTEM FOR CONDYLE AND TOOTH MOVEMENT ANALYSIS

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Purpose of this work was to examine the interincisive and condylar border movements in normal subjects by means of a new non invasive system in which two TV cameras record, at a sampling rate of 50 Hz, reciprocal position and movements of 9 passive markers strategically applied on a steel frame cemented on the lower incisors and on spectacles. The TV raw signals are fed into a real time movement analyser and processed by a Fast Processor for Shape Recognition and a Central Processor Unit. An AT personal computer and a dedicated software provide for calibration, head movement compensation, three dimensional reconstruction and graphic representation. A geometrical model is employed to reconstruct the condylar movements. The system was applied to 10 normal subjects which were asked to perform three opening, closing, protrusive, retrusive and lateral border movements. The following observations were made: during opening and closing condylar movements are similar but not always identical and closing may run downward or upward respect to the opening path. During protrusion and retrusion movements are more repetitive. During lateral movements the working condyle shifts sideway and backward most of the times while in the balancing condyle an immediate side shift and a progressive side shift are easily recognizable. The system seems to be adequate for minute tooth and condyle movement analysis in normal and dysfunctional subjects. Supported by a grant of the Italian Department of Education.

MASTICATORY FUNCTION ALTERATIONS IN CRANIOMANDIBULAR DISORDERS

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To study the different parametres of the masticatory function in normal conditions and TMJ and/or muscle dysfunction we employed two jaw tracking devices (Elite System and Sirognathograph) and an electromyograph with surface electrodes applied on the belly of the masseter and temporal muscles. A dedicated software provides, at 21 different degrees of jaw opening and closing, for the following mean data: mandibular displacement and velocity, amount of times the maximum jaw closing position is reached, masseter and temporal EMG activity.

12 normal subjects and 18 patients with TMJ and/or muscle disorders were tested. In the normal subjects mastication appeared to be a rythmical event in which chewing cycles show a symmetrical and well balanced distribution of functional loads. During closure two phases can be distinguished: in the first half of closure the muscle contraction pattern is mainly isotonic in nature since velocity and EMG values increase in a roughly parallel way; in the second half of closure muscle contraction is mainly isometric since velocity decreases while EMG activity further increases up to a peak which is rather close to the position of maximum jaw closure.

In the TMJ patients the following observations were made: the cycles distribution becomes asymmetrical with more cycles performed on the lesion side; the position of maximum closure is seldom or never reached; a tendency to reduce or suppress the isometric contraction phase is present. This can be interpreted as an adaptative mechanism to reduce loading on the impaired TMJ.

This phenomenon is not observed in patients suffering from muscle dysfunction only.

This work was supported by a grant of the Italian Department of Education.

OCCLUSAL ADJUSTMENT AND ELECTROMYOGRAPHY: ADVANTAGE OF PARAMETRIC MODELING IN DENTISTRY

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F.Gasmi, H.Poublan (ENSEEIHT Toulouse, France)

Abstract
The aim of this paper is to determine whether parametric modeling techniques applied to surface electromyographic signals from the masticatory muscles can be used by the dentist during the follow up of patients wearing anterior bite plates and after occlusal adjustment.

Material and Method
Autoregressive modeling techniques allow the electromyographic temporal signal to be replaced by a smoothed spectrum which can easily be read and assessed by the dentist.

The data processing equipment used was developed at the "School of Electronics, Electrotechnical Engineering, and Hydraulics" of Toulouse (ENSEEIHT).

Electrical signals of the patient's muscles were recorded by surface electrodes (40 mm diameter silver, silver-chloride surface electrodes employed in cardiology for children).

Ten patients (aged between 20 and 45, eight females and twelve males), with painful spasms of the masticutory muscles and for whom bite plates had been prescribed were enrolled in the study.

Electromyographic recordings were performed in patients, before, during and after wearing the anterior bite plate and after occlusal adjustment.

Results
The observation of the right-and left-autoregressive spectra, prior to the patient wearing a bite plate, showed a discrepancy between the two plots.

When the plate was inserted and properly adjusted the relationship between right and left sides shows a marked tendency to superimposition and a dramatic decrease in signal power.

On the other hand, when the bite plate was not well-adjusted, a right-left disequilibrium increased. In particular, the collapse of power spectra and the improvement of signal symmetry are of paramount importance to highlight the efficacy of the anterior bite plate. Symmetry obtained from the bite plate seems to be preserved after occlusal adjustment.

Conclusion
Electromyography shows that correctly adjusted bite plates worn for a sufficient amount of time improve the efficacy of the masticatory system.

The autoregressive spectrum represents a frequency curve which can easily be read and assessed by the dentist.

The material and techniques presented in this study are an objective means of evaluation of bite plate efficiency with respect to the consequences of occlusal disorders.

ELECTROMYOGRAPHIC ANALYSIS OF THE OMOHYOIDEUS, VENTER SUPERIOR AND VENTER ANTERIOR OF THE DIGASTRICUS

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On looking the literature about the infra-hyoideia musculature, we note that this musculature has been studied very little, as apposed to the supa-hyoidea musculature.

Since we only came across electromyography work on animals as regards the omohyoideus muscle, the purpose of this paper was to research the electromyographic activity of the venter superior of the omohyoideus muscle in conjugated action with the venter anterior of the digastric muscle, in various man mandibular movements.

For this purpose, we used 20 volunteers whose ages varied between 17 and 30 years; the electromyographic registrations (EMG were carried out on a TECA TE-4 electromygraph with two channels. A pair of electrodes were used for each muscle being:

- Two monopolar needle electrodes for yhe venter superior of the omohyoideus muscle.
- One monopolar needle electrode with "BECKMAN" type surface for thee venter anterior of the digastricus muscle.

The mandibular movements analysed were:

1. lowering
2. elevation
3. propulsion
4. retropulsion
5. lateraliity to the right
6. laterality to the left
7. intrusion
8. extrusion
9. retrusion
10. protrusion

By the registrations obtained we observed that:

- In the movement of lowering the mandibule the two muscles showed themselves to be active, this movement being the most marked.
- The muscles showed themselves to be active in the movements in which a component of the lowering of the mandible is associated , being propulsion, laterality to the right, laterality to the left and retrusion.
- In the movements of elevation, retropulsion, extrusion and protrusion of the mandible, the two muscles remained inactive.

RELATIONSHIP BETWEEN OCCLUSION AND POSTURE

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In the last years, the medicine is become tightly specialistic: for any singular problem exists the doctor specialised in a particular medicine sector, that means the loss of the global vision of the body with its relationships. The aim of this work is to identify the characteristics of the problem and the area of the body in which is itself located, starting from a global approach to the system-body.
Our protocol uses a motion analysis system, ELITE, and it submits the subjects to a series of tests in which we modify the proprioceptive sensitivity at the stomatognatic level and at the extra-stomatognatic level using for example bites and soft soles.
The analysis is based on three steps: first we acquire information about the subject in natural condition with a proprioceptive stomatognatic stimulation; then the acquisition is done with an extra-stomatognatic (plantar) stimulation; in the end we construct some indexes evaluated as

Index=(Presence of Stimulus/Absence of Stimulus)*100

in the following fields:

KINEMATIC
Using the two TV-cameras of the system, we identify the spatial position of three planes of the body: head, shoulders and pelvis. Aften we calculate the angles between these planes and the three reference axis in order to define an index called

OLI = Orthostatic Line-Up Index

ELECTROMYOGRAPHY
Always in the three stimulation conditions, we record the activity of the temporals and masseters muscles during a clench test. The computed index is called

CEI = Cranium-Mandibular Electromyographic Index

We record also the activity of the dorsal muscles during an extension movement computing an index called

VEI = Vertebral Electromyographic Index

GROUND REACTION FORCE ANALYSIS
This analysis is based on the Romberg test. Always in the three stimulation conditions the computed indexes are:

RIA = Romberg Index Area
RIL = Romberg Index Length

The comparative analysis of these indexes give an information about the compensation responses that the body give when we modify the postural parameters: the positive or negative responses to these tests give us an information about the birth body area of the eventually disorder and about the characteristic of the problem.

COMPARATIVE STUDY OF TWO TREATMENT METHODS: SPLINT VERSUS PHYSICAL THERAPY FOR TEMPORAL MANDIBULAR PAIN DYSFUNCTION

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Many aspects of temporal mandibular joint pain dysfunction (TMJPD) are subjects of considerable research. There is no consensus among researchers with regard to its nature, diagnosis, aetiology or methods of treatment. Sixteen patients who suffer from this syndrome were chosen to participate in this study. Eight patients were chosen randomly to receive treatment with an occlusal splint. The other eight were chosen to receive physical therapy that included passive mobilization and active exercises. The pain level and maximum mouth opening range were measured for each group before and after the treatment. The physical therapy group showed a significant decrease in total average, pain level ($p \leq 0.020$). The occlusal splint group failed to show any significant decrease in total average pain (p=N.S.). The occlusal splint group also failed to show a significant increase in average maximal opening. In summary, there are still different opinions regarding the syndrome of TMJPD, but a close and full cooperation between the physiotherapist and orthodontist staff is essential for successful and comprehensive treatment.

ELECTROMYOGRAPHIC ANALYSIS OF THE STERNOHYOIDEUS MUSCLE AND THE DIGASTRICUS, VENTER ANTERIOR MUSCLE IN MOVEMENTS OF THE TONGUE AND NECK

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In spite of the Hyoide bone being considered the skeleton of the tongue, the supra hyoideus and mainly the infra hyoideus musculature little have heen studied electromyographically, so much so that even BASMAJIAN & DE LUCA (1985) consider the study of these muscles an open field for electromyographical study. On the other hand, many text books cite that the infra hyodeus musculature may have a participation in the kinesiology of the head. Therefore, we decided to study the Sternohyoideus and the Digastricus venter anterior muscles in moviments of the tongue and neck. The muscles were studied in 20 young volunteers in a TECA TE-4 electromyograph. A pair of electrodes was used for each muscle, one being superficial and the other a monopolar needle, made up from a short, fine "MISE " type needle used in odontological syringes, once the muscles studied are very thin. The following movements were studied, for the tongue: propulsion and retropulsion, laterality to the left and to the right, attempt to place the tip of the tongue on the soft palate, the hard palate and on the floor of the mouth. For the head: flexion and extension, rotation to the right and to the left, inclination to the right and to the left. For the movements of the tongue the muscles were active in all the proposed movements, with the exception of retropulsion and little activity in the attempt to place the tip of the tongue on the floor of the mouth. The hyoglossal membrane and the median fibrous septum are fibrous structures of the tongue wich are fixed below the hyoid muscle. Various muscles of the tongue are connected directly into the hyoid bone, such as: genioglossus muscle, hyoglossus muscle and the inferior longitudinal muscle of the tongue. The stiloglossus muscle is related indirectly through the stiloglossus ligament in which it is partly inserted. And even the transversal muscle fo the tongue, although intrinsic, is related to the hyoid bone through the median fibrous septum. Therefore, any movement in one of these structures results in movement in the other, which provokes an isotonic contraction in the muscles under study. Thus it happens that by palpation of the Hyoide bone we perceive taht its greatest dislocation occurs when an attempt is made to place the tongue on the soft palate coincidentally exactly with the greatest electrical potentials of the muscles under study. The two muscles were inactive in all the movements of the head, which demonstrates that they do not have any relationship with their kinesiology.

BASMAJIAN,J.V. & DE LUCA. Muscles alive: their functions revealed by electromyography. 5.ed. Baltimore, Willians & Wilkins, 1985, p. 469.

ELECTROMYOGRAPHIC ANALYSIS OF THE OMOHYOIDEUS, VENTER SUPERIOR AND DIGASTRICUS VENTER ANTERIOR MUSCLES

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The venter superior of the omohyoideus muscle and the venter anterior of the digastricus were analysed electromuographically, in 20 adult volunteers, in movements of the tongue and head.

The electromyographic registers (EMG) were carried out on a TECA TE-4 electromyograph with two channels. The following movements were analysed:

Movements of the tongue:

1. propulsion
2. laterality to the right
3. laterality to the left
4. retropulsion of the tongue
5. placing the tip of the tongue on the hard palate
6. placing the tip of the tongue on the soft palate .
7. placing the tip of the tongue on the floor of the mouth.

Movements of the head:

1. flexion
2. extension
3. rotation to the right
4. rotation to the left
5. inclination to the right
6. inclination to the left

By the registrations obtained we observed:

- In the cinesiology of the tongue, the muscles participated actively.
- In the movements of propulsion, laterality to the right, laterality to the left, placing of the tip of the tongue on the hard and soft palates, placing of the tip of the tongue on the floor of the mouth, the two muscles acted simultaneously.
- The two muscles show themselves to be inactive in the movements of retropulsion.
- In the cinesiology of the head, the two muscles venter superior of the omohyoideus, and venter anterior of the digastricus muscle did not show any activity.

ELECTROMYOGRAPHIC STUDY OF THE STERNOHYOIDEUS AND DIGASTRICUS, VENTER ANTERIOR MUSCLES IN MANDIBULAR MOVEMENTS

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Although electromyographic studies about sketal muscles are increasing, and each muscle or group of grooved muscles have hadtheir actions revised, the infrahydoideus muscles still have not been extensively studied by electromyography, so much so, that BASMAJIAN & DE LUCA (1985) consider the study of these muscles to be a field open to electromyographic investigation. Many text books such as Gray(1979) among others, consider a joint action between the supra hyoidea and infra-hyoidea musculature. The contraction of the sterno hyoideus muscle would fix the hyoide bone providing a fix plataform for the action of the supra- hyoideus muscles, particularly the venter anterior of the Digastric Muscle, lowering the mandible. Thus, owing to these facts, we decided to study the joint action of the Sternohyoideus and the Digastricus venter anterior muscles in mandibular movements. The muscles were studied in 20 young volunteers in a TECA TE-4 electromyograph. A pair of electrodes was used for each muscle, one being superficial and the other of the monopolar needle type "MISE", being a short, thin needle used in orthodontical syringes, since the muscles are very thin. The following pairs of mandibularmuscles were studied: lowering and elevation, propulsion and retropulsion, laterality to the left and to the right. The Sternohyoideus and Digastricus venter anterior muscles in the movement of lowering the mandible showed great action potential which was also observed in the propulsion and laterality to the left and to the right, being inactive in the remainder of the movements. From this, it is concluded that the muscles under study present action potentials in all mandibular movements in which there is a component of lowering of the mandible. This fact bears out the conclusions of authors such as: CARLSOO(1950), STEPOVICH(1965), ALHGREN et al (1978) and PANCHERZ et al (1986) who report movements of the Hyoide bone during mandibular movements, although they disagree among thenselves as regards the direction of these movements. Although, there is a contraction of the muscles under study, there is a movement of the Hyoide bone, which is only permitted by an isotonic contraction and not an isometric one, of this musculature. These results disagree with the test books, which attribute a fixation of the hyoide bone to the Sternohyoideus muscle for the action of the supra hyoideus musculature.

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STUDIES OF HEAD AND BODY MICRO MOVEMENTS USING FLAT SENSORS DEVICE

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The head micromovements sensed by physiologists and osteopathes, were never objectively measured on the human being. They could have two origins : the cranio-sacrum mobility or the arteriolar vasomotion [1],[2].

We have realized a device which allows frequency measurement of micromovements on a level with the head but also another part of the human body [3]. This device can detect very small amplitude periodic movements (1µm and less). The measurements must be neither invasive nor constraining for the patient. The system uses several flat eddy currents proximity sensors [4]. They are include in oscillator which frequencies are function of a target sensor distance. The acquisition system realizes sets of 562 frequency measurements every 250 ms. The treatment system works out digital filtering. The spectrum is calculated by Fast Fourier Transform or autoregressive (AR) model.

Every sensor must be placed in front of metallic target which fits perfecty with the patients's skin ; we use a thin sheet of aluminium paper of about 5 centimeters square applied with paste on the patient's skin.

At first, we have utilized this device to study the head micromovements on or several sensors are placed on a level with the brow of the patient. Another sensor can be placed where the respiratory movements are important, such as a level with the clavicle ; it allows to eliminate the breathing movement on the spectra.

A set of about hundred tests on healthy subjects shows that a periodic micro movement of an mean frequency of 9,7 cycles a minute is detected. By the presence of the same periodic micro movement on a level with the hand, it seems that the origin of these movements is the arteriolar vasomotion rather than the cranio-sacrum mobility.

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A BIOMECHANICAL STUDY OF THE FORCES EXERTED BY THE FACIAL ORTHOPEDIC MASK (DELAIRE)

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The effects produced by maxillary protraction appliances on the craniofacial complex have been described in several experimental and clinical studies. Although several aspects of the biomechanics of the maxillary protraction are now reasonably well understood, there are factors which remain unclear. One is the direction and the magnitude of the forces exerted by the orthopedic mask on the chin and on the forehead. Consequently, the relative role of these forces during the orthopedic treatment of a given patient is obscure, in particular for the force on the chin.

On a more practical basis, the entire process of fitting the mask to a given individual is done empirically, using qualitative guidelines concerning the position and the geometry of the mask relative to the subject's head. This study presents a model for the computation of the forces exerted on the chin and on the forehead by the facial orthopedic mask (Delaire) in skeletal Class III malocclusions. Cephalometric data as well as geometry of the mask (see figures below) are taken into account to simulate in quantitative terms the entire approach. A computer program (Grandori et al., *American J. Orthodont. Dentofac. Orthop.*, 1992, vol. 104, pp. 140-148) has been implemented to validate the model on a group of six patients. Despite the approximations about the mechanical characteristics of the appliance and of the constraints (rigid body, ideal constraints) and despite the unavoidable errors in the estimation of the geometrical parameters (dimensions and angles), it is shown that the computation of the forces (in orientation and strength) at the forehead and at the chin is possible. Some practical applications of the model are presented. These data are discussed also in relation to the use of electrical stimulation to decrease the duration of the treatment.

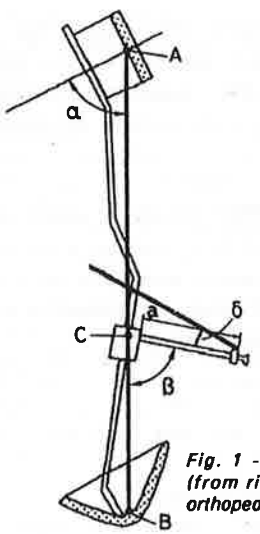

Fig. 1 - Schematic view (from right) of the facial orthopedic mask (Delaire).

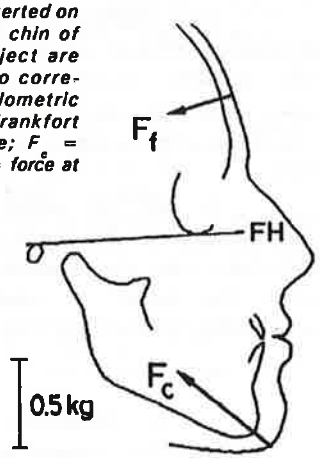

Fig. 2 - Forces exerted on forehead and on chin of one typical subject are superimposed to corresponding cephalometric tracing. FH = Frankfort horizontal plane; F_c = force at chin; F_f = force at front.

RESTORATION OF FUNCTIONAL GRASP-RELEASE IN TETRAPLEGIA WITH IMPLANTABLE NEUROMUSCULAR STIMULATION AND TENDON TRANSFER

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Abstract. An implantable system has been developed to restore functional grasp and release in the C5 and C6 tetraplegic patient through neuromuscular stimulation of the paralyzed extremities. The system provides control of both lateral and palmar prehension and release by electrical activation of the paralyzed forearm and hand muscles. The neuroprosthetic system uses a surgically implanted multichannel stimulator for activation of the muscles. The patient proportionately controls the position of the digits and grasp force by voluntary movement of either the shoulder or the wrist. The implanted stimulator is both controlled and powered using radio signals generated by an external portable controller. Tendon transfers are used to augment function by moving the insertion of muscles which are electrically excitable to provide enhanced functionality.

1. Introduction

Individuals who are paralyzed due to spinal cord injury have retained electrical excitability of many of the muscles of the extremities. This supplies a potential source of energy for restoration of functional movement, by electrically exciting the innervating motor neurons in sequences which can elicit controlled movements.

In the case of mid- to high level cervical level injury to the spinal cord, central control is often lost to the muscles of the forearm and hand, although peripheral innervation is retained in many of these same muscles. Many of the individuals with this level of injury retain volitional, albeit weakened, control of the proximal musculature and joints, but are unable to utilize the extremity unless encumbered by extensive orthotic devices. Our work [1], and that of others [2,3], has sought to elicit controlled motions in these paralyzed muscles, and thus restore functional control in the distal limb for restored function. The objective is to provide the individual with sufficient control of the distal extremity that he/she is able to perform most activities of daily living such as eating, grooming, and office tasks independently, with minimal orthotic support, and without the assistance of an attendant.

2. Discussion

2.1 *Implant System Description*

The approach that we have taken is to electrically excite the muscles with indwelling electrodes, implanted near the nerves innervating the paralyzed muscles. The delivery system has evolved over the period of development from a percutaneous approach to utilizing a fully implantable stimulator-receiver. Our clinical experience, gained over 15 years in our own center and up to 5 years in four collaborating centers [4], has demonstrated the viability of the percutaneous approach, but also confirmed limitations which primarily originate from the need for maintenance of the implant site and unacceptability of the external lead wires which traverse along the arm to the electrode terminals on the distal forearm. Consequently, we elected to implant the stimulator-receiver unit in the pectoral area, and subcutaneously tunnel leads to electrodes placed near the motor points of the paralyzed muscles. Because the size and flexibility constraints associated with the percutaneous leads were now eliminated, we were able to design and construct a new lead and electrode system. The lead uses a closed helix design, in which multi-filament lead wires are helically coiled and inserted in an encasing tube. Many potential terminations of the lead are possible, including electrodes in the muscle or applied to the nerve. To date, we have elected to use only the former type with the clinically implanted system, although extensive acute and chronic testing of the others is underway. The epimysial electrode generally is placed on the superficial surface of the muscle at the proximal one-third junction of the length. Its advantages are its overall robustness, including the ease of installation and replacement, relative insensitivity in contractile response to the precise placement, and absence of adverse biological response when used on a variety of muscles over a range of stimulus levels and movements.

The implanted receiver-stimulator unit is an eight channel device which is powered and controlled by a 7 MHz carrier [5]. Each channel is individually addressable and generates monopolar current regulated charge balanced biphasic stimulus with the maximum parameters of the cathodal pulse of 20 ma, 200 usec at 50 Hz. The electronics is hermetically enclosed in a titanium case, which serves as the common return electrode.

2.2 *Surgical Implantation*

Surgically, incisions are made over the muscles to be stimulated, the muscle is localized, electrode sites are identified by probing and stimulating,and the electrode is sutured to the muscle surface [6]. Acceptable response can be measured intraoperatively [7,8], or qualitatively assessed. The acceptance is based on selective and graded response eliciting a MRC grade 4 contraction without substantial change in contraction strength due to internal shortening-lengthening of the muscle (ie. "length dependent contraction"). The lead wires are then tunnelled subcutaneously to a mid-humeral site. Generally seven muscles are implanted, allowing one channel to be used for electrotactile sensory feedback [9]. This electrode is placed in the supraclavicular sensate region with the conducting side toward the skin. It is used to provide state feedback such as on/off, mode changes, and lock/hold/regain control. Briefly, the stimulus amplitude and pulse width are set and fixed at a discriminable level. All coding is with frequency cues, with frequencies separated by two JND's for reliable identification. The most complex (but still simple) frequency coding provides frequency changes at 20% increments in the input command. For example, in active control when between 0 to 20% command the stimulus is applied at 4 Hz; from 20 to 40% command the stimulus changes to 6 Hz, which is a two JND step. Eliciting a lock command removes the cutaneous feedback. This code is not optimized, but has proven to be simply recognized and easily learned. It replaces an auditory feedback code used previously, and is reported by the subjects to be desirable because it is intimate and it neither draws attention or is obscured by environmental noise.

The implant is placed in the pectoral region with the return electrode facing toward the skin, and its lead wires are in turn tunnelled to the humeral site. Junction is made at this point between leads and receiver [10]. This site allows subsequent replacement of any failed component, or for future upgrades in the system.

2.3 *Surgical Alterations in Biomechanics*

One issue which has been identified from patients in our program and the multi-center evaluation is that key muscles may be denervated as a result of the spinal cord injury. The muscles affected generally are those which receive their innervation at or just caudal to the site of lesion. The denervated muscle is unexcitable with the stimulus parameters available. We have elected to use conventional hand surgical procedures to transfer the insertion of electrically excitable muscles to provide the functions lost by paralysis and denervation. For example, if the finger extensors are denervated but the extensor carpi ulnaris (ECU) is innervated, it may be transferred to EDC and electrode appropriately placed on ECU elicit finger extension. The selection of muscles to be implanted is thus based on which are under volitional control, which are electrically excitable, and which functions are required. Since the latter is defined by the grasp configurations desired, the allocation of channels is between muscles that have intact peripheral innervation, those that require transfer, and which transfers are feasible to perform with a good probability for success.

2.4 *Selection of Muscles and Movement Synthesis*

The objective generally is to provide both lateral and palmar prehension and release. This requires, at a minimum, six channels for finger and thumb flexion and extension and thumb ab-adduction. Finger flexion is provided by two muscle groups, thus allocating a seventh channel. Since the eight channel provides sensory feedback, all channels are utilized. If one or more muscle functions is absent due to denervaton, that channel is allocated to a muscle for transfer to provide the substitute function.

The grasp template specifies the orderly phasing of muscle activation to achieve the desired movement and is described elsewhere [11]. This uses graded activation of each muscle, in response to a single graded input command. The resulting movements for both lateral and palmar grasp are proportional to the graded command, with control provided for both digital position and grasp strength.

2.5 *Command Control*

The command signal employed primarily is from the contralateral shoulder. As described by Johnson [12], combinations of elevation-depression and protraction-retraction are available to supply both proportional and logical signals. While this is the most commonly employed command site, wrist extension has also been employed if the individual retains this movement voluntarily. This site is particularly liked by subjects. At this point in development, both require use of an externally placed sensor. For the wrist, external placement presents particular issues such as exposed lead wires which are similar to those experienced with percutaneous leads, and thus internalization of the sensor system is felt to be a priority for clinical acceptance of the command control site.

2.6 *Clinical Deployment*

The implanted system has been implemented in three subjects, the longest being implanted for five years and the others for nearly one year. No surgical complications have been experienced. After two years in the first subject, the original receiver unit was replaced because of increased power usage (subsequently determined to be caused by a high impedance antenna weld), but all leads have remained functional. Physiologic measures of individual muscle force and grasp strength have been performed. The results show that the force vectors generated by the transferred and stimulated muscles effectively substitute for denervation, and that functional levels of grasp and pinch strength are developed. The functional results are expressed by quantitative tests which measure overall system performance. They are (a) repeated acquisition, movement, and release of standardized objects of graduate size and weight, and (b) acquiring common objects of daily living. Functions previously unattainable are achieved by both measures with the neuroprosthesis.

3. Conclusions

The clinical deployment of a complex neuroprosthesis is a complex process. It involves engineering technology, surgical experience, and clinical judgement. These can best be assembled by a multi-disciplinary team with access to the full dimensions of the problem. In the case of the deployment of the upper extremity system for hand control, these elements have been brought together to provide a first generation motor prosthesis that enables cervical level spinal injury patients to regain functional use of their paralyzed hand. The system has been in clinical use outside of the hospital with percutaneous electrodes for over ten years and in four satellite sites for as long as four years. The implanted system, which replaces the percutaneous leads with an implanted receiver, has been in clinical usee for as long as six years on a daily basis. The system requires minimal maintenance and technical support. Undoubtedly and hopefully, this

first generation system which we have introduced will be improved upon by ourselves and others, with the result that advanced electrodes, stimulators, control strategies, and so forth, will improve the clinical functionality. In the meantime, this basic system is to undergo multi-center testing to introduce it into clinical usage. Through this process, we hope to provide a substantial number of disabled users with improved capabilities for independent function and an enhanced quality of life, as has been enjoyed by the relatively few users who presently have access to the neuroprosthesis.

4. Acknowledgement

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FES AND BALANCE CONTROL

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Functional electrical stimulation FES has developed within the last decades into an important therapeutic modality for locomotion rehabilitation in stroke and spinal cord injured SCI patients. In many cases FES needs to be applied chronically like in patients with almost complete or incomplete SCI /1/. This last category of patients is while walking by means of FES using the upper body - non plegic part for voluntary controlling and inefficient maintaining of balance due to the total proprioception and exteroception loss from the plegic body part. They are relaying on and utilizing the visual and balance sensation but also incorporating the sensation provided across crutches and hands and the minimal left rudimentary sensation. Because of missing knowledge and technology we have not been able to provide by means of FES an active balance assistance yet. Therefore, and also due to motor inputs provided by patients arms across the crutches, the resulting gait is functionally closer to a four point crutch supported walking than to a normal biped gait. From a biped walking gait point of view, the active balancing abilities of a completely injured SCI subject are drastically reduced and therefore he has to relay on supporting and equilibrium provided across forces exerted by his hands. Owing to it the FES enabled "biped gait" in SCI patients is consequently - and by definition a quadrupedal locomotion with the characteristics of static stability. This presentation discusses the drawbacks of quadrupedal static stability gait modes which are limiting the FES enabled gait in SCI patients and resulting in a gait mode which is very different in essential characteristics from a normal biped gait /2/. These important facts have so far not been objectively incorporated into the design of FES systems for SCI patients. It is more than evident that the current FES systems and also those developed in the next decades, will not be able to provide equilibrium and thus hands free walking. Therefore for the enhancement of FES systems for gait in SCI patients the incorporation of logistics and control of four points supported body is vital and promising. The key for progress is to substitute the statically stable weight transfer phase for statically unstable but dynamically stable body weight transition phase which has to be included between two statically stable postures being necessary for balance control and equilibrium recuperation. Such walking pattern is enabling periodic potential into kinetic and vice versa exchange of energy but requires also active FES means for restoring of the lost energy in gait cycle. This energy can be obtained by FES providing propulsion (push-off) or lifting of energy of the body. We are experimenting with such gait modes and studying the necessary FES sequences, patient control and training. At the end of this presentation a discussion is provided highlighting advantages and disadvantages of incorporating FES joint closed loop FES versus patient closed feedback loop. It is shown that the here explained philosophy of FES gait enhancement in SCI patients is opening an entire new area of research, modifying the existing patient training methodology but also triggering the development of new FES software with hardware solutions.

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CONTROLLING THE MAGNITUDE OF HIP FLEXION USING THE FLEXION WITHDRAWAL RESPONSE

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Background
The flexion withdrawal response has been the main method of producing hip flexion for FES-assisted gait using surface electrodes. If this response is to be used in a practical system to provide ground clearance during gait and/or stair climbing, then the patient must be given the means to control the magnitude of the response. This will allow him/her to negotiate obstacles, compensate for the effects of habituation and climb steps of various heights.
Method
A method of controlling the magnitude of the hip angle elicited by the flexion withdrawal response was studied. The response was elicited using surface electrodes and hip angle was monitored using a flexible goniometer (Penny & Giles, Gwent).
Two targets, representing desired hip angles, were displayed on a screen in front of the patient, and the magnitude of the actual hip angle achieved was represented by a vertical bar. Two methods for controlling the magnitude of flexion were tested. These were 1) the control of pulse width of the stimulation using a force transducer and 2) control of time of activation of the stimulation using an on/off switch. These controllers were compared to the use of fixed stimulation levels. The absolute error between the hip angle obtained and the target angle was used as the performance criterion.
Results
It was found that maximum hip angle obtained could be controlled by stimulus parameters of pulse width, frequency and duration of stimulation. The errors using the patient controller were significantly less ($p < 0.001$) than using pre-defined stimulation levels. The lower target was more difficult to match than the upper target with both controllers.
Conclusions
Pulse width, frequency and duration of stimulation train can all be used to control the magnitude of the flexion withdrawal response. The magnitude of the flexion withdrawal response can be controlled by the patient which could enable variable height obstacle clearance during gait and single step negotiation. This, in conjunction with methods for dishabituating the response, will extend its applications.

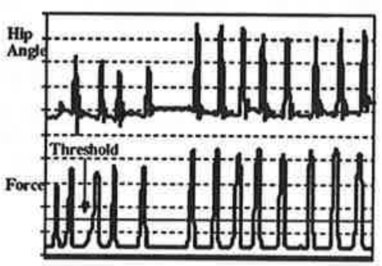

Figure A

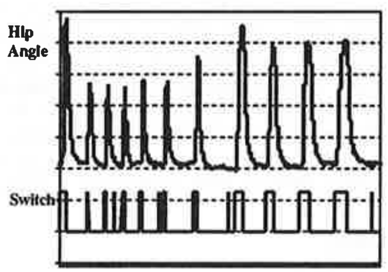

Figure B

Figure A shows the relationship between force applied to the transducer and hip angle obtained. Figure B shows the relationship between time switch is ON and hip angle obtained.

THE NONLINEARITY OF MUSCLE RESPONSE WITH TIME EFFECT OF BLOOD SUPPLY ON FATIGUE RESISTANCE IN PARAPLEGICS WITH SURFACE F.E.S.

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Muscle fatigue using surface electrical stimulation represents a highly reproducible phenomenon in spinal cord injured patients. We recorded the torque output of the quadriceps muscle electrically stimulated as a function of time using a force transducer. We used a two channels current regulated stimulator. An ammeter, serially mounted, was specially developped for the test to be sure that torque decrease was due to fatigue and not to current decrease. The curves obtained with a duration of 120 seconds belonged to sigmoïd curves. Three main parts were described; a first plateau of short duration, followed by a slope more or less steep and then a second plateau maintained for long time. We modelized this phenomenon using three different equations (one reffering to inverse trigonometry and two using exponential) in two populations; a group of patients composed of 11 paraplegics, with a level of injury comprised between T4 and T11, and a group of 10 able-bodied subjects without regularsport practice. The computed coefficients of determination, r2, were of very high values, always better than 0.99 for the check sample and 0.999 for the group of patients. Therefore, the predictions of the model were excellent. Muscle output was not different between populations during the first phase, i.e., the firqt plateau but, since 30 seconds was lower in paraplegics. The residual torque after 120 seconds was representing 20% of the maximum torque in patients, while it was around 60% in the able-bodied group. We also studied the effect of blood supply in the generation of fatigue, repeating, for the two populations the same tests using a tourniquet. This test permitted to determine the different metabolic phases and therefore the recruitment of the different populations of muscle fibers within the quadriceps. The first plateau represented the output of FF fibers, non affected by the deprivation of blood supply. We did not find diffrence between groups. The main slope was the reflect of the FR fibers output, partially affected by the tourniquet. This slope was of same value for both groups. The second plateau highlighted the residual torque and, thus, the tension developped by the S fibers, whose output was dependent to blood supply. It appeared a lack of type S and FR fibers in patients, certainly due to non use. This plateau was always higher in the able-bodied group. In some patients, the curves obtained without and with tourniquet did exhibited difference, signing the poor effect of tourniquet on torque output in this sub-population. Fatigue appeared as essentially due to the recruitment mode, which is inverted from normal physiology, and was not dependent on vascular phenomenon in paraplegics.

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A FIRST APPROACH TO RGO PARAPLEGIC WALKING

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Reciprocating gait orthosis (RGO) has been recently introduced in our country. This pilot study has been designed in order to: a) ascertain the feasibility of restoring standing and walking capabilities in paraplegic by means of the RGO device; b) evaluate the advantages of the use of functional electrical stimulation (FES) in association with RGO and as a tool to recover muscular trophism; c) study some kinetic and dynamic aspects of RGO walking. Six paraplegic patients, with levelsof spinal lesion ranging from C6 to T12, participated in the research. In three patients the use of the RGO was preceded by a three month period of physiokinetic rehabilitation using FES. In two subjects forces transmitted by the foot on the floor and by the hands on the deambulator were measured together with leg trajectories. Muscle activation patterns in the trunk and upper limbs were obtained on EMG tracing. FES training has been demonstrated to be very effective in restoring force and trophism in quadriceps femuris and hamstring muscles. Furthermore all patients reported a beneficial effect of FES in reducing muscle spasms. When no FES of paretic muscles is used, and the patient can only rely on the work done on the ambulator to move the legs, the forces applied on it still did not exceed 25% of body weight. In fact, since the head, arms and trunk are constantly projected forward, the body center of gravity is placed anteriorly with respect to the supporting foot: in this way the body weight operates the torque which moves the legs. The RGO is fixed at the knee and ankle, thus obliging the user to a compass gait. Therefore, major work involving the shoulders and arm muscles is required in the lateral tilting of the body which is necessary to avoid the swinging foot hitting the floor. The prime actuators of this work, as is noted in EMG tracings, are the triceps, latissimus dorsi and trapezius muscles. The flexores digitorum continuously contract for grasping the ambulator. The efficiency of the body tilting is very different in patients depending on the level of spine lesion. In low spinal injuries, when control of all trunk muscles is retained, the task is easily accomplished. In upper thoracic patients the movement becomes fatiguing, and even worse in cervical patients. In the latter the lack of a firm grip of the hand further complicates the RGO applicability. In this case the advantages of using FES were clearly evident. The main objective of future RGO improvement shall concentrate on reducing the work needed for tilting the body. In addition, correct physiotherapy should aim to strengthen the above mentioned muscles.

CONTROL OF FES BY NEUROPROTHETIC NETWORKS: A COMPUTER SIMULATION STUDY

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Introduction The rèstoration of complex functions such as free stance via functional electrical stimulation (FES) is difficult to achieve. This is partially due to the related control problems such as nonlinearity, stochastic variations, and non-stationarity. Therefore, we decided to investigate the use of artificial neural networks as a part of a feedback control-loop. As a first approach we tried to manage the control of free stance and simple body movements in computer simulations.

Patient model The biomechanical patient model is planar and consists of 3 limbs and 6 muscles (4 mono-, 2 biarticular) [1]. The outputs $\boldsymbol{u}$ of the controller are the current amplitudes of 6 stimulators, which are transformed into a motion of the model via the nonlinear dynamics of muscle physiology and body mechanics. Finally, a set of sensory data is communicated back to the controller.

Neural Networks Three layer artificial neural networks were applied for system identification (forward dynamics) as well as control (inverse dynamics) [2]. The special design of our networks yielded a combination of standard feedforward network and radial basis function behaviour. This allowed for a fast deterministic initialization of synaptic weights, followed by iterative learning steps for fine-tuning.

Results Our computer simulations demonstrate that the proposed feedback-loop with integrated neural networks was capable of stable control of the model system. Moreover, by a repetition of the training procedure the size of the working area could be enlarged and the response dynamics were improved. The figures below show the response to a disturbance of the initial state for two networks obtained from the first two cycles of the training procedure. It can be concluded that regions of the model's state space can be controlled where no valid linearization of the plant is possible, a fact that severely limits the use of non-adaptive linear PID-controllers.

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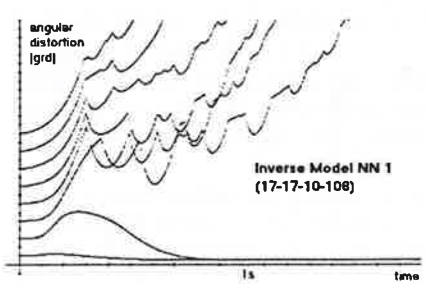

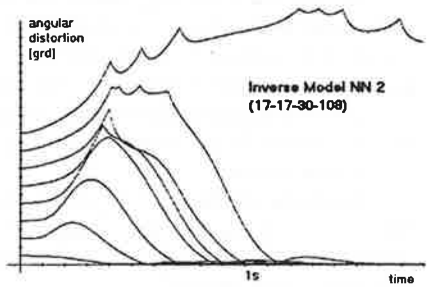

RECRUITMENT OF INTRANEURAL STIMULATION, USING MULTIPOLAR ELECTRODE COMBINATIONS

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A simulation study on the recruitment behaviour of intraneural multi-electrode devices for selective nerve stimulation is presented. In the low force range, for monopolar stimulation, a cubic dependence between the muscle twitch force and the stimulation current can be derived, (F ~ I^3). Recruitment curves are presented as log-log plots of this F-I relation, and the power of I is called the steepness P of these curves. So for monopolar stimulation P equals 3. For tripolar stimulation lower P values are to be expected. Experimentally obtained recruitment curves however are often found to be steeper. Steepness values of 6-9 are often found. It was investigated wether this effect might be caused by clustering of nerve fibres in sub bundles in the fascicle.

Simulations are done for the α-motor fibres of the EDL muscle in the peroneal nerve of a rat, with the fibres distributed randomly throughout the nerve, or clustered in a sub bundle. The EDL has only about 40 α-motor fibres, and therefore the density of these motor fibres is low. The recruitment curves are fitted to a straight line with an upper fitting force limit of 15 grams. In fig. 1 a typical monopolar recruitment curve is shown for one realisation of the fibre distribution, in case of no clustering.

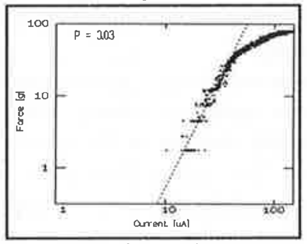

fig. 1. A typical monopolar recruitment curve.

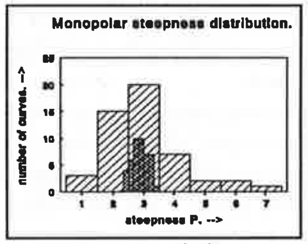

fig. 2. Histogram of the steepness distribution.

From a study over fifty realisations, the next conclusions can be drawn. A considerable spread is found in the steepness distribution, for both monopolar and tripolar stimulation, and also for clustering and no clustering of motor fibres. Because of the low density the actual fibre positions strongly influences the steepness found for individual recruitment curves. In fig. 2 the bright histogram shows the results for monopolar stimulation in case of no clustering. The steepness values vary from 1 to 7 with a mean of 3. For high fibre densities, (2000 fibres), this effect is much less, as shown by the inserted histogram, which is narrowly centred around the theoretical value of 3. For tripolar stimulation, (not shown), the steepness values vary from 1 to 3 with a mean of 1.4.

In case of clustering, when the electrodes are placed outside the cluster, for both monopolar and tripolar stimulation an increase of the steepness of recruitment curves is found. For monopolar stimulation the steepness increases from 3 at the edge of the cluster to almost 7 far from the cluster. For tripolar stimulation the steepness increases from about 1.4 to almost 5. Monopolar curves remain steeper than tripolar ones. For tripolar stimulation the orientation of the tripole with respect to the cluster also influences the steepness of recruitment curves.

When the electrodes are placed outside the cluster, log-log recruitment curves are no longer linear in the low force range. In that case the steepness value found is depending on the range over which is fitted, and one has to be careful with the interpretation of the steepness value obtained.

ELECTRIC CURRENT DISTRIBUTION DURING INTERFERENTIAL THERAPY

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By means of computer calculations, this paper offers a better insight into the electric current distribution during quadrupolar interferential therapy, and it points out the particular possibilities to perform transcutaneous stimulation of muscle fibers which are situated far beneath the skin.

In a 2-dimensional structure, currents from the two separated circuits do not flow in the same direction (mostly perpendicular) and the currents have to be added just as vectors. As a consequence :

1. in phase, the amplitude is maximally $\sqrt{2}$ times the original one, and the direction is given by the vectorial composition of the two currents
2. destructive interference is impossible : when the currents are in counter phase, the resulting current is not zero, but is given by the vectorial composition
3. in quadrature, there exists no single moment at which the total current equals zero, and with optimal interference conditions, a constant magnitude is found with a rotating direction.

It is because of the absence of destructive interference and the fact that the magnitude remains practically constant in the periods of quadrature that the special stimulating properties occur.

Using a computer program for solving 2D problems the electric current densities in a thigh cross section have been calculated [1].

The current density (without taking into account the direction of the current) during one period is displayed in two particular points inside the cross section (see figure). For pure rotating currents, the density is displayed as a constant value.

In point A (Vastus Lateralis muscle) a quit long period of non zero current is found (good interference).
In point B (between Femur and fat tissue) a high intensity is found due to the lateral-medial current, but a few interference occurs due to the low dorsal-ventral current density.

The rotating current is of particular interest because it is now possible to obtain a nearly constant amplitude (without zero's) without skin load (AC). Therefore this 2D interference is more powerful than an ordinary AC interferential current with amplitude changing from 0 to 2 × A.

At this moment the theoretical results have only been verified on normal muscles. From this it becomes clear that during the periods of quadrature, the person feels the constant intensity by a tetaneous contraction of internal situated muscles.

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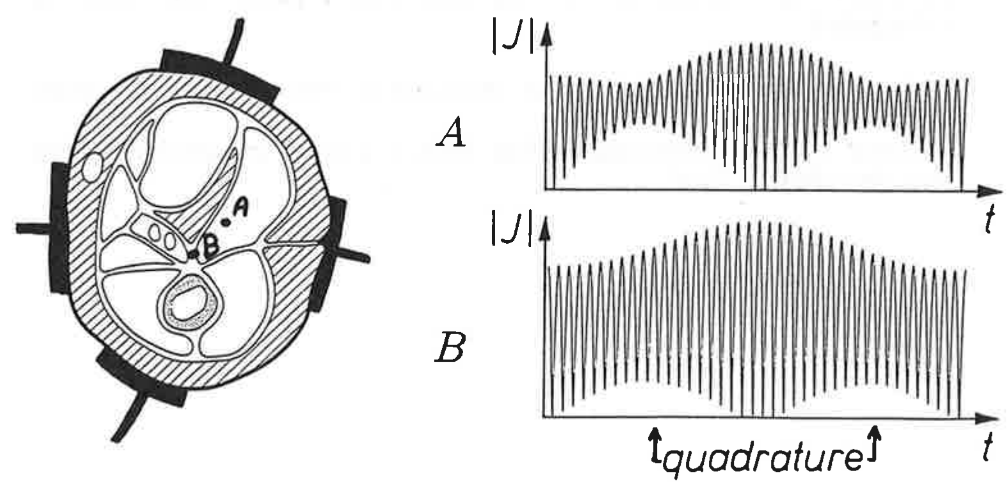

EFFECTIVENESS OF ELECTROMYOSTIMULATION IN THE EARLY REHABILITATION AFTER TOTAL HIP REPLACEMENT

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Introduction: Muscular atrophy, which is usually present in serious coxarthrosis, greatly affect the result of the intervention of total hip replacement. The aim of this work was to study the effectiveness of muscular electrical stimulation in the early post-operative period to increase muscular strength and the mobility of the operated joint.

Methods: Eighty patients entered the study. All the patients, affected by coxarthrosis, underwent surgical intervention for the implantation of a non-cemented total hip arthroprosthesis. Patients were randomly divided into two groups. A standard phyisiokynesitherapic protocol, including isometric voluntary contractions and passive and active mobilization, was administered to the first group of patients (the control group); in the second group of patients this treatment was supplemented by elecrical stimulation. Surface electrical stimulation (40 to 70 mA; 2.500 Hz sinusoidal current, modulated at 50 Hz; 10 sec. stimulation followed by 50 sec. rest) was administered starting 3-5 days after surgery for 30 days, 30 minutes per day. Muscular groups stimulated were in all cases: hip adductors, abductors, extensors and flexors. Parameters evaluated were pre- and post-operative maximum voluntary isometric torque and passive and active mobility of the hip.

Results and discussion: A significant ($p < 0.05$ Student's t test) increase in maximum voluntary isometric torque and in active joint mobility was observed in the treated group when compared with the untreated one.
Electromyostimulation is a safe and easy method to achieve strength and movement improvements in the early rehabilitative period after total hip replacement.

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FES SYSTEM FOR SELF-ACTIVATION: AN ELECTRICAL STIMULATOR AND INSTRUMENTED WALKER

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Independency in performing daily function is the main goal in patients following spinal cord injury, Differents methods in the field of rehabilitation are aimed to permit the paraplegic some degree of independence. One of these methods is tthe use in functional electrical stimulation (FES) to the muscles of the lower limbs. In this work, we describe a self developed system for the functional activation of spinal cord injured patients. The system combines a micro computer.-controlled electrical stimulator and an instrumented walker. It allows the following functions: strengthening exercises for the muscles of the lower limbs; standing up form the sitting position; maintaining the standing position; and, performing reciprocate gait while using a specially adapted walker. These function can be accomplished by the user himself, without any external assistance. The system components and method of function are described.

INTERFACE CONTROL OF THE PORTABLE FES SYSTEM IN THE UPPER EXTREMITY

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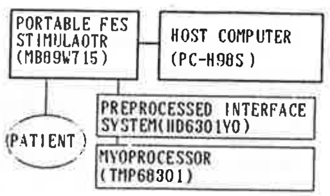
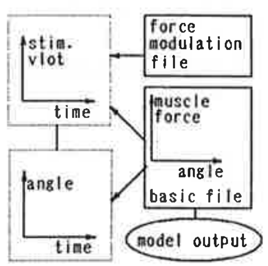
INTRODUCTION: This paper discuss with a multichannel FES system for the upper extremity in the severe upper motor neuron paralysis such as the C4,5 quadriplegia.
From a viewpoint of the man-machine interface, a new portable stimulator(maximum;48ch) has been developing, which has the functions of force and angular control, in a clinical use, by using the stimulating pattern estimated on the musculoskeletal system. Also some of the switch controls and the assistive devices has been revised to be adopted for the patients individually.
MULTICHANNNEL FES SYSTEM: A multichannel portable stimulator is supported by (1)the hostcomputer, and (2)the preprocessed interface system. The one mainly generates an electrical stimulating pattern. The other handles (1) the mechanical input switches, (2)the biofeedback signal made up of the voluntary EMG and the elctrical response of the stimulated muscle, etc. Concerning the electrical specifications, The output, whose electrodesare percutaneous, is voltage-controlled, andhas a rectangular wave with 200 micro sec pulse width.



MULTICHANNEL FES SYSTEM

STIMULATING VOLTAGE PATTERN: The composite motion of the upper extremities consist of 5 basic movements of shoulder, cubital, radioulnar joints, and the hand. Estimation of the stimulating pattern is based on the integrated EMG signals with sitting on the electric wheelchair. But, on the elbow and wrist jionts, a revised musculoskeletal model has been applied for estimating the muscular forces. Its basic data structure to create the stimulating patternis shown in the Figure. We point out that the musculoskeletal model has been induced to make a more flexible and self-learnig system for the stimulator.
[PATIENT]-MACHINE INTERFACE SYSTEM: The interface between the FES system and its user should be built to make the most of the remaining voluntary movement, and to be useful for the environmental condition of the patients in daily living. Usually the shoulder joint or head movements are used by the mechanical, blower switches, etc. These switches are adopted by its simple implementation in a ordinary therapeutic course.

Here, the preprocessed interface system possesses (1)a flexible and selective function of the number of switches, and its operating order, by using the operating table file, and (2) the display to monitor the switch-operating condition, and the biofeedback signals(M wave) of the stimulated muscles. Also it can access to the other processors, such as theelectromyographical signal.
ASSISTIVE DEVICES: The patients seldom have a functional movement except the shoulder joint. They are usually in bed, and use a electric wheelchair. Here,a reviced balanced forearm orthosis is introduced to assist the movements of the shoulder and elbow joints. Also a seating device is useful to sit up in electric wheel chair.



DATA FILE STRUCTURE

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METHODS FOR ANALYZING ELECTROMYOGRAPHIC PATTERNS IN HUMAN GAIT

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Electromyography (EMG) is an essential component of the evaluative process for normal and pathological human gait. Because of this several quantitative methodologies have been developed for analysis utilizing the linear envelope (LE) as the mode of representing the EMG. These methodologies perform two different procedures. One allows the comparison of populations of LE and determines if any significant differences occur between the populations. The other methodology subdivides a pooled population of LE into groups of different patterns.

1. INTRODUCTION

Studies for investigating the characteristics of abnormal gait and the effect of treatment programs have been conducted in many institutions during the last fifteen years. The representation and assessment of gait performance is obtained through the measurement and analysis of kinematics, dynamics, and electromyographic (EMG) patterns [2]. Much emphasis is placed on the kinematics and dynamics of gait. A monograph by Sutherland [6] discusses the utility and necessity of these performance measures. More recently, especially for those disabilities having a neuromuscular etiology, the assessment of EMG patterns has become an essential component in order to determine the correct or incorrect phasings of muscles. Recently, Winter [7] has shown the feasibility of using linear envelopes (LE) of normal EMG patterns as a basis for assessing pathologic patterns. Several quantitative methodologies have been developed which perform the analyses of LE. They implement two basic but different procedures. One allows the comparison of populations of LE and determines if any significant differences occur between the populations. The other methodology takes a pooled population of LE and subdivides it into groups of different shapes.

2. EMG PROFILE ANALYSIS

2.1 Ensemble Averaging

The LE enables the creation of the average pattern of activity and a quantitation of the variability patterns generated over several strides, sometimes called the EMG profile [3,8]. The ensemble average is formed after the LE is acquired or calculated. The ensemble average is formed by:

1. scanning the foot contact patterns for successive strides;
2. normalizing the time base of the stride to a specified number of points;
3. ensemble averaging the time-normalized patterns.

Simultaneously the standard deviation of each time point is also calculated [4]. Together these form an EMG profile for a subject; an example is shown in Figure 1 [8].

Similarly, one can form a profile for a population of subjects by performing the same procedure on the average patterns of individuals in a group. One of the applications of this technique has been to study the variability of EMG patterns with walking speed in the normal population. One creates a measure of variability called the variability to signal (V/S) ratio by summing the variance at each time point and dividing by the summation of the squared average values. The V/S for different muscles is shown in Figure 2. Notice that as walking speed increases the variability decreases and that the more proximal muscles are more variable at any speed.

Care must be used when performing population calculations. The amplitude of the LE depends on measurement conditions; skin resistance, amplifier gain, etc. Thus there is a different proportionality between EMG amplitude and muscle force for each subject. One must normalize the amplitude in order to reduce the variability caused by these factors. Studies have compared using the EMG magnitude at maximum voluntary contraction, maximum value, and average value and have shown that the average profile value is the best amplitude-normalizing factor [3].

2.1 Population Comparisons

Comparisons among populations of subjects can be made in order to determine significant differences in EMG patterns. This can be done using either a nonparametric procedure called the Smirnov test or a Gaussian test [5]. Usually the nonparametric tests are necessary because the LE distribution of normalized amplitudes is not Gaussian. The Smirnov test is implemented in the following manner:

1. for each 5% of the stride calculate the average LE value for each profile of each subject;

2. in each population calculate the probability distribution function (PDF) of amplitudes for each 5% period, 20 sets;
3. find the maximum difference between the two PDFs;
4. if they are greater than the significance limit, then the populations are different for that time period.

Figure 3 shows the results comparing RF patterns between population of normal individuals and those who had a surgically repaired anterior cruciate ligament.

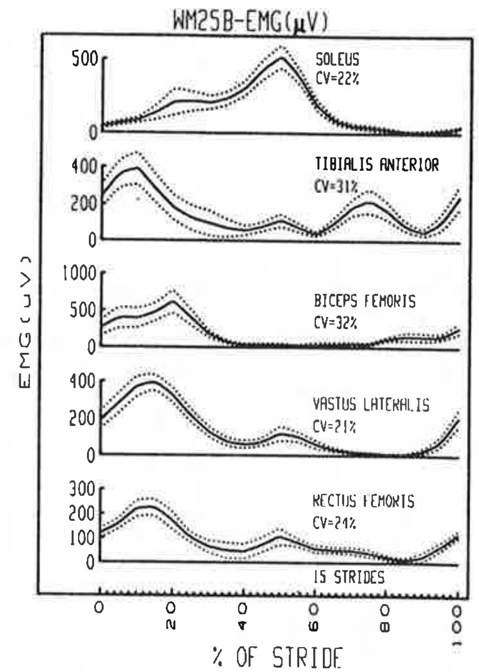

Figure 1. EMG profiles for a normal subject.

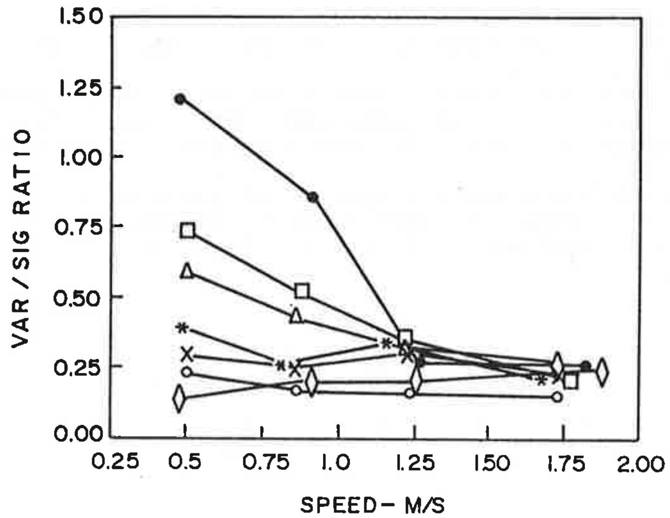

Figure 2. Variability to Signal ratio for a normal adult population.

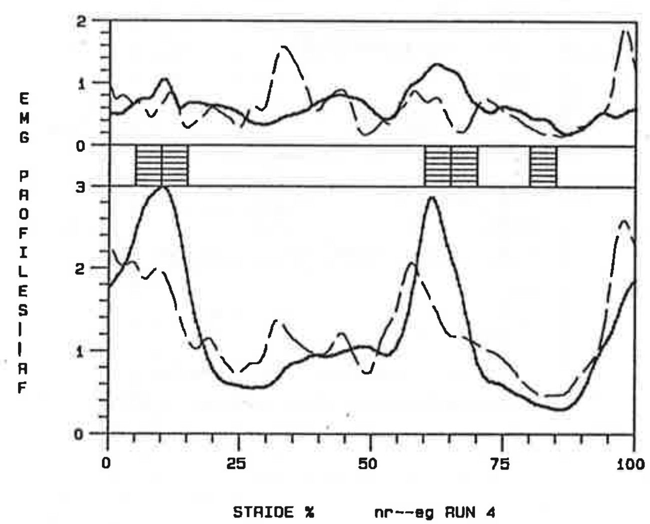

Figure 3. RF EMG profiles for a normal (solid) and surgically repaired ACL population (dash). The hashed areas in the middle window indicate time periods of significant difference.

3. PATTERN ANALYSIS

Pattern analysis differs from profile analysis in that it seeks to determine the pattern types within a population instead of determining the average and variability characteristics. This has just become important recently because it is realized that subjects who have a pathologic gait often have normal patterns in some muscles. Similarly, normal individuals sometimes display "atypical" patterns in some muscles. The goal of this type of analysis is to quantitatively determine the various pattern types that exist in a population. Two of the approaches are presented.

3.1 Fourier Series Approach

Walking is a cyclic task; thus, all of the measurements form periodic time series and it is feasible to calculate the harmonic representation of the EMG profiles. Since the stride duration is normalized to 1, the fundamental frequency, f_o, is also one and

$$y(t) = C_o + \sum_{n=1}^{M} C_n \cos(2\pi nt + 2\pi n\tau_n)$$

where C_n = amplitude, n = harmonic number, τ_n = time shift, and M is the maximum harmonic number. The magnitudes and time shifts of each cosine component are the features used to describe an EMG pattern. The pair for n=1 is called the first component, n=2 the second, etc. Figure 4(a) shows the time varying portion of an EMG LE from a rectus femoris muscle. Figures 4(b), (c), and (d) show the signals reconstructed by using the 1st to 4th, 1st to 7th, 1st to 10th harmonics, respectively (the dashed lines represent the original LE). Ten harmonics represent the LE quite adequately. In fact, subsequent usage of the features in pattern analysis has demonstrated, in general, that eight features are adequate and account for more than 90% of the waveform power [8,9]. Thus, the LE can be represented by a set of 16 features comprised of 8 pairs of harmonic amplitudes and phase angles. As one would expect, LE with different shapes have features with different magnitudes.

The next step is called clustering. This is a procedure which searches through all of the sets of features and groups those with similar magnitudes. The result from applying this approach to a pooled population of normal and ACL injured subjects is shown in Figure 5. Six different pattern types have been produced. Pattern one is predominantly produced by normal individuals whereas patterns 1, 4, and 5 are produce by injured individuals [9].

3.2 Tauberian Series Approach

The Tauberian series was investigated for describing EMG LE because the basic form of the series is a summation of pulses or bursts which look similar to the LE [1]. It is very descriptive because its parameters, features, indicate the time, duration, and amplitude of the phases of activity, thereby mimicking physiologic terminology. A sum of basis functions, $x(t)$, is used to produce an approximation, $y(t)$, of the linear envelope;

$$y(t) = \sum_{i=1}^{M} a_i x(t - \tau_i)$$

where a_i and τ_i are the features which designate the amplitude and time of occurrence of EMG periods and M is the number of terms. The basis function that most resembles EMG phases of activity is the unnormalized Gaussian function

$$x(t) = \exp[-t^2/2\sigma^2]$$

with the spread factor σ, standard deviation parameter. The features are calculated using a scale-space technique. An example of the results of the technique is presented in Figure 6. The smooth curve is the EMG envelope and the pulses representing the feature sets are superimposed with dashed lines. Each phase of EMG activity is represented by a set of time, τ_i, amplitude, a_i, and spread, σ_i, features. The features are valid and very accurate because they represent directly the characteristics of the LE used for evaluation of its state of normality or abnormality. This approach has been applied to normative data bases in children and cerebral palsy data bases [1]. These features have been found to be excellent as variables when applying clustering techniques when each waveform is represented by two phases.

4. SUMMARY

This goal of this brief review has been to give an overview of the utility of some procedures for analyzing EMG in the study and clinical evaluation of pathologic gait. There are many medical centers which use EMG as a diagnostic component for gait disabilities arising from a variety of causes including cerebral palsy and trauma. The patterns of EMG activity are represented in several ways. The measured interference pattern is still used, although the linear envelope is becoming more frequently used. The LE lends itself to quantitation for clinical and research purposes.

Because gait research usually entails using large data bases, several quantitative methodologies have been developed for analysis utilizing the LE as the mode of representing the EMG. These methodologies perform two basic but different procedures. One allows the comparison of populations of LE and determines if any significant differences occur between the populations. The other methodology takes a pooled population of LE and subdivides it into groups of different waveform, LE, shapes.

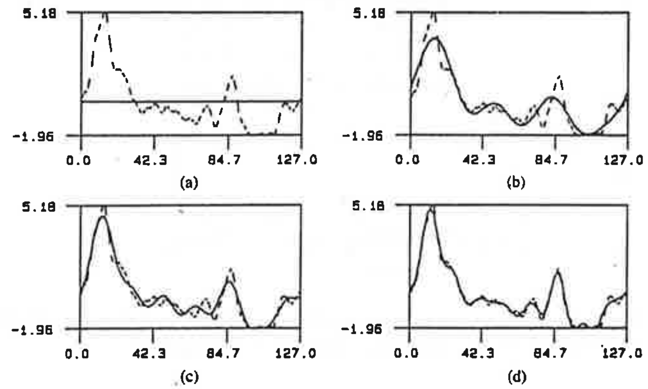

Figure 4. Fourier series representation of an LE from an RF muscle.

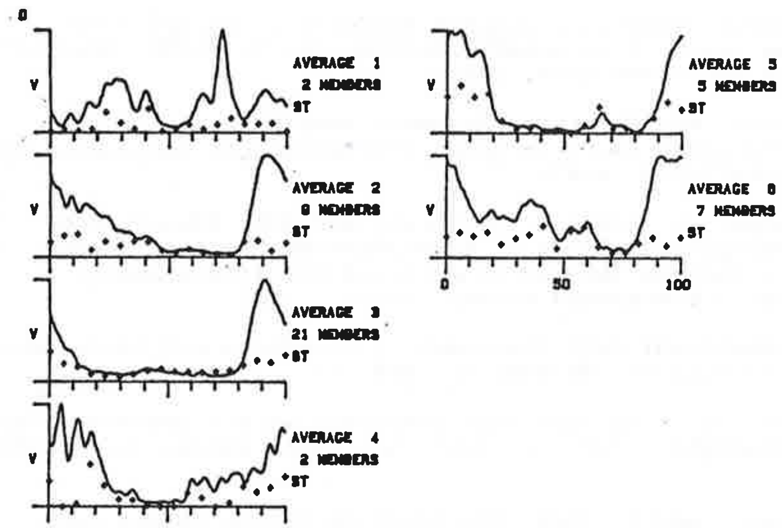

Figure 5. Pattern types in a population of semitendinosus (ST)

Temporal Features				
	1	2	3	4
α	3.471	0.496	0.284	1.808
σ	0.064	0.068	0.044	0.061
τ	0.056	0.298	0.462	0.681

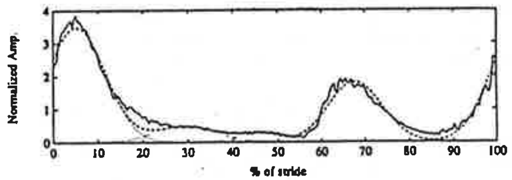

Figure 6. The features and resulting model of an LE from a tibialis anterior muscle pattern.

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FREQUENCEY ANALYSIS OF THE DYNAMIC POSTURAL MOVEMENT IN THE ELDERLY

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INTRODUCTION: This paper discusses with the postural sway characteristics to floor surface perturbations in a upright standing, by using a movable platform. Nashner LM, Diener HC, etal have reported that there are a few typical strategies of postural responses. Here it is studied how the frequency components will change in the angles of the ankle, knee joints and the upper body. Also some of characteristics of the floor reaction are described by the vector representation. It is useful to presume the effects of sensory inputs, and the co-ordination of the lower extremities in the elderly.

MATERIAL AND EXPERIMENTAL METHOD: The perturbation is generated by a new movable platform(1). The signals measured are as follows; (1)the vertical components of reaction forces of one pairs of the platforms(300×600mm), (2)the anglular displacements in the lower extremities and the trunk by the optical measuremnet system with the PSD cameras, etc.
25 male volunteers aged 24 to 72 years who were tested while standing in the indicated position. The stimulations used are follows; (1)ramp inputs and (2)sin-wave inputs in the forward/backward and lateral directions.

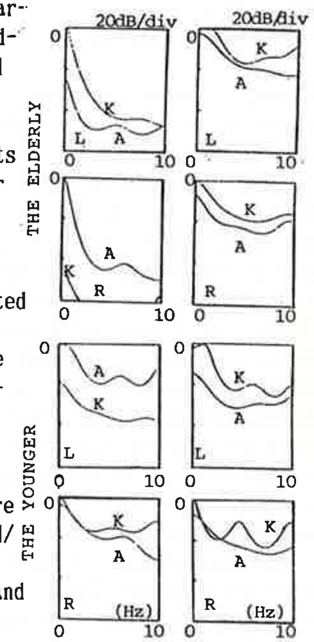

Fig.1 POWER SPECRUM OF THE KNEE(K) AND ANKLE (A) JOINTS

POSTURAL RESPONSE PATTERN: (only the results of the forward and backward perturbations are discussed in the abstracts.)
Concerning the so-called automatic postural response, information of the joint angles in the lower extremities is processed by a techniqe of the frequency analysis. The power spectrum(Fig.1) used here, is calculated by the lattice least square method, based on the autoregressive model. In the younger, though a little imbalance between two legs, they have showed the 'ankle strategy' in the slow perturbation, and the 'hip strategy' in the fast. However, in the elderly, the knee angle is invloved in the slow perturbation. This is often observed in the elderly. It strongly suggests that the distal muscles, can't have a faster reaction of the sensory input, and these lateresponse results in a compensated and complicated postural control. Lastly the angular displacements of the trunk have also confirmed a similar response.

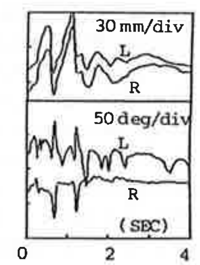

Fig.2 VECTOR PRESENTATION; The upper = amp.,the lower=angle (R=right,L=left foot)

DYNAMIC AND STATIC IMBALANCE: We have tested the relative extent of each leg to the weight-bearing. The vector representation of the center of foot pressureis introduced to describe the loci as shown in Fig.2. It is characterized that near 1 Hz titubation, similar to natural frequency of the human body, is obserbed in the higher perturbation, though in the younger, there is usually a small amplitude. The sine-wave perturbation was induced to test the effect of sensory inputs in this sway. Here wepoint out thatthe aged paerson was relatively late.

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IS THE TRUNK AXIS A REFERENCE FRAME FOR CALCULATING THE LEG POSITION?

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During unilateral leg movements performed while standing, it is necessary to displace the center of gravity towards the other leg in order to maintain equilibrium. In addition, the orientation of particular segments, such as head and trunk, which are used as reference values for organizing the motor act, needs to be preserved. The coordination between equilibrium control and the ability to maintain the orientation of given segments (head, trunk) was previously studied in standing subjects instructed to raise one leg laterally to an angle of 45° in response to a light. Two sources of light placed at eye level indicated the side on which the movement was to be performed. Two control strategies were identified (Mouchnino et al., 1992). An "inclination" strategy was used by the naive subjects (n=5). This consisted of an external rotation of the body around the antero-posterior ankle joint axis; a counter-rotation of the head with respect to the trunk was observed, which ensured the stabilization in the horizontal plane of the interorbital line. A "translation" strategy was used by the dancers (n=5). Here the external rotation of the leg around the ankle joint was associated to a feedforward counter-rotation of the trunk around the coxofemoral joint so that the horizontality of the interorbital line and the verticality of the trunk axis were maintained. This new coordination resulted from long term training and indicates that a new motor program has been elaborated.

The maintenance of the verticality of the trunk axis during leg raising in dancers raises the problem of its functional significance. The analysis of the final position reached by the leg whom being instructed to raise a leg to an angle of 45° and to maintain the final position provides a cue for answering that question.

The kinematic analysis showed that in naive subjects, the new leg position was always higher (56° +/-9) than the instructed position (45°). In dancers however, the instructed angle (45°) was approximately adopted (48° +/-9). The possibility that the trunk axis may have been used as an egocentric reference frame to calculate the final position of the moving leg was explored by measuring the angle formed by the trunk axis relative to the vertical axis. This angle was 13.2° (+/-3.2) and was statistically equivalent (r = 0.84) to the additional angle (12.9° +/- 5.4) reached by the moving leg beyond 45°. These results suggest that in both naive subjects and dancers, the final leg position might be calculated with respect to the trunk axis orientation. A possible interest of keeping the trunk axis vertical is that the trunk axis could serve as egocentric reference for the performance of leg movement. As the trunk axis coincides with verticality, the transformation of the limb position or trajectory from intrinsic coordinates, calculated on the basis of the trunk axis, into extrinsic coordinates, calculated with respect to the vertical or gravity vector, would be simplified.

Mouchnino L. Aurenty R., Massion J., Pedotti A. (1992) Coordination between equilibrium and head-trunk orientation during leg movement: a new strategy built up by training. J. Neurophysiol., 67: 1587-1598.

THE CONTRIBUTION OF RAPIDLY ADAPTING RECEPTORS TO THE POSTURAL CONTROL. STUDY OF THE FORM AND DISTRIBUTION OF PACINIAN PLANTAR CORPUSCLES

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The present study includes the ultrastructural features of fast-adapting Pacinian corpuscles of human planta with the aim of understanding their action in the postural control. These receptors represent the first units of the reflex medullary activity;they,together with the mechanoreceptors of Meissner,Ruffini and Merkel, transmit the description of mechanical events from the body surface to the central nervous system. The Pacinian and Meissner corpuscles are rapidly adapting mechanoreceptors that emit one or two impulses to a constant pressure stimulation,while the corpuscles of Ruffini and Merkel are slowly adapting receptors that emit a discharge of impulses during the whole period of the tactile stimulation.Our study concerns only Pacinian corpuscles. METHODS. The approximate locations of Pacinian corpuscles were determined by dissections of six small areas of a human planta. Tissue removed at surgery was fixed in 10% buffered formalin and embedded in paraffin. Blocks were sectioned at 7 to 9 micron and stained with hematoxylin and eosin. A total of 84 specimens was examined.

RESULTS. In the 84 specimens we found 56 units (33 in the right foot and 23 in the left one).24 of them were in bunches, the other 32 were solitary. Four couples of corpuscles in bunches had a common capsule and were vascularizated by their own artery. 42 corpuscles were in the subcutaneous fat below the transverse metatarsal arch,7 were in the subcutaneous fat of hallux and 7 along the posterior part of the external longitudinal arch. The dimention of the major axis varied from 0,3 to 1,8 mm., with a mean of 0,6 mm.. The profundity of receptors varied from 2,5 to 14 mm., with a mean of 7 mm.. Our analysis points out that the big bunchy units are deeper than the others.

CONCLUSION. From this study come out the following results: a) Pacinian corpuscles are more numerous in the forefoot;b) their trophism depends on the fat tissue;the larger is the quantity of fat tissue,the larger is the number of corpuscles. The metaplasia of adipose tissue in fibrous tissue causes a loss in the number of these receptors;c) from a comparison with the international literature comes out that the number of the plantar receptors we observed is larger than the one found in the hand. This may indicate the presence of smaller receptive fields and, consequently,a more precise perception. In conclusion, this peculiar plantar perception could be properly utilized by the mechanisms of reflex postural control.

POSTURAL ADJUSTMENTS ASSOCIATED WITH FAST ELBOW FLEXION MOVEMENTS IN THE ELDERLY

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Upright posture is maintained by specific activations of postural muscles which are aimed to counteract joint reaction moments arising from voluntary (focal) movement and to reach static balance at the end of the motor act. The aim of the study was to investigate differences due to age in the recovery phase of equilibrium after performing fast symmetric elbow flexions. We studied 8 aged (66-79 years)and 9 young (21-32 years) healthy subjects. Young subjects performed elbow flexions with angular accelerations in the same range of the elderly. Subjects were requested to stand upright holding a horizontal bar with their forearms supinated and their eyes open (EO) or eyes closed (EC). After 10-15s of quiet stance they were requested to perform fast elbow flexions and, successively, to maintain the new position for 13s more. Anterior-posterior displacements of center-of-pressure (COP) were sampled (100 Hz) using a force platform (Kistler). Elbow, trunk, hip, knee, and ankle angle movements were sampled (50 Hz) with a 3-D video based motion analysis system (ELITE). Off-line, from the computer recordings, periods of 8 s were selected for each trial (3 s before and 5 s after the onset of elbow flexion movements). These periods were further divided in 500 ms intervals in which the total lengths of COP (mm) and joint angular (degrees) movements were calculated. With EO, before the onset of the motor act, the length of COP, trunk, hip and knee movements was greater in the elderly while no differences were observed for elbow, and ankle. In the time interval in which the focal movement occurred, the length of move ments of COP and joint angles showed a sharp augmentations in both groups. However, the length of movements of trunk, and hip and knee was greater in the elderly. After the end of elbow flexion, lengths of movements rapidly decreased. The time required to achieve values similar to those before the motor act was longer in the elderly for COP (figure 1) and all postural joints. With EC, the time required for recovering stability was prolonged in both groups. Results indicate that elderly subjects adopt a different strategy to maintain standing posture while executing fast arm movements and require longer time to recover from instability induced by voluntary movements.

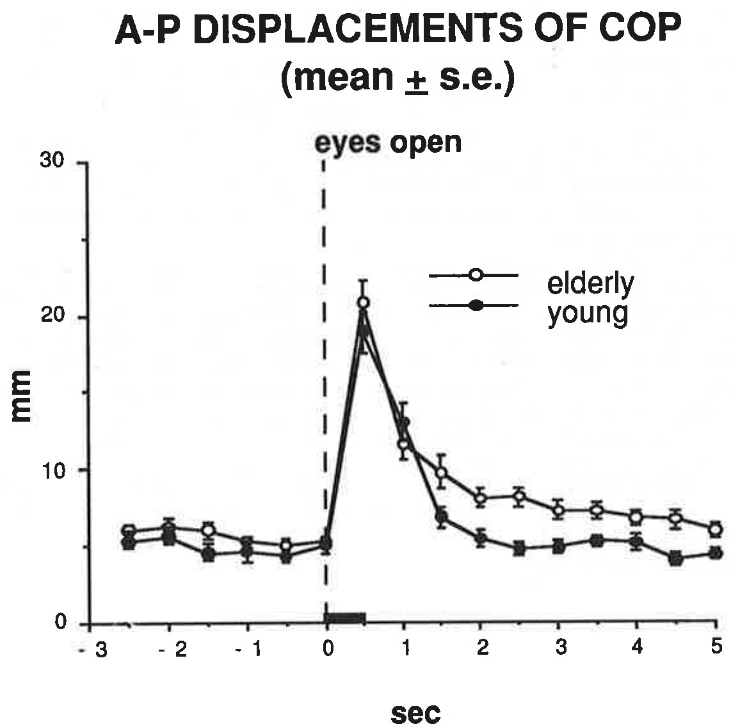

Figure 1. Length of the Anterior-posterior sway path of the center of pressure computed for each of the 500 ms interval. The broken line indicates the onset of elbow flexion movement. The broad segment indicates the time interval in which the focal movement occurred. The time required to achieve values similar to those before the motor act was longer in the elderly.

A COMPREHENSIVE SYSTEM FOR CLINICAL GAIT EVALUATION: METHODS, PROTOCOLS AND ELABORATION PROCEDURES

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Techniques for gait analysis have achieved a high degree of reliability and accuracy, so that application to clinical investigation is now conceivable.
At the Bioengineering Center of Milan, an initiative has been undertaken in order to transfer to the clinical side most of the expertize acquired in several years of activity in the field. In this process several problems arise that are not merely technical in nature, but also are problems of communication with the clinicians, organization of the routinary activity, definition of protocols for every phase of the elaboration procedure.
Two different levels of data analysis have been identified: the first is a pure description in biomechanical terms of the main determinants of the pathological gait; the second consists in the definition of a physiopathologic profile, based on different correlations between the variables analysed. Methods, protocols and elaboration procedures have been designed to match different requirements: first of all to furnish good measurements of the variables of interest; secondly to present the results in a form which can be easily interpreted in clinical terms; thirdly to provide correlation between different variables according to the definition of physiopathological factors.
Basically, the method consists in the analysis of kinematics, ground reaction forces, EMG signals. A special set-up of TV-cameras and a particular marker arrangement allow us to detect the movement of the head, upper limbs, column, pelvis and lower limbs in three dimensions. A special purpose software package has been implemented to make elaborations such as: reconstruction of points internal to the body, identification of the segment axes, computation of space angles, joint angles, joint torques, joint powers and global time/space parameters. Comparison between different variables and between variables and EMGs is always possible in a friendly interaction. Further elaboration are possible such as muscle length and muscle velocity computation, correlation between muscle kinematics and EMG output, correlation between joint moments and joint angles. The whole package of programs is controlled by a manager system which updates a data base and allows us to compare different subjects, different populations, single subjects with a reference population. A preliminary activity on patients has already been started and the approach seems to be adequate for answering several clinical problems.

ELECTROMYOGRAPHIC AND DYNAMIC MOVEMENT CHANGES FOLLOWING SURGICAL TRANSFER OR RELEASE OF RECTUS FEMORIS MUSCLE IN CHILDREN WITH CEREBRAL PALSY

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Children with cerebral palsy frequently suffer from restricted range of motion at the ankle, knee and hip due to muscle contractures around those joints. Spastic overactivity of the muscles is the primary problem causing abnormal gait patterns. Surgical intervention to release, lengthen, or reattach the muscles is indicated in many cases to allow more efficient gait. It is not well understood what the effect of surgery has on the muscle activity during gait. The purpose of this study was to examine electromyographic (EMG) and dynamic changes of the lower extremities following surgical transfer or release of the rectus femoris muscle in children with cerebral palsy. In a gait laboratory telemetric EMG and motion analysis were recorded pre- and post-operatively on 66 spastic hemiplegic, diplegic or quadriplegic patients (range = 4.1 to 18.1 years of age) allowing observation of 119 lower limbs. Rectus femoris EMG was recorded and quantified as the percent of the gait cycle during which it was active. The percent of gait cycle for time to peak knee flexion was chosen as a measure of dynamic motion, because this commonly occurs very late in the cerebral palsy gait cycle. Surgical procedures on the distal tendon of the rectus femoris muscle were classified into four types: 1) release; 2) medial transfer to the sartorius muscle; 3) lateral transfer to the iliotibial (IT) band; or 4) other medial transfer (to the gracilis or semitendinosis). A 4 x 2 (type of surgical procedure by pre- and post-surgical gait assessment) analysis of variance was performed for percent of rectus femoris EMG and time (as percent of gait cycle) to peak knee flexion. A .05 significance level was used for all comparisons.

Significant main effects for pre- and post-operative rectus femoris EMG, $F(1,115) = 10.3$, $p = .002$, and pre- and post-operative time to peak flexion, $F(1,122) = 8.1$, $p = .005$ were observed. No other main effects or interactions were statistically significant. Following surgery the average rectus femoris EMG decreased from 86.1% to 77.7%. Post-surgically, the time to peak flexion also decreased significantly from 85.0% to 80.7%. Although the post-operative measures were not reduced to normal levels (normal rectus femoris EMG is approximately 35% and time to peak flexion is 70%), the differences generally resulted in greatly improved gait.

Although not statistically significant the transfer to the iliotibial band resulted in the least change in EMG (IT transfer average = 1.3% decrease, all others average = 10.7% decrease) and time to peak flexion (IT transfer average = 1.6% decrease, all others average = 5.2% decrease). Surgical procedures on the rectus femoris resulted in an overall improvement for reduction of overactivity of the muscle and improved gait function during dynamic knee flexion. Perhaps of clinical significance the release and medial transfer procedures of the rectus femoris resulted in a more favorable outcome than the lateral transfer procedure for both variables, although these differences were not statistically significant.

THE EFFECT OF TENDON VISCOELASTIC PROPERTIES ON THE DYNAMIC PERFORMANCE OF THREE LOAD MOVING MUSCLES

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Many reports have discussed the energy storage properties of tendon during activities such as running and jumping. The viscoelastic nature of tendon has the potential of modifying the dynamic response of the musculotendon complex.

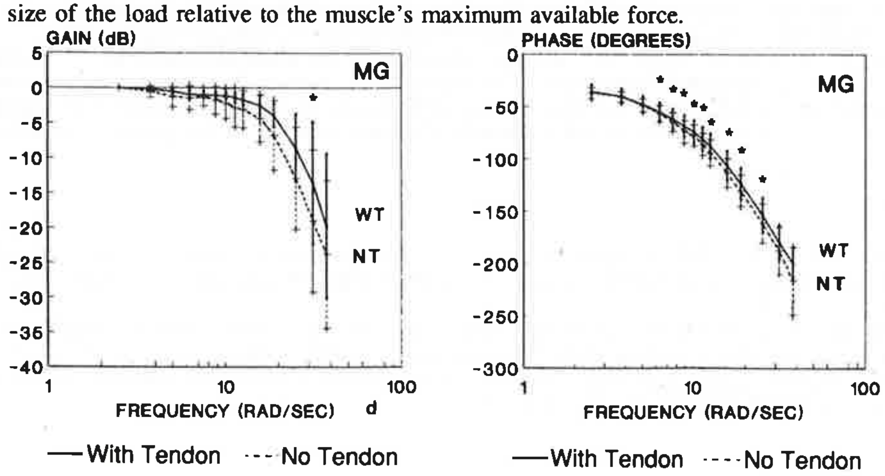
Previous reports examining the effect of tendon properties on the dynamic response of isometric muscles showed that its effect was insignificant. It was indicated, however, that the possibility of the tendon's properties having a significant effect during load moving contractions was not ruled out. The aim of this study was, therefore, to examine the effect of the tendon's visco-elastic characteristics on the dynamic behavior of load moving muscles.

Using sinusoidal variations in activation of the muscle over the frequency range of .4 to 6 Hz, frequency response models of the loaded Medial Gastrocnemus, (MG), Tibialis Anterior (TA) and Extensor Digitorum Longus (EDL) were performed. The MG and TA were loaded with a 1 Kg suspended weight, and the EDL with a .5 Kg load. After the frequency response test of the muscle with tendon was taken, the tendon of each muscle was removed, and the frequency response test repeated. The data were presented in frequency vs gain and frequency vs phase Bode plots (**Figure 1**).

Analysis of variance with repeated measurements comparing the gain and phase data of each muscle before and after tendon shortening showed that in the MG and EDL, the gain is significantly (asterisks in **Figure 1**) lower and the phase lag is greater without the tendon at some of the test frequencies. Conversely, the TA shows a significant increase in the gain at medium frequencies as well as decreased phase lag without tendon.

Harmonic distortion calculations show no significant change due to the tendon removal indicating that the tendon was in its linear viscoelastic region.

The fact that in every muscle there was a small defined area of statistically significant difference in the dynamic response of load moving muscle with and without tendon suggests that in this frequency range there is some interaction between the load mass and the tendon which was not observed in isometric conditions. By the nature of the results, it is apparent that these interactions vary from muscle to muscle, possibly depending upon the muscle architecture such as pennation, muscle/tendon length ratio, tendon thickness as well as on the size of the load relative to the muscle's maximum available force.



POSTURAL DISORDERS IN MULTIPLE SCLEROSIS WITH RESPECT TO OTHER KINDS OF CNS DISEASES: AN IDENTIFICATION APPROACH

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A black box modelling approach and subsequent identification were found useful in recognizing Multiple Sclerosis (MS) patients at a preclinical stage of the disease with respect to a normal population.[Corradini et al. 1990a, and 1990b]. In this study the comparison has been extended to other kinds of Central Nervous System pathologies.

In particular a population of 11 MS patients was analysed and compared with respect to:
- 5 subjects affected by hemiparesis resulting from Cerebro-Vascular- Diseases (CVD),
- 1 patient suffering for a cerebellar disease,
- 1 patient affected by cranial cerebral injury.

The population of the control group consisted in 26 normal volunteers, matched for sex and age with the pathological population.

The standing upright subject has been modelled as a single inverted pendulum in his sagittal plane. The assumption has been made that equilibrium is maintained by the subject applying a torque at the ankle joint. A measure of the efficacy of the posture control system is the subject's body sway. This latter quantity has been considered as the controlled variable in the chosen model of the posture control model. A black-box (ARMAX) model of the posture "controller" has been assumed and its parameters have been identified.

Every subject has been analysed in open eyes and in occluded vision conditions by means of an optoelectronic kinematic measurement system (CoSTEL) and of a force platform.

Every MS patient was at the lowest degree of neurological impairment: EDSS ranging from 0.0 and 1.5 according to Kurtzke's EDSS score [Kurtzke 1983].

As far as MS patients are concerned, results confirm those already reported in literature (with a lower number of MS subjects): while normal subjects are characterised by a specific order of the controller ARMAX model, 7 MS subjects out of 11 showed a different model order with respect to the control group. Two of the subjects characterised by the same controller order as the healthy population, had had in their clinical history only optic neuritis.

All the 5 CVD patients were characterised by the same order of the "controller" as the healthy subjects, in both the experimental conditions. As well as for the healthy subjects the order remained unchanged both interindividually for subjects performing the same task, both between the 2 tasks performed by the same subject.

The same results hold for the cerebellar patient while the subject suffering from traumatic cerebral injuries behaved like MS patients as far as the controller model order is concerned.

This may be due to the diffused signal disorders in the neural nets of balance regulation.

Results put into evidence the high level of selectivity of the approach based on the system identification of the ARMAX "controller" block and seems to suggest that its power of detection could depend on a suitable threshold related to the EDSS score. Moreover the model order change of the controller observed may reveal a possible different strategy of the posture stabilizing behaviour.

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Kurtzke J.F.(1983): Rating neurologic impairment in multiple sclerosis: an extended disability status scale (EDSS). Neurology 33, 1444-1452.

A SIMPLE VIBROMETER FOR STUDYING BIOMECHANICS DURING HUMAN STANDING

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Summary

The objective of this study was to develop and test a new method to estimate some biomechanical factors such as the joint stiffness and muscle activities during human standing. Although the device used in this study has the similar feature in some aspects to a kind of vibrometer, the usage of the device in this study is quite different from that of any kind of vibrometer.

The device consisting of a plate supported by four columns resembles a type of force plate, however, the natural frequency of the device is very low. The dynamic characteristics of the device itself are basically same as those of a column with a lumped mass at the top end. The complex system with the device and the man on the plate, however, cannot be represented by a simple model. The human body in this situation can be modeled as a spring-mass-damper system with multiple degrees of freedom. The total system can be modeled by an AR(autoregressive) process with unknown input, and the output measured as the motion of the plate. The model parameters were calculated by using Burg's algorithm, then power spectra were estimated. This method is sometimes called MEM (Maximum Entropy Method).

A series of a pseudo impulse response can be obtained by giving some impacts in sequence on the leg during standing. Considering these results, a simple model with one degree of freedom was used to estimate the joint stiffness. The estimated values, although negative, showed a consistency for different settings. Some examples of the spectra for different conditions of standing were presented.The peak in the high frequency region (5-10Hz) of the spectrum showed a shift to the lower or higher side according to the condition.

Several indices to estimate the activity level of the human body during standing were derived from the results of the model analysis. Although the activity level seems not to be directly related to the stability of the posture, those indices will be useful for evaluating the human performance during standing. Improving the frequency resolution will increase the usefulness of this method.

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RELATIONSHIP BETWEEN MORPHOFUNCTIONAL POSTURAL ATTITUDES OF THE SHOULDER (MUSCLE TONE AND MORPHOLOGY) AND INTERPERSONAL CONTACT STYLES

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The aim of the present research was to examine the relationship between muscle tone at rest (frequency and amplitude) and the styles of interpersonal contact. Following the literature on nonverbal communication, we consider the problem of contact as complex, with different levels of articulation: putting one subject at the centre of an ideal microcosmos, we consider around him there are different concentric areas of contact. We can represent this phenomena by different concentric circles: the first one coincides with the surface of the body of the subject and represent the concrete physical barrier that separates him from the environement. The second hypothetical circle defines the area of phisical erotic intimacy. The third one indicates the area of spatial distance, also called "personal space" which we have interpreted as an ethological space. The fourt level corresponds to social area of formal interpersonal relationship. On each indicated level subjects can assume different styles of contact. For example, in relation to the area of physical erotic intimacy, subjects can also refuse the interpersonal contact in this area. To describe these different conditions we have suggested the concept of "conflict of contact" and of "refusal of contact". For the last concept, in relation to the idea of concentric functional circles, we have emplyed the term "Barrier" to contact. So we have for each level of interpersonal contact two possibibilities, one of conflict and one of a barrier to contact.

SUBJECTS

The sample of research was composed by 54 women incluted 24-32 years. All subjects declared themselves right handed.

METHOD

It has been used an electromyographic record with surface electrodes applied on particular points of the neck and of the shoulder. The fequency changes of EMG potential, amplified by an amplifier with band passing between 60 and 1200 Hz, were transferred into a digitalized signal by means of zero-crossing apparatus. In turn the digitalized signal was converted into a tension signal by an integrator, expressing a difference in potential as proportional to the EMG potential frequency. The tension-signal was sent to a tension-frequency conveter and then to a decoding counter. The decoding counter showed the mean integrated frequencies appearing in 1 sec.

We can study the styles of interpersonal contact using a questionnaire about some usual interpersonal behavior of the subiect.

RESULTS

The increase of myographic frequency was positively and significantly correlated , using Sparman Rho, with the scores on barrier in interpersonal contact, while the increase of myographic amplitude shows a trend to facilitate the interpersonal contact (positively and significantly correlation).

MOVEMENT BALANCE SYSTEM THEORY: THE ROLE OF APPLIED BIOMECHANICS

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The aim of this study is to examine the implications on clinical practice after a description of the general theory. The Movement System Balance Theory proposes that the path of the external and internal forces is an index of the quality of the system's functional status. The interactive nature of human biomechanical system requires a complete examination with quantitative kinematic and kinetic analysis to determine the balance of internal and external forces. This permits an understanding of the structural alterations through the knowledge of kinematic parameters and ground reaction data during the action. To determine the balance of the locomotor system it's also useful to assess the properties of muscle work by surface EMG. The imbalance of locomotor system is a widespread pathology. If acute trauma has never occurred, it is caused by cumulative microtraúma that lead to the disfunction of locomotor system. Therefore, microtrauma is a fundamental aspect of the pathology related to movement imbalance.By microtrauma ,we mean a traumatic phenomenon each of whose occurrences is of insignificant intensity. By numerous repetitionsof this event it is able to cause changes in the bone-muscle-tendon apparatus , without pointing out any kind of the subjective syntomatology. The microtrauma often associated with all forms of stéreotyped repetitive activities.
According to the Movement System Balance theory , symptomatic treatment can not correct faults in movement. For this reason it's important to define the dynamic posture and its forces during the gait.
Our clinical éxperience from 1986 on permitt to state that the gait analysis data give some information which cannot be obtained by clinical observation and allows quantitative evaluation of the therapies and functional aids.

A NEW STABILOMETRIC METHOD TO EVALUATE ANKLE FUNCTIONAL INSTABILITY IN A CLINICAL SETTING: TEST/RETEST RELIABILITY

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1 - Introduction

Reliability of force platform measurements has been proved in laboratory settings [1,2], but not in clinical ones. The validation of a new stabilometric method we developed in a clinical setting to evaluate ankle functional instability was the aim of our study.

2 - Methods

2.1 - Subjects

12 subjects, males (24 ankles) were examined standing on one foot (arm around chest, eyes open) for 60 seconds; 7 of them (14 ankles) were retested immediately after repositioning descending platform (Group A); 11 (only right ankle) were reevaluated after about 10 minutes, having meanwhile performed other tests with braces and taping (Group B). Linear regression was computed on these two sets of data and Pearson's Correlation Coefficients (r), R^2Test and ANOVA on theoretical model developed were used to investigate test/retest reliability.

2.2 - "Balance Platform" system

We used the "Balance Platform" system made by Cosmogamma. 3 mechanical-electrical force transducers under the platform calculate via computer the movements of Center of Pressure (ground projection of Gravity Center of the body) using ground force reactions on feet. Software offers data base management and video and printed output too. Examinations were carried out always by the same physician, minimizing visual and earing disturbances. There was one fixation point, a vertical line, at two meter distance.

3 - Results and Discussion

In Group A correlation coefficients were high for Space (r= 0,929; R^2 =86,3%), Speed (r= 0,909; R^2 = 82,7%), Sagittal Speed (r= 0,923; R^2 = 85,2%) and Frontal Speed (r= 0,854; R^2 =73,0%). These variables presented r< 0,750 (and P< 0,01 in ANOVA on the model) in Group B too, where other disturbing elements (learning, fatigue, etc.) were presumably present. In Group A also maximal deviations on both axes demonstrated good correlations (Sagittal: r= 0,729 - Frontal: r= 0,596), but not in Group B. These data offered a valid representation of the entire phenomenon, and demonstrated a good reliability, unlike Sagittal and Frontal Oscillations whose variability was too high.

4 - Conclusion

It was possible to conclude that some variables are reliable at intra- examiner test/retest in a clinical setting too, offering data valuable in time.

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DEVELOPMENT OF PROTOCOLS FOR CLINICAL USE OF OBJECTIVE AND QUANTITATIVE MOVEMENT ANALYSIS TECHNIQUES

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INTRODUCTION

One of the objectives of the Clinical Evaluation Centres Network active in SubGoal A.2 of the CAMARC II project is the Definition of Concerted Functional Evaluation Protocols. This paper describes some views towards this problem from the rehabilitation clinician's perspective. It can help establish a conceptual framework which may serve as a guide in the development of clinically relevant applications of Movement Analysis under CAMARC II.

PROBLEMS IN CLINICAL APPLICATION OF MOVEMENT ANALYSIS (M.A.)

Although there are a considerable number of laboratories with M.A. facilities it appears that there are only very few clinically accepted applications of movement analysis techniques at this moment. For instance, a study by Messenger (1987) surveyed the clinical impact of gait laboratories in the United Kingdom; they found that only 6 out of 35 laboratories were more or less routinely involved in clinical work.

Leo (1991) described four reasons for this limited clinical acceptance. Independently Kleissen (1990) analysed the same problem.

A common conclusion in these analyses is that the problems involved are not only of technical nature: development of clinical applications also requires the answer to methodological, organisational and conceptual questions. The following will discuss some important concepts which have to be shared by the groups involved in the development of clinical protocols. Otherwise it is not possible to define clinically relevant, concerted functional evaluation protocols.

OPERATIONAL DEFINITION OF "CLINICAL APPLICATION OF M.A."

First of all, there has to be agreement with respect to the notion of "clinical application". As long as "clinical use" is loosely understood as "application in a hospital setting" this notion is too ambiguous to be of practical value in directing the development of clinical protocols for M.A..

For the clinician it is vital that the use of M.A. techniques in one way or another materially affects the treatment of his patients. He sees two ways in which this can be achieved:

1. M.A. supports clinical decision making
2. M.A. provides objective and quantitative data in research

It is obvious how clinical decision making applications of M.A. affect the treatment of patients.

Research applications of M.A. are only relevant to the clinician if they answer well-defined research questions which eventually lead to changes in the patients' treatment procedures by changing the empirical or theoretical basis for the treatment. In the development of clinically relevant concerted functional evaluation protocols only those protocols will be considered which can be convincingly shown to affect treatment of patients in this manner.

CLINICAL APPLICATIONS IN REHABILITATION: THE ICIDH

For successful use of M.A. in rehabilitation it is important to be aware of the questions which are relevant to specifically the rehabilitation clinician. It was already observed a decade ago that quantification of observed movement is not necessarily useful per se (Brand, 1981).

The International Classification of Impairments, Disabilities and Handicaps (ICIDH) published for discussion by the World Health Organisation in 1980 is a useful tool in creating this awareness (Rozendal, 1989). In order to be able to specify the domain in rehabilitation where M.A. can successfully applied it is, in our opinion, necessary to distinghuish Impairment and Disability as defined in the ICIDH.

The definition of Impairment is (WHO, 1980, quoted in Rozendal, 1991): "In the context of health experience, an impairment is any loss or abnormality of psychological, physiological or anatomical structure or function".

The definition of Disability is: "In the context of health experience, a disability is any restriction or lack (resulting from an impairment) of ability to perform an activity in the manner or within a range considered normal for a human being".

Several authors (Rozendal, 1991; Zilvold, 1990) indicated that current tools for M.A. only allow measurement of variables describing the impairment. We will refer to this type of variables as I-level variables.

For applications of M.A. in rehabilitation it is essential to be aware of the fact that the rehabilitation clinician primarily focusses on restoring a patients functional abilities, particularly those important for activities of daily living.
Rozendal (1989) pointed out that M.A. techniques are not suited for describing patients' disabilities. He speculated that disabilities can only be measured with a new class of instruments, ".. developed from social sciences and not from biomechanics." We shall refer to variables describing the functional abilities as D-level variables.

The important question now is: how can I-level measurement using M.A. be useful for the rehabilitation clinician who is primarily concerned with a patient's disability?

The answer can be stated as follows. Rehabilitation treatment encompasses interventions in the psychological, physiological or anatomical qualities, described by I-level variables, of a patient to reduce his or her disabilities. Conceptually, a treatment should be effective on the I-level before something can change on the D-level. Success of a treatment on the level of the impairment is conditional to successes on the level of disability. Thus the relevance of measurement of I-level variables in this concept of rehabilitation practise becomes clear.

It is worth emphasising here that besides I-level variables also D-level variables will be relevant to the rehabilitation clinician.

MEDICAL DIAGNOSIS AND REHABILIATION DIAGNOSIS

Rozendal (1991) wrote on the issue of the diagnostic potential of M.A. techniques that "The physiatrist or rehabilitation specialist does not ask for a medical diagnosis, because he will be perfectly aware of it already". It is true that a stroke patient with for instance a spastic hemiparesis appears to have a perfect neurological diagnosis. However, this medical diagnosis is not sufficient for the rehabilitation clinician, because the functional abilities of patients with the same medical diagnosis can be extremely different.

Therefore the rehabilitation diagnosis can be improved by measurement of those I-level variables which provide a more complete and accurate picture of the impairment of the individual patient. This means that M.A. techniques for purposes of rehabilitation diagnosis and decision making should provide information on the level of impairment which is complementary to the available clinical information. In other words: the information obtained by M.A. is only useful if it makes information available which cannot be obtained in another way.

Another important consideration in this respect is that the rehabilitation diagnosis is not neccesarily improved by the quantitative format of the information. The emphasis should be on the complementary rather than the quantitive nature of the M.A. data.

As an example of this approach we would like to present to you this video of our MYOVOSION system which visualises the surface EMG muscle action during movement.

CLINICAL DECISION MAKING IN REHABILITATION

In clinical decision making in rehabilitation it is vital that the clinician is certain that an intervention at the psychological, physiological, or anatomical level of a patient has been effective i.e. the intervention resulted in the desired alteration of I-level variables. If the intervention did not have the foreseen I-level effect, it is unlikely that D-level improvement can be expected. Thus, an ineffective intervention can be unmasked and an alternative chosen.

RESEARCH USING M.A. IN REHABILITATION

Success of an intervention in rehabilitation is eventually measured by the improvement in the functional abilities of a patient. Because the relation between the impairments of a patient and his functional abilities is ambiguous it is difficult to predict how a treatment of certain physiological characteristics of a patient will affect the functional abilities. The clinician chooses a treatment strategies for improvement of the level of ability depending on his ideas concerning the relation between impairment and disability. This ideas can be based on experience or embedded in theories, or can be plain guesswork.

An important task for M.A. in rehabilitation can be the contribution to the scientific basis for choice of the treatment strategies.

INTEGRATION OF M.A. IN DAYLY ROUTINE OF REHABILITATION PRACTISE

The previous thoughts lead to the following conclusion. If applications of M.A. are to be developed for use in the daily routine of rehabilitation practise, this development should focus on the definition of protocols for improvement of the rehabilitation diagnosis and for rehabilitation decision making.

It is important that, in the development of these protocols, we bear in mind the usual clinical approach towards diagnosis and decision making. Whittle (1991) observes that clinical assessment is based on three things: history, clinical examination and special investigations. In this clinical context M.A. is no other special investigation than an X-ray investigation, also in the sense that an X-ray picture in itself is not sufficient for making a clinical decision.

The M.A. investigation should be conducted in a cycle of hypothesis formation and hypothesis testing (Rose, 1983; Gage, 1983). This implies that before a any M.A. investigation is performed a testable hypothesis should be available. Then M.A. should be used to confirm or reject the hypothesis. Sometimes one investigation may be enough to give clinically satisfactory information, sometimes an alternative hypothesis may have to be tested.

This specific clinical context is an important constaint in the development of concerted functional evaluation protocols for clinical use.

CONCERTED CLINICAL PROTOCOLS FOR M.A. FOR CLINICAL DECISION MAKING

The considerations above lead to our opinion that the priority in the development of concerted clinical functional evaluation protocols should be in the decision making protocols.

A clinical decision making protocol can be defined as a set of rules guiding the clinician in using M.A. techniques to support his decision making.
The ideal decision making protocol would consist of a set of unambiguous rules which (for a given decision making problem):

- instruct the clinician in choosing the data-gathering procedure
- describe the interpretation of the obtained data
- transform the interpretation into the solution of the decision making problem.

The definition of concerted functional evaluation protocols for decision making within the framework of CAMARC II therefore requires for a given decision making problem:

- Harmonisation of Data Gathering Procedures
- Harmonisation of Interpretation Procedures

PRACTICAL CONSIDERATIONS

Given the limited time available in CAMARC II, it is not possible to develop protocols for all patient categories relevant to rehabilitation. We propose to confine the work to the categories Cerebral Palsy, Hemiplegia, Amputees, Multiple Sclerosis, and patients with joint disease.

Also, the three year time span for CAMARC II is likely to be too short for the development of new, complete decision making clinical protocols. In the given time, we propose to focus the development of clinical functional evaluation protocols on expanding the pioneer work by e.g. Gage, Perry, and Sutherland. With this work as a starting point, we shall emphasise harmonisation of data-gathering procedures. Once this is achieved exchange of measured data and experience can begin, so that there will be a basis for further development of ideal clinical decision making protocols.

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THE APPLICATION OF BIO-MECHANICAL COMPUTER AIDED ANALYSIS IN REHABILITATION ENGINEERING

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We report on progress in the application of a computer aided mechanical modelling technique for the solution of problems in design for rehabilitation.

At Brunel Institute for Bioengineering Rehabilitation Engineering Unit we are concerned with the development of optimised designs of products for elderly and disabled people. In order to provide these solutions and to reduce the amount of physical modelling ,for testing initial concepts and at later stages of the design process, we have developed a computer aided analytical procedure based on Mechanical Computer Aided Engineering .

Our approach to the analysis of problems associated with design for rehabilitation is based on an analysis of the complete system rather than the provision of a simple aid, since it has been shown that narrowly conceived devices may induce additional problems or may be rejected by the client.

The advent of computer aided analytical tools has in recent years been accompanied by its application to biomechanical systems as structural entities. Finite element analysis has been particularly useful in the stress analysis of bones, and such techniques have been used to solve problems connected with orthopaedic biomechanics. The analysis of the kinematics and dynamics of the internal forces due to motion and externally applied forces, using the techniques of mechanism analysis, has been valuable, in for example vehicle impact and gait analysis. We review those aspects of this work which relate to our analysis for the design of assistive devices.

At the Brunel Institute for Bioengineering we are undertaking research and development funded by research charities such as the Motor Neurone Disease Association, Action Research, and The Spastics Society. This work has included the design and development of an arm support system which provides the user with enhanced movement in both horizontal and vertical planes .

In parallel with this work has been an analysis and development of hand grips undertaken in collaboration with arthritis care groups. Demographic data indicates that individuals with an arthritic condition form the largest group within the disabled population with a hand dysfunction. Our attention has focussed generally on pans and cookware and specifically on pan handles, we discovered that very little research in relation to biomechanical , anthropometric and ergonomic data existed. A simple model of the hand is being developed through MCAE and will be compared with the results of user and clinical trials on hand grips. The techniques of MCAE available through the ADAMS software are applied to these systems taking into account not only the assistive device, but the biomechanical system Appropriate constraints are applied to the system and the results of user trials analysed to provide new anthropometric data. We consider the implications of this work and discuss likely future developments.

BIOMECHANICAL ANALYSIS OF PARPLEGIC WALKING WITH HIP GUIDANCE ORTHOSIS (HGO)

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An evaluation protocol to analyse the gait pattern of paraplegic walking with mechanical orthosis has been developed in the frame of a pilot project consisting on selecting training and supplying with HGO-ParaWalker five Spinal Cord Injured patients.

The study includes Clinical assessment, Ergonomic assessment, Biomechanical evaluation, Energy Consumption measurement and Bone Mineral Content assessment, to be performed both during training period and after six months of functional use of the orthosis.

Concerning Biomechanical evaluation an experimental set-up has been adopted in which Kinematic data (ELITE system), Ground Reaction Forces (KISTLER platform) and myoelectric signals from eight supralesional muscles were analysed.

The results show the strategy adopted for such an assisted gait allowing a quantitative description of the locomotor pattern. This could be useful to compare different walking restoration techniques as RGO, Hybrid System, FES. Moreover the differences between experienced and inexperienced patients in the use of the orthosis are pointed out, permitting an evaluation of training procedure and its optimization.

Another important application of results obtained by this method is the study of the optimal adaptation of the orthosis according to single subject characteristics.

In these figures examples of Kinematic, Kinetic and EMG data acquired on a T1 complete paraplegic patient are shown.

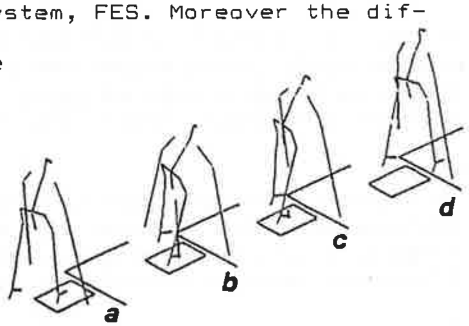

Fig.1 - Total body stick diagram

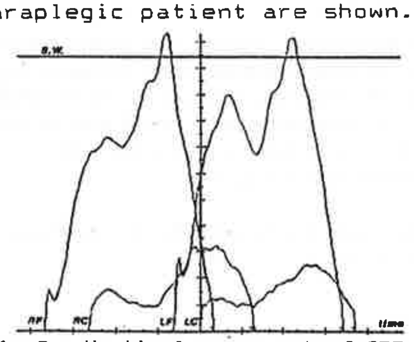

Fig.2 - Vertical component of GRF of: Right Foot, Right Crutch, Left Foot, Left Crutch. B.W. = Body Weight

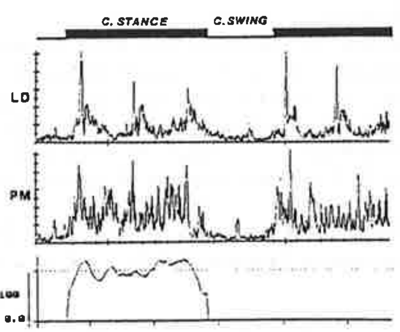

Fig.3 - EMG of Latissimus Dorsi (a) and Pectoralis Major (b). c) Vertical component of ipsilateral crutch GRF

EFFECT OF ANKLE FOOT ORTHESES AND FUNCTIONAL ELECTRICAL STIMULATION (FES) ON STANCE SYMMETRY AND WEIGTH-SHIFTING IN HEMIPARETIC PATIENTS

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Introduction:
Because of their reduced ability to control one part of their body hemiparetic patients prefer not to use the involved half of their bodies. This results in asymmetric postures and movements. This can be seen in their stance when only the noninvolved leg is loaded while at the involved side the hip is retracted and the knee bent.

This posture does not only cause additional stress on the noninvolved limb and gives a bad starting position for walking but causes contractures of the muscles of the involved part as well, and in the long run therefore reduces the mobility of the patient.

To prevent this, during the rehabilitation period the patient should learn to load his legs in stance equally. During weight-shifting he can learn to trust his ability to load his involved leg. Evaluation of the patient's performance of weight-shifting can be used as an indication for his ability to include his impaired body part in his posture and movement.

Methods:
We employed 2 methods to help the patient trust his ability to load his involved leg, stand symmetrically and shift his weight to his involved leg.
1. rigid ankle foot orthosis (Valenser Schiene)
2. functional electrical stimulation - hip, knee stabilisation (FES) (Stratec Device)

The performance of stance and weight-shifting were evaluated using 2 Kistler force platforms.

Results
Ankle foot ortheses and FES provides patients with better stability in stance and in weight shifting. The effect of current application not unambiguous. Some patients show very good results (protraction of the hip, perfect knee extension, taking more body weight in weight-shifting). Some show only little reaction to stimulation even if the current amplitude is increased. Therefore for applying FES more investigation is needed to find indicators for which patients benefit best from this method.

With these restrictions concerning FES, both methods are helpful in the therapeutic process to improve stability and mobility of the patient.

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A NEW STRATEGY FOR PROSTHETIC LIMB CONTROL

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Recent work at our Institute has focused on the development of multifunction myoelectric control systems. This research has resulted in the design of a new 5-state proportional control system based on the classification of myolectric patterns by an artificial neural network classifier [1,2].

According to the accepted myoelectric signal generation models, the myoelectric signal measured using surface electrodes is stochastic, [3]. This is due to the random nature of the pooled activity of the motor units within the pickup region of the electrodes. The model implies there is no information in the instantaneous value of the myolectric signal. Confirmation of this theory is given by Figure 1a. This is an ensemble average of 60 records of steady state myoelectric signal. It illustrates that the steady state myoelectric signal is indeed zero mean and has no apparent structure. There is a factor of 56 reduction in variance of the ensemble average waveform over the average variance of the individual waveforms in the ensemble. The reduction in variance agrees with that expected for an ensemble average of 60 random waveforms.

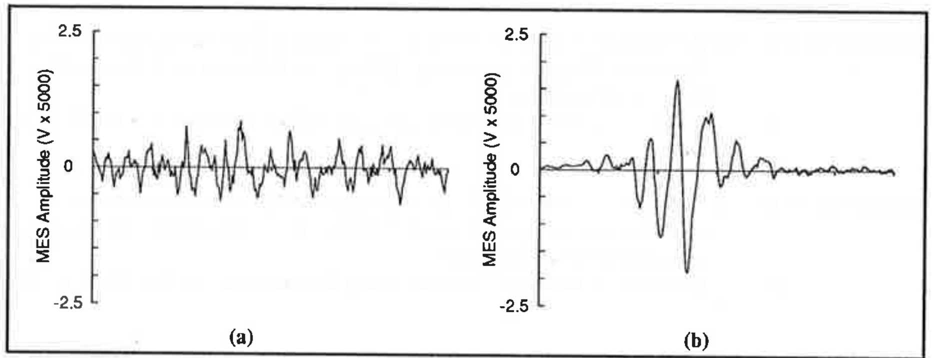

Figure 1 The ensemble average of sixty 300ms records of MES (a) steady state (b) initiation of elbow flexion.

Figure 1b shows the ensemble average of 60 records of myoelectric signal measured from the same electrode arrangement but taken during the initiation of elbow flexion. Figure 1b demonstrates that much structure is maintained in the ensemble suggesting that the myoelectric patterns of the ensemble records are reproducible and deterministic. The reduction in variance is only 6.9 rather than 60, which indicates there is a nonrandom component in these transient waveforms. This deterministic component is of short duration and occurs during the initial phases of the contraction. It has been observed that this structure is distinct for contractions which produce different limb functions, [1]. For the new control system the actual structure of the myoelectric signal over time is used to discriminate limb function. The result of this discrimination is used to control the selection of a prosthetic limb function. Once selected, the speed of the function is controlled in a manner proportional to the strength of the contraction. Tests on normally limbed and amputee subjects have shown that this method can provide the degree of classification accuracy necessary for multifunction control.

This work has resulted in a novel approach to the control of a multifunction myoelectric prosthesis. The new control scheme increases the number of functions which can be controlled by a single channel of myoelectric signal but does so in a way which does not increase the effort required by the amputee. The control signals are derived from natural contraction patterns which can be produced reliably with little subject training. The new multifunction control strategy has been implemented using an artificial neural network to classify myoelectric patterns. The system can be adjusted to the individual and retrained, if necessary, to meet the changing needs of the amputee. As well, the control system can adapt to slow varying changes in the control patterns. The state selection delay is below 250ms and the new control scheme integrates function selection with proportional speed control. State selection test verified that this method provided good classification accuracy with little subject training, [1]. Other tests have confirmed that the performance of the neural network based classifier is unaffected by small variations in feature values and that the network can continually adapt to changes in the pattern class features, [2]. This is particularly useful during subject training. During this time the user will become more proficient at using the control system. If the network is allowed to adapt to these training patterns it will also become more capable of recognizing the user patterns. This will allow the user to adopt an approach to generating the desired pattern classes which is comfortable and efficient rather than forcing the user to continue to use the same strategy which was used when the task was unfamiliar.

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HYPERTONE- THE MEASUREMENT SYSTEM IN CLINICAL DIAGNOSTIC OF UPPER MOTOR NEURON LESION

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Spastic syndrome with its characteristic phenomena interferes with motor functions and concomitantly with the performance of daily living activities. Different therapeutic procedures have been used to overcome spasticity, and different mechanical techniques for measuring spasticity described [1]. From clinical point of view different clinical scales for the assessment of spastic hypertonia have been used [2], which unfortunately offer only qualitative information. Moreover muscle stretch reflexes and passive muscle tone are assessed or the method of assessing function based on the natural progression of functional return is used.

For the assessment of properties of spastic syndrome measurement system "Hypertone" which can be used for the measurement of spastic knee or elbow has been introduced in clinical work [3]. 44 post-stroke hemiplegic patients (20 -60 years old, 2 to 8 months after acute stage) and 16 normal subjects have been observed and clinically classified with respect to the graduation scale from 0 (normal) - 5 (severe). Stretch reflexes of m. biceps and m. tricpes and passive muscle tone and other properties of spastic elbow in sitting position have been calculated. Results in 18 patients selected for the diagnostic procedures before and after the treatment have been compared.

The aim of our study was to determinate the relationship (or equality) between the clinical spasticity graduation scale and calculated classification represented in form of a decision tree [4,5]. The results obtained by the measurement system are presented in various informative graphical ways together with all necessary data of patients before and after treatment. Concerning the classification of 44 patients (class 0:3, class 1:4, class 2:10, class 3:14, class 4:10, class 5:3) significant graphical and numerical parameters before and after the treatment have been drown out. Decision tree was determined by Assistant program [5] on the set of 60 measurements (i.e. objects) four times. Each time 10-15% of objects were randomly taken out of the whole set for testing. About absolute average accuracy of the decision tree was 37,04% (i.e. 10 correct out of 27 objects to classify). However, when the criterion was relaxed so that classification in neighbouring classes was allowed relative accuracy was 92,59% (i.e. only 2 out of 27 were misclassified for more than one class). The most important numerical parameters according to which decision trees were built were: first local maximum of angle velocity during pendulum test, amplitude ration of the goniogram, relative passive range of motion, selectivity of voluntary flexion in elbow.

In this way the follow up of therapeutical progress after the treatment (i.e. lessening of tonic reflex activity, reciprocal inhibition, facilitation of voluntary movements) was enabled. Preliminary results show that hypertone enables the assessment of spastic phenomena using gravitational method (pendulum drop test), determination of volitional control (flexion and extension of elbow joint), reliable recording of static (tonic) reflex activity, but not of phasic component.

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LONG-TERM RELAXATION OF SPASTICITY IN HEMIPLEGIC PATIENTS BY ELECTRIC STIMULATION (ES)

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An unsolved question in Rehabilitation is the duration of spasticity relaxa= tion after ES. Alfieri, with clinical evaluation and Ashworth Scale (1982), and Alfieri and Vitale, with EMG (1982), showed a long term antispastic effect on pectoralis major muscle by ES of its antagonistic deltoid muscle.

We are at present engaged in a study on ES of extensor muscles of wrist and fingers of hemiplegic patients, directed to verify how much time the relaxation of spastic flexor muscles lasts after 30 real days of ES.

The study protocol provides an observation period of 5-7 months and states also that each patient is control of himself, in the sense that every patient receives both ES and conventional treatment; but they are randomly assigned to 2 groups: the one beginning with ES and finishing with conventional programme, putting between the 2 periods an interval of 4 weeks of no hand treatment; the other group vice-versa. More detailed information on the protocol will be pub= lished later on. The evaluation is made with the SRI-Test (Alfieri and Knutsson 1991), a new instrumental, numerical and diagrammatic method, conceived to mea= sure the resistance opposed by the spastic muscles to a rapid and passive stretch.

The results till now acquired (Feb. 92) on 8 patients seem to confirm that: i) the spasticity decrease by ES is always present in a more or less extent,de= pending on the initial grade of spasm; ii) the spasticity decrease is keeping 40-70 days after the end of the entire period of treatment; in the cases beginn= ing with ES, 3-4 months after the end of ES; iii) with the relaxation of flexor muscles, voluntary activity of extensors appears in some cases (unmask = ing), also when ES started more than one year after the stroke, during which constant physiotherapy was performed.

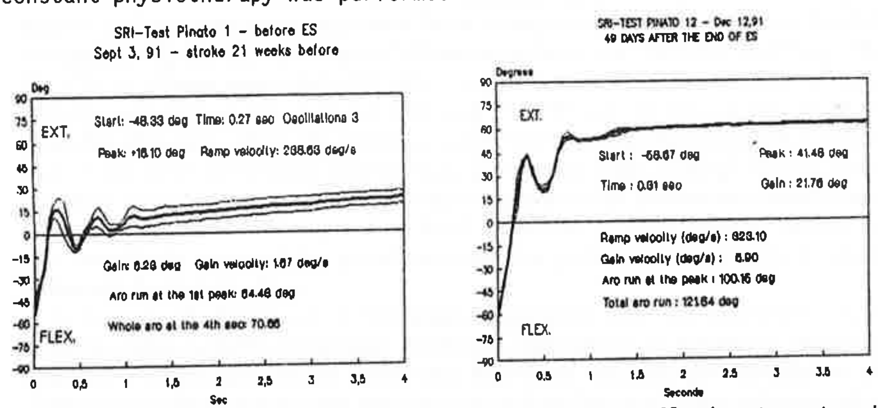

Fig.1.- Diagrams relating the passive movement from flexion to extension of the affected wrist of a hemiplegic patient, before a set of 30 session of ES and 49 days after the end of the treatment.

THE LOCOMOTOR OUTCOME OF STROKE PATIENTS AS ASSESSED BY POSTURAL SWAY PARAMETERS

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Investigation of postural activity and sway stability during standing requires information on the external forces acting on the body, if the swaying motion is to be interpreted through the basic laws of dynamics. Apart from body weight, these forces include the reactive foot ground forces. However, if measured on both feet together, these latter forces are summed, as the lower limbs are then treated as an equivalent single support. If, indeed, only application of the equations describing the motion of the centre of mass is required, the information thus obtained may be sufficient; however, if knowledge of the activity of each leg is desired, separate force measurements would be necessary on each of the legs. The latter case especially applies to post–CVA, hemiplegic patients in whom comparison of the leg activities, one in relation to the other, is of great interest.

Bilateral force measurements on the supporting limbs were made to provide a new representation of postural sway, which was implemented to evaluate post–CVA patients during rehabilitation. Postural sway was measured by two Kistler force platforms, which were collaterally installed for adjacent positioning of the left and right feet during standing. The foot–ground reaction forces in the vertical, anteroposterior and mediolateral directions were simultaneously monitored for both feet during the sway tests. From the measurements made, the following parameters were determined: (a) synchronisation of the forces acting on both legs; (b) vectorial patterns of sway activity of the legs in relation to each other (RSTFV); (c) timings of the oscillations; (d) amplitudes of the oscillations; (e) weightbearing imbalances between the legs; (f) total sway activity, defined to express the vector summation of the absolute values of the horizontal force component acting on both legs and invested to maintain equilibrium; (g) asymmetry, defined to enable comparison of the activities of both legs on the horizontal plane.

Fifteen stroke patients, aged 41 to 78 years, from 5 to 42 weeks after they sustained their first stroke took part in this study. To evaluate the results obtained, a control group of normal subjects, of approximately the same age average, was included.

The results presented in this study disclose the reactive force patterns acting on each of the legs of post–CVA hemiplegic individuals in postural sway while standing still. These forces appeared to be asymmetrical in nature, with a typical pattern for every individual, which generally differed from that of normal subjects. Sway activity and asymmetry were found to be considerably higher in the hemiplegic subjects compared with the normal controls.

The correlation of clinical features with biomechanical data and locomotor outcome (LO) produced interesting results. Surprisingly, we found that disorders in proprioception did not highly correlate with LO. Instead, hemianopia and superficial sensation were significantly correlated with LO. Irrespective of the clinical picture, a patient's tendency toward reduced total sway activity and asymmetry values in due course, and the possession or eventual adoption of a "regular" RSTFV pattern was associated with a favorable LO. It is thus suggested that these biomechanical parameters can be used for prognostic and monitoring purposes.

THE TRANSFER FROM SITTING TO STANDING IN SOME MUSCULAR DYSTROPHY PATIENTS

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Introduction

The Muscular Dystrophies are characterised by progressive muscle weakness and disability. The aetiology of the postural deformities which occur is a complex mixture of muscle weakness and compensation (Khodadadeh1985). The biomechanics is ill understood by most clinicians and this has, in the past, resulted in the patient receiving the wrong treatment.

We looked at 4 normals and 10 patients with different forms of muscular disease and various degrees of difficulty in rising from a chair. Each subject had been clinically examined. Then we took sagittal and coronal plane video recordings of the patients transferring from a chair. These video sequences were viewed in slow motion and all major manoeuvres made by the patients were analysed.

Results

During the normal process of getting up from a chair, the trunk is moved forward to bring the centre of mass over the support area. This is followed by extension of the knees and hips and relative plantar flexion of the ankles as the subject achieves an upright position.

Muscular Dystrophy patients employed more individualistic 'trick' manoeuvres which may or may not be related to the pattern of remaining muscle strenght.

This can be listed as follows:-

1. Initial use of a chair which has arms so that upper limbs may aid in pushing upwards.
2. Motion of the trunk forward on the hips to bring centre of mass over support area and use of subsequent momentum to assist in knee extension (used in conjunction with No 1).
3. Use of hands or knees or lower thighs to (a) initially stabilise hip and (b) by then extending at the elbows to produce a knee and hip extensor moment.

There are then a number of further options available depending upon strength of other muscle groups:-

a. Use of plantarflexors - used in conjunction with the hand/thigh complex to raise the centre of mass and allow extension of the contralateral knee and hip.
b. From the position of hands on thighs to release one hand and hyperextend the back whilst bringing the free shoulder and arm back. This results in a truncal rotation with extension of the ipsilateral hip bringing the trunk to a position where hip abductors can act to bring the trunk upright.

Conclusion

From this study two important principles of treatment emerge. The first, is the prevention of joint contractures in order that progressively weakening muscles may continue effectively to provide a balance of moments.

The second principle is that the clinician should develop an understanding and awareness of the compensatory mechanism that may be used.

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NON-INVASIVE ANALYSIS OF POSTURE

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Non-invasive and non-ionising three dimensional approach to biomechanics of posture has largely improved the number and the quality of information available for diagnostic and prognostic purposes. Various measurement systems, based on optical or opto-electronic principles of image capture, have been proposed to provide a specific set of results which are able to quantify, without any risk for the patient, the posture performance. Depending on the field of application (i.e. the analysis of rachis and the related effects induced by this disease on posture, the study of static equilibrium, the evaluation of the lateral and forward displacement during stepping, and so on), such modern measurement systems can reconstruct in three dimensions the positions of suitable landmarks and render the data as graphic displaying of markers and links (during both static and dynamic acts) or, in static tests, as a surface interpolating the back shape. In order to join the advantages provided by simultaneous measures of both the body topography and the spinal shape, the AUSCAN system has been largely upgraded to accept and process the additional data that are necessary for surface recontruction even during dynamic conditions. The present version of such a system, which was designed a few years ago for the analysis of the spinal repere points in static and dynamic conditions, is featured by:

- two couples of electronic shuttered cameras to detect a number of markers which are present on the patient's front and back;
- a specially developed processing unit, based on parallel computation, for the recognition of the markers and for the estimation of their centroids;
- a software support for three dimensional reconstruction of the coordinates associated with all the marked skeletal repere points and for back surface analysis.

The markers placed on various skeletal repere points and used for the analysis of spinal geometry are simple plastic semispheres coated with reflective paper. Conversely, the back surface reconstruction may be based on either the use of passive markers (in dynamic analysis) or, in case of a static topographic detection, on the joined effects of a laser beam, scanning the patient's back side, and the electronic shuttering of the cameras. The paper will be particularly focused on the description of such a technique for surface reconstruction.

MOTOR CONTROL OF GAIT AND ITS CHANGE DURING THERAPY IN SPASTICITY AND RIGIDITY

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Basic characteristics of the motor control during gait were studied in patients with disorders of the pyramidal and extrapyramidal nervous pathways. Analysis of the electrophysiological and biomechanical data was repeated after treatment with the electrical stimulation or drugs in order to assess the therapy.

Locally amplified surface EMG signals from 10 lower limb and trunk muscle groups were full-wave rectified, filtered with 3ms and averaged together with force basograms, obtained by measuring shoes with 9 force transducers each. Kinesiological data of about 30 strides per measurement were sampled with 200Hz and processed by the PC AT computer with Tecmar Labmaster data acquisition system.

Besides 15 healthy adults, 16 ambulatory spinal cord injury patients were studied. The measurements were also repeated after approximately 1 month of treatment with 4-channel electrical stimulation in some of them. Quadriceps muscles and peroneal nerves were stimulated bilaterally during gait, in order to provide the support and flexor response during the stance and swing phases of stride.

In 17 stroke patients the measurements were performed before and after 3 to 4 weeks of treatment with 6-channel electrical stimulation of gait. Hip abductors and flexors, knee extensors and flexors, peroneal nerve and plantar or elbow flexor muscles were stimulated in required sequences on the impaired side.

Gait of 6 patients with Parkinson's disease was measured without medical treatment and again after administration of neurotransmitter supporting drugs (Sinemet).

Patients with the impaired pyramidal nervous pathways due to spinal cord or cerebral lesions displayed slow, poorly coordinated gait, strongly dependent on the support of upper limbs or accompanying person. With reduced volitional control, they showed poorly expressed EMG patterns and coactivation of antagonists. In the spinal cord injury patients with preserved segmental and impaired suprasegmental control, the presumed released segmental reflex activity did not prevail, while the suprasegmental functional stretch reflex was not evident. The residual cerebral motor control, although disturbed and diminished, enabled their gait. The stroke patients however showed possible suprasegmental functional stretch reflex activity, suppressed segmental reflexes and reduced cerebral preactivation patterns.

After the electrical stimulation, both spinal cord injury and stroke patients walked faster with better weight shift to the lower limbs. Reciprocal activation was expressed better, the coactivation was diminished especially in stroke patients. They were also independent from the accompanying therapist. Central preactivation patterns were facilitated.

Dysfunctions of the extrapyramidal pathways due to Parkinson's disease reflected in markedly decreased stride length and increased double stance. Stride time was not as long as in dysfunctions of the pyramidal tract, yet these patients moved slowly with poor loading and push-off in the lower limbs. They did not require additional support. Sequences of muscle activation were disturbed and without expressed peaks towards the beginning and the end of stance phase. No significant muscle coactivation was observed. With no evidence of segmental reflex activity, both suprasegmental functional stretch reflex response and cerebral preactivation patterns were diminished.

During repeated measurements with medical treatment faster and longer strides were recorded. With reduced rigidity, both the biomechanical and EMG activity approached the patterns of healthy persons, displaying a direct dependence of the motor control on the level of neurotransmitter activity in these patients.

EVALUATION OF ENERGY EXPENDITURE DURING DAILY LIVING ACTIVITIES IN HEMIPARETIC STROKE PATIENTS

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Hemiparetic stroke patients often show early fatigue in standing and during deambulation.Consequently most of them have a sedentary life.

We studied 8 hemiparetic stroke patients(male;mean age 62 years; mean height 173cm;mean weight 73 Kg;mean time after disease onset 8 months)and a control group(5 subjects;male;mean age 67years; mean height 168cm;mean weight 70Kg).We used a portable instrument COSMED K2.This mini-device (weight 700gr)allows the evaluation of oxygen consumption(total VO2,VO2/Kg),heart rate(HR),respiratory frequency(RF),expiratory volume(VE),tidal volume(VT), oxygen equivalent(EQO2),oxygen percentage extracted from air(O2%), oxygen pulse(VO2/HR). We have verified the energy expenditure in sitting,in standing position and during deambulation.

We found(hemiparetic/control group):1)sitting:RF +26%; VE +13%; VT -14%; VO2/Kg +4%; VO2 +7%; EQO2 +5%; HR -3%; VO2/HR +7%
2)standing: RF +27%; VE +15%; VT -10%; VO2 +19%; VO2/Kg +11%; HR +4%; VO2/HR +12%
3)deambulation: RF +20%; VE =; VT -16%; EQO2 +5%; VO2 -2%; VO2/Kg -11%; HR -4%; VO2/HR -8%.

The oxygen cost of deambulation (Cal/mt/Kg)was 1.9 in hemiparetic patients and 0,754 in control group(hemiparetic/control +152%).

These data suggest:1) a different respiratory pattern between hemiparetic patients /control group during sitting,standing and deambulation.
2)an increase HR in hemiparetic patients during sitting/standing position.
3)in hemiparetic patients a lower cardiovascular work during deambulation,associate with a higher oxygen cost of deambulation.

Therefore cardiovascular training,as well as neuromuscular training,is an important goal of rehabilitation programs with the aim to avoid early fatigue during daily living activities and to improve recovery functions in hemiparetic stroke patients.

TRANSIENT NEUROMUSCULAR BLOCKADE: AN ASSESSMENT TOOL FOR PHENOL NEUROLYSIS

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A potential risk of phenol neurolysis of mixed peripheral nerves is the development of a deafferentiation syndrome resulting in severe, long-term pain. Phenol neurolysis however is an effective intervention in certain types of hypertonus and may provide functional improvement in selected patients. To justify the risk of the procedure a safe, temporary diagnostic intervention identifying those patients who will substantially benefit from a longer lasting neurolysis is important. A local anesthetic temporarily mimics the effect of phenol without creating long term deafferentiation.

Fifteen subjects (age range: 20-67 years) with unilateral or bilateral ankle plantar flexor hypertonus as a result of traumatic brain injury (N = 2), spinal cord injury (N = 4) and cerebrovascular accident (N = 9) were recruited from our in- and outpatient populations. Using a modified Petrillo technique percutaneous local anesthetic blockade of the tibial (medial polpiteal) nerve was performed. Marcaine 0.25% (Bupivacaine hydrochloride) was chosen for its rapid onset (2 to 10 minutes) and long-lasting effects (up to 7 hours). Using a Cadwell 7200 the tibial nerve was located percutaneously. Then 10 to 15 ccm of Marcaine were injected at the site using clonus as a intraprocedural parameter. Seven parameters were monitored before and after the procedure: Clonus, passive ankle range of motion (PAROM), active ankle range of motion (AAROM), ambulation time (AT) for ten meters, integrated resting EMG (RIEMG), integrated EMG response to fast stretch (FIEMG) and integrated EMG response to slow stretch (SIEMG) of the triceps surae. Patients were asked their subjective rating of the benefit of the block after 24 hours. All patients tolerated the procedure well and apart from minor soreness developed no complications.

Using a paired T-test, statistically significant changes following blockade were observed in Clonus: ($p<.0009$), FIEMG: ($p< .0009$), SIEMG: ($p<.0305$) and AAROM: ($p< .0509$) while change in PAROM proved to be a trend: ($p<.0686$). No significant change was observed in AT and RIEMG.

Nine patients rated the results as good and went on to phenol blockade. Six patients reported no improvement and declined phenol blockade of the tibial nerve. No correlation was found between the individual measures and the patient's subjective rating of blockade effects (two group T-test).

The data suggest that Marcaine bockade of the tibial nerve affects sensory and motor axons of the triceps surae and decreases hypertonicity. The absence of observed changes in ambulation time is explained by the relative short duration of the block and the patients inability to rapidly intgrate changes in tone and range to functional activity. The accuracy of the subjective impression may have been influenced by the time delay between the procedure and the follow-up interview.

All patients showed significant objective changes yet a relatively high proportion reported no subjective improvement and were not willing to undergo phenol tibial nerve blockade. This fact should urge clinicians to use short-term diagnostic blockade before subjecting patients to the risks of long-term neurolysis.

SYMMETRY OF GLUTEUS MAXIMUS ACTIVATION IN BELOW-KNEE AMPUTEE GAIT IN COMPARISON TO NORMAL

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INTRODUCTION

Often rehabilitation clinicians assume that improving the symmetry of an amputee's gait would inherently result in a more optimal gait pattern and that it would reduce the energy cost of ambulation (Hannah and Morrison, 1984). Winter and Sienko (1988) indicated that there is no scientific justification for this assumption. As a contribution to this debate the symmetry of muscle activation of the gluteus maximus was studied empirically both in normals and experienced below-knee amputees.

METHODS

Computer Averaged Surface Electromyographic Profiles (CAEPs) were determined of the Gluteus Maximus muscle of both legs in two groups of subjects at comfortable walking cadence using the instrumentation and methods described by Kleissen et al. (1989). One group consisted of 6 male below-knee amputees, age range 28 to 70 years (mean age 52 years) and mean mass 84.6 kg. These subjects had been using their prosthesis for longer than 1 year without complaints or problems, and had a normal walking function in daily life. Amputation had been performed after trauma, without affecting quadriceps and hamstrings. All subjects were tested using their own prosthetic device and shoewear. The normal group comprised 10 male subjects, age range 28 to 71 years (mean age 43 years) and mean mass 79.6 kg. Also the normals used their regular shoewear. Surface EMG data of at least 15 cycles of gait collected in the CAEPs.

RESULTS

Figure 1 presents the CAEP of the Gluteus Maximus of the right limb for the normal group. Figures 2 and 3 give the CAEP for the amputee group for the prosthetic and healthy limb respectively. Solid lines indicate the average profile, dashed lines represent the range for the standard deviation.

Statistical analysis of the data yielded, among others, the following observations:

- The symmetry of the profiles for left and right limb, expressed in the Pearson Correlation Coefficient, proved to be significantly lower for the amputee group (Wilcoxon test, $p < 0.05$).
- The total EMG activity over the gait cycle was significantly higher in the prosthetic limb than in the healthy limb. (Signed Rank test, $p < 0.05$).
- The EMG activity in the prosthetic limb during early stance was nearly twice the activity in the healthy limb.
- The EMG activity in the healthy limb was significantly higher than in the normal leg (Wilcoxon test, $p < 0.05$).

Fig 1: CAEP of Gluteus Maximus Right Limb, Normal (n=10)

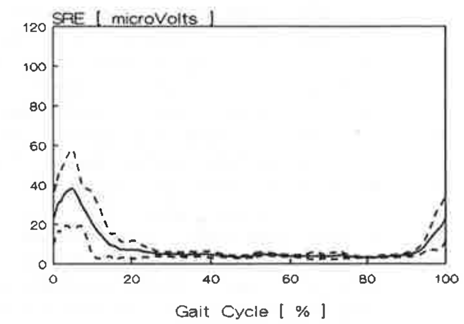


Fig 2: CAEP of Gluteus Maximus Prosthetic Limb, Amputee Group (n=6)

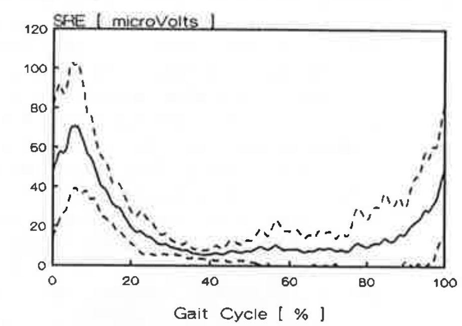
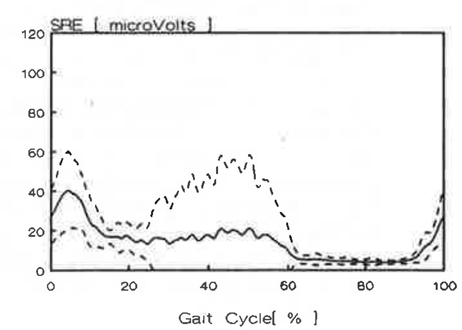


Fig 3: CAEP of Gluteus Maximus Healthy Limb, Amputee Group (n=6)



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KNOWLEDGE REPRESENTATION METHODS FOR THE MOVEMENT ANALYSIS ASSESSMENT IN REHABILITATION

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The problems faced by a clinician involved in motor rehabilitation require the evaluation of a large variety of parameters. Some of these are perceptive and can be directly inspected by the clinicians, other parameters cannot be directly observed even if very important in motor behaviours. The last parameters, as the angles at the joints or the velocity of the body segments, etc., can be on the other hand quantitatively measured through Movement Analysis (MA) systems.
At present the MA systems are based on an advanced biomechanical technology but it is necessary a strong effort to make their use simple in the clinical environment. This effort could reach also the objective to unify different rehabilitative approaches by the comparison of the quantitative data collected and analyzed through the MA systems.
The first step to achieve these goals is the integration of the MA outputs with the clinical reasoning: to make this possible it is necessary to single out entities and relations which can be specified and quantified through the MA systems and to define the correspondences among the entities subjectively perceived and the output parameters of MA. In this respect it is necessary to define clinical protocolos suitable to be used for the functional evaluation of the patient.
The next step for the integration of the biomechanical and clinical knowledge is the use of Artificial Intelligence methods for knowledge representation and formalization. This process can be subdivided in two main phases: first, the expectations of the clinician about a particular pathological behaviour are elicited and formalized in a theoretical structure; second, the results of the clinical experiment are compared with the contents of such a structure to search for common and different features. In this way it is possible to make objective the results of clinical reasonings.
Inside the European Project CAMARC (Computer Aided Movement Analysis in a Rehabilitation Context), together with the U.S.L. n.31 of Ferrara, we have conducted several sessions of interviews in which clinicians discovered togeteher with knowledge engineers what kind of relationships there are between the concepts expressed in clinical textual format and the scores of the evaluation tests. This inductive procedure has produced the first ground rules representing the clinical reasoning in rehabilitation for the class of hemiplaegic patients, and through the integration with deductive procedures, the bases have been posed to realize an explanation module of the clinical procedures.

METHODOLOGY FOR GAIT EVALUATION IN PATIENTS WITH TOTAL HIP REPLACEMENT

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INTRODUCTION
The present methodology for gait evaluation of patients with total hip replacement performed at Laboratory of Kinesiology in Ljubljana includes qualitative and quantitative methods. As our evaluation procedure has still some limitations, the patient's gait is also recorded by TV camera for visual documentation of patient's gait before and after surgery. In the abstract only the qualitative evaluation part is presented.

MATERIAL AND METHODS
The qualitative part of the method consist of visual clinical observation and manual tests. By observation, a physical therapist appreciates the patient gait and notes the observed anomalies in the 'Gait analysis full-body FORM'. This Form includes a set of characteristic anomalies of patient's gait. Anomalies are estimated according to severity: with one for small, with two for medium and by three for severe one. In the table, there is also a place for different comments on patient's gait in order to describe it more completely. The quantitative part of the gait evaluation methodology is based on the measurements of biomechanical parameters. The measuring system consists of ground reaction force system, 3D goniometric system for both hip, knee and ankle joints, and a system for measuring velocity of the patient's gait.

RESULTS
In the group of 31 arthrotic patients analysed before total hip replacement, limited flexion was observed in 84% of patients and inadequate extension was observed in 71% of patients, being thuse the most frequent anomalies. After surgery these numbers drooped in limited flexion to 32%, but the number of inadequate extension is still the same 71%. But on the other hand the total sum of scores for inadequate extension anomaly was reduced from 51 before operation to 35 after operation, what proves the global benefit of the surgery.

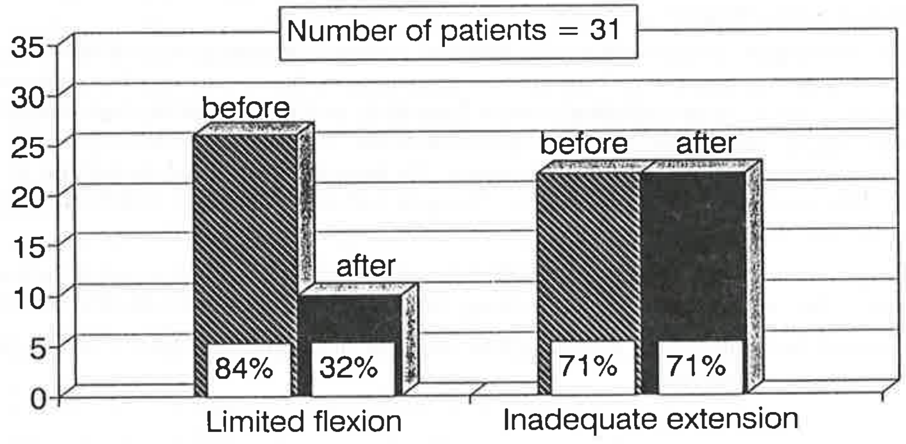

Fig. 1 The improvement of most characteristic hip anomalies due to total hip endoprostheses.

CONCLUSIONS
Out of 27 anomalies, that have been observed according to our methodology, limited flexion and inadequate extension were the most frequent ones. Beside these two, the following anomalies: Trunk: forward lean, lateral on affected-health side; Pelvis: trandelenburg, symphysis down; Hip: external rotation seems to be very important and give the major information on gait in patients with total hip replacement. These observations coincide with the results obtained by measurement of 3-D goniograms and is also reflected in ground reaction forces.

THE ITALIAN EXPERIENCE WITH THE LSU RECIPROCATING GAIT ORTHOSIS

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The RGO is an orthosis studied in Louisiana, at the State University, and applied to paraplegics during the '80s. Generally speaking it is long leg brace characterized by a double cable system which allows one leg to extend when the other flexes.

At the beginning of 1991 Officine Ortopediche Rizzoli started to supply the RGO, on a sperimental basis, to some paraplegics in cooperation with several Rehabilitation Center in Italy. Before this project there was little direct knowledge in our country of the device. The objective of our study was to personally verify the clinical improvements in subjects using the RGO and, eventually, make the device commercially available.

The sperimental experience has been based on few subjects, all complete traumatic paraplegics with mid dorsal lesions. They received a pre ambulatory training mostly devoted to increase their ability to maintain an erect posture and to increase the strenght of the upper limbs. After such a period they were fitted with the RGO and followed a gait training for few weeks.

Since the beginning we have fitted more then 20 paraplegics, males and females with complete traumatic lesion between T3 and L2. If compared to the traditional long leg braces the walking performances with the RGO are totally different in terms of gait pattern, energy consumption, indepence, daily use.

The first important result obtained is that after one week of training most of the subjects were not only able to don and doff the orthosis independently but also to walk for a distance never attained before using traditional devices. They were able to maintain an erect stance with no hand support and to perform some manual activities. The gait pattern is very smooth and the cadence, once they are trained, very regular. We have also analyzed some patients in our gait lab. The most evidente positive result, anyway, is that once at home our subjects still use their orthosis on a daily basis.

In the last months we have also applied the original LSU electrical stimulator designed to reduce the energy consumption of walking and give our subject a more efficient orthosis.

Some of the subject fitted with the RGO and some of the results obtained will be presented by means of a short video.

KOTZ CURRENTS, BRANCHED ANIMO ACIDS AND MUSCULAR STRENGTHENING IN THE ELDERLY

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Physiological aging leads to an irreversible reduction in muscular performance: an analysis of the muscular parameters recorded in normal subjects of various ages shows that average muscular strength has a maximum peak around the third decade, lasting up to the end of the fourth decade and then decreasing by 25% at the age of 65, 40% at 70 and subsequently at a rate proportional to age. The maximum strength peak shows a decline according to age at a rate of 6% per decade. The decrease in mean and maximum muscular force affects the lower more than the upper limbs. Associated with this phenomenon is a loss in muscle volume which seems to be reduced but not in proportion to the above-described decrease in strength peak.
We thought it would be interesting to use alternate currents with the aim of increasing muscular power in subjects who are no longer young. KOTZ currents increase muscular power (the product of strength multiplied by contraction speed) by increasing muscular diameter parallel with an increase in muscular mass and therefore muscular strength. The best frequency to achieve a maximum increase in strength is 2500 Hz, both with continuous sinusoids and interrupted sinusoids (in packets) in which the wave form consists of 10 m/sec of sinusoids and 10 m/sec pauses in continuous succession.
Electrically induced contractions provide the following advantages
1) they guarantee a selective training of individual muscles or a group of muscles;
2) they allow a more powerful contraction of the individual muscle than could be obtained with spontaneous effort;
3) the maximum muscular tension caused by electrostimulation can be prolonged for longer than is possible with spontaneous effort;
4) muscular training occurs without any effort on the part of the patient so the processes of central fatigue which occur in voluntary training are not involved;
5) some of the muscular fibres which are more difficult to involve in voluntary training since they are superficial are easier to stimulate when induced.

Clinical study

We decided it would also be interesting to associate electrical stimulation with the use of the drug FRILIVER, bearing in mind the activity of this drug in the recovery of muscular mass in liver disease patients.
Twenty patients, aged between 50 and 65, some of whom practised sports at an amateur level, were enrolled in the study.
They were followed for 40 days and received the following treatment: 10 patients were treated with Kotz currents at a frequency of 2500 Hz for 20 minutes; 10 received Kotz currents at a frequency of 2500 Hz for 20 minutes plus the administration of 5 sachets per day of Friliver equal to a dose of 50 g.
All the patients were tested with Cybex 340 at base time T0, T1 (after 20 days) at 'r2 (after 40 days). The test provides data on peak torque and total work at both 60° and 180- of angular speed.

Conclusions

The data obtained with the Cybex 340 tests show an improvement in peak torque and total work in both groups but to a greater extent in the group treated with Kotz currents plus Friliver.
These data show that the action of Kotz currents associated with the administration of branched amino acids give excellent results in maintaining muscular performance as age increases.

PREGNANCY AND SCOLIOSIS PROGRESSION: A PRELIMINARY STUDY

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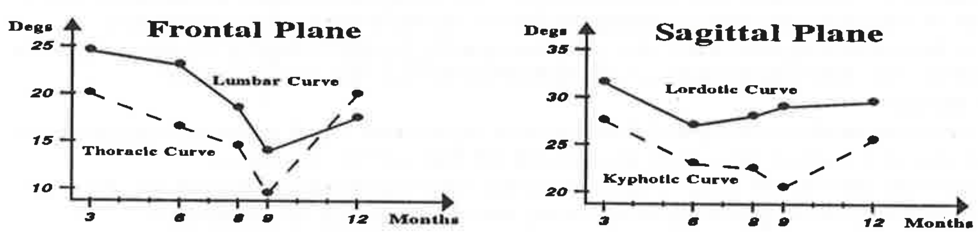
Introduction: The development of non-invasive and non-ionising measurement systems for the analysis of spinal deformities has greatly encouraged experimental experiences in many fields of application where the patient's state should be repeatedly assessed during time without inducing any risk of side effects. The progression of scoliosis in pregnant patients represents one of the most debated and not yet clarified topic in such a frame right because of the inherent hazard for mother and fetus in the traditional x-ray examination. The aim of this work is to present a preliminary investigation about the effects induced on the scoliotic deformity by pregnancy.

Method: The spinal angles of sixteen scoliotic pregnant subjects have been monitored by means of the AUSCAN system, a specially developed opto-electronic measurement device for 3-D analysis of human posture, from the third month on to three months later the childbirth. The experimental protocol included static and dynamic (lateral bending) measures of the spatial co-ordinates of several body landmarks on the patient's front and back side: zygomatic bones, mentum, acromions, sterno-clavicular joints, apophysis of sternum, ASIS, PSIS, knee joints, heels, spinous processes from C7 down to S3 every second vertebra, five equidistant points on the abdomen along the surface line joining the apophysis of the sternum with the pubis. Just a few among the large number of the coming out results concerning the spinal curvatures (projections on the frontal and the sagittal plane of the estimated spatial angles between the various segments of the rachis) and the realted rotations (projections on the horizontal plane of the spinal angles) have been preliminary considered to quantify the changes induced by pregnancy.

Results and Discussion: In order to summarise the results obtained, the changes of the surface measured Cobb and kypho-lordotic angles (i.e. the non-ionising approached estimation of the x-ray based parameters) and the associated trunk rotations during the course of pregnancy of one subject are shown in figure 1. Although the increased abdominal mass has induced a common tendence of all the patients to adjust their posture features by moving back the acromions, the changes of the measured spinal parameters follow a case by case evolution and not the course of pregnancy itself. In fact, significant aggravation of scoliosis after childbirth has been noted only in some patients and, given the personal strategy adopted to compensate posture during pregnancy, the type of parameter supporting such aggravation depends on the subject considered.



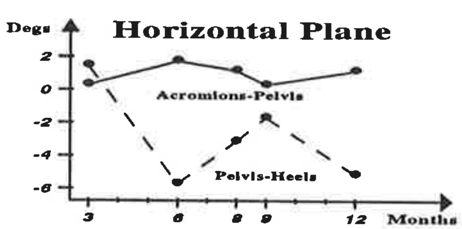


Figure 1 - Diagrams of the spinal angles as measured by the AUSCAN System. The data (black circles), referring to a 25 years old patient presenting a double curve scoliotic deformity, were captured during the 3rd, 6th, 8th and 9th month of pregnancy and the 3rd month after childbirth.

FUTURE DIRECTIONS IN KINESIOLOGICAL RESEARCH: HOW CAN WE MAKE USE OF NEW TECHNOLOGY TO FURTHER OUR DISCIPLINE

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Abstract. Use of a multifactorial approach in the assessment of kinesiological problems is discussed and examples are given, drawing from work of the author and others in related fields.

The focus of this discussion regards the use of a multifactorial approach for the analysis of traditionally based problems in kinesiology. Approaches that build on the foundation of functional anatomy but incorporate other measurement tools and new techniques are important for maintaining kinesiology as a discipline that furthers our knowledge of how movement is organized. For example, many current investigations address factors that are involved at the molecular level of joint articulation and organized movement. While these investigations may seem removed from studies that address the use of a particular prosthesis, or recuperation of normal joint function after an injury or an intervention, we, as kinesiologists, can ask questions that may direct this research into areas that relate to our own interests and endeavors.

When looking at the traditional textbooks, kinesiology is organized with respect to structure and function of joints and the surrounding musculature from a normal and pathological perspective (Fig 1a). However, in approaching problems today, we need to study these fields from a broader, multilevel viewpoint (Fig 1b) which incorporates components such as all connective tissue (bone, cartilage, fibrocartilage, ligaments tendons, blood), muscle, including the musculotendon unit and neuromuscular interface, and the neural component using analyses at both the micro and macroscopic level. This is true for any type of analysis such as those listed in figure 1b. For example, one could address joint motion from a variety of levels or perspectives, all of which are interrelated: a) global biomechanics of joint movement incorporating bone, muscle, transfer of momentum, energy and power transfer; b) specific interactions such as compartmentalization of muscle activity and its contribution to joint motion during a specific task or tasks, or c) "microscopic" biomechanics to examine strength or stiffness as a function of crossbridge formation during lengthening or shortening contractions. Similarly, with a focus on biochemistry, one could choose to examine a particular enzyme such as calcitonin gene related peptide (CGRP) from any of a number of different perspectives that relate directly or indirectly to joint movement. This particular peptide is released from small unmyelinated C fibres found in muscle, the periosteum, and other connective tissue in response to tension or shear, and acts as a vasodilator and mitogen, is interactive with prostaglandins, and has been termed a "noceffector" (Kruger,1988). Thus one might look not only at the particulars of its manufacture, structure and function, but at its effect on overall muscle or bone growth, how pain regulates its release or, alternatively, how pain in a particular joint or muscle may initiate adjustments in movement which modulate the ensuing CGRP release, and lastly, the effect of CGRP regulation on muscle or bone deterioration or recuperation. And while topics such as these may interest us only peripherally, or at only one level, as investigators, it behooves us to incorporate into our overall view of movement analysis these various contributing factors so that we may guide investigators that are more interested in pure molecular biology or biochemistry to address questions that are relevant to whole body movement.

With this in mind, then we can take one of three basic approaches in addressing

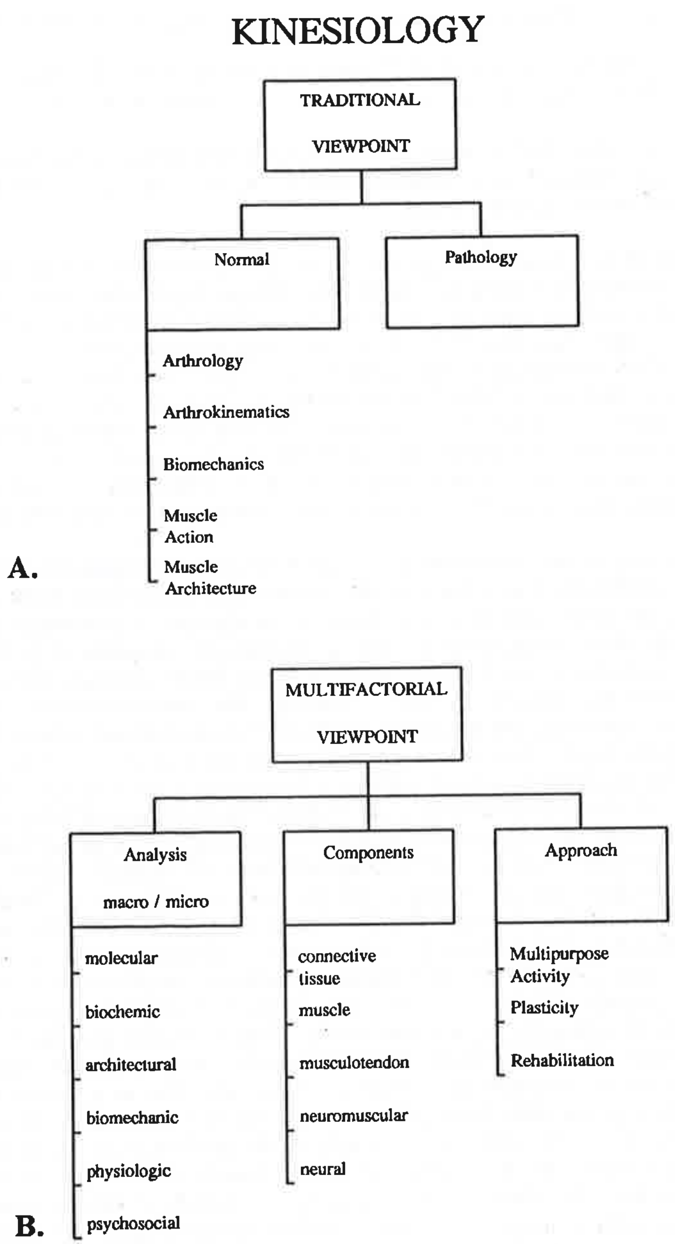

Figure 1. Two viewpoints of kinesiology

the problems of movement analysis from a kinesiological perspective and guide or direct questions to fit in with these approaches. The three basic approaches discussed here are that any whole biological system is 1) designed for multipurpose activity, 2) must be adaptive, and therefore have the property of plasticity, and 3) can be rehabilitated, using the property of plasticity, but in such a way that the organism retains the ability of executing multiple tasks.

Multipurpose Activity

In approaching problems from a multipurpose perspective, we have to make certain assumptions that are critical for understanding how any biological system - be it primate or echinoderm - organizes movement. Some of these assumptions are as follows: 1) a system design is imprecise and reflects the best solution for all circumstances that are likely to be encountered; 2) there exists, therefore, a weighting of factors relative to a given species and to each component within a functional unit; 3) thus, there is an interactive effect between the components so that the functional unit is *different* than the sum of its parts, and the operation of each functional unit is dependent on this interaction and may be different than other like units. A comparison between balance, stance and walking in quadrupeds and bipeds as well as organization of the shoulder and forearm in these two groups easily clarifies how factors two and three become important, while the notion of imprecision as being important can be addressed from the perspective of the variety of locomotor patterns available to a given organism and its ability to escape from or adapt to a particular environment. The ability to locomote forward, backward or sideways is to an animal's advantage, as anyone who has chased a cat or caught a frog knows. Catching a frog is as simple as putting your hand in front of it, as the animal can only jump forward, while getting the cat in for the night is another matter. And, while a river, lake or stream does not present an impenetrable barrier to a land vertebrate, with the exception of the walking catfish, land is a considerable barrier to a fish.

How does this viewpoint of multipurpose activity affect our understanding then of an organism's ability to organize and effect movement - from neural activity to joint articulation? We can begin by categorizing some principal functions for which each component of the system is responsible. Across the three main divisions, neural, connective tissue, and muscle, we see that each of these systems is important in maintaining posture, while each system subserves a different aspect of movement itself: the neural structures perceive the necessity to act, and organize movement either voluntarily or in response to a perturbation; the muscle and tendon effect the movement by the generation and transmission of force; the bone, ligaments, and cartilage articulate movement, while withstanding stress of force application.

If we use as a starting point (for the purposes of this talk) the mammalian primary motor cortex (MI), we can recall, that early gross electrophysiological mapping of this region suggested the presence of a fairly fixed, somatotopically organized map within MI, separating control not only of the face, arm and leg movements, but also of individual muscles found acting about a specific joint (Penfield and Rasmussen, 1950). With the advancement of more precise recording techniques, recent analyses of the motor cortex representation have forced a revision of this central theme, as many investigators have found that not only is the representation more distributed and overlapping, in that individual MI corticospinal neurons have functional contacts with several synergistic muscles (Cheney and Fetz, 1985; Buys et al, 1986), but muscle and movement maps generated by intracortical electrical stimulation of MI (Gould et al, 1986), neural recording (Fetz et al, 1980), and functional metabolic mapping (Colebatch et al, 1991) reflect the observed divergence of MI projections to multiple pools of motor neurons. In addition, it has been observed that distal and proximal muscle representation within a limb is often intermingled, and re-representation of the same muscle produces a mosaic where the same muscle is found several times, but each time, adjacent to different muscles (Waters et al, 1990); in addition, the size and location of the representation of the body parts was idiosyncratic. This variability between maps of different animals may reflect the role that experience plays in development and organization of somatotopic maps (Merzenich et al, 1988). Indeed, weighting of a particular muscle representation appeared to coincide with often repeated movements for an individual (McKinley, Waters and Dykes, unpublished observations). As the characteristics of overlap and functional distribution are observed in both motor (see above) and sensory (Wall and Cusick, 1986) maps, it is probable that this form serves as the basis for neural control of voluntary movement.

The notion that individual muscle representation is broadly distributed and therefore organized along the lines of synergistic action becomes important, because it affords the flexibility in muscle activity that we observe in natural voluntary movements. When observing movement at the muscle level, activation of a muscle, muscle compartment or a group of muscles together is task dependent. For example, in a recent series of articles, Loeb and colleagues (Chanaud and Macpherson, 1991; Chanaud et al, 1991 a,b; Pratt and Loeb, 1991; Pratt et al, 1991) have observed that timing of muscle or muscle compartment activity, and magnitude of activity appear to be task dependent not only when the speed of the movement is altered, but when the interaction of the limb segments is different. Thus increasing speed of locomotor activity increases activity in muscles such as the Tibialis Anterior and can add an additional burst of activity in superficial compartments of muscles such as the Tibialis Anterior and Semitendinosus (Chanaud et al, 1991b). In addition, changing motor behavior from a stride to a paw shake alters the way in which muscle groups work together, as the knee extensor, Vastus Lateralis, is coactive with the plantarflexors, Gastrocnemious and Soleus, during locomotion and with the ankle dorsiflexor, Tibialis Anterior, during paw shakes; in addition, compartments of bi-articular muscles such as the Sartorius show differential recruitment that is task related and may be dependent on the mechanical requirements of the task (Pratt and Loeb, 1991; Smith et al, 1985).

Adding to the complexity is a behavioral component, which is perception of the task goal. McKinley and colleagues have examined the task of landing from a jump with different environmental constraints. Although the task is the same under all conditions, that is, to take off from a platform and land with both feet together without taking a step forward, the timing of muscle onset latencies depends on the specifics of the task. For example, when landing from different heights, but on a surface that is always of the same stiffness, leg muscle onset is invariant (Thompson and McKinley, 1988; McKinley, 1992). By contrast, when one must land on surfaces of different rigidity, but from the same height, muscle timing is adapted to the surface to be experienced (McKinley and Pedotti, 1992). Thus, the context in which a movement must be performed, as well as the movement itself, can determine how muscle synergies are invoked.

To help us understand the way in which these muscles might need to work, are models of the way in which the articulations can move during a given activity. Former simple models of articulations as hinge joints have given way to newer complex multiaxial joint motion, as researchers have taken advantage of new imagery tools, such as the pulsed X-ray technique, to explore in detail the multiaxial motion of simple joints such as the knee (Bassett et al, 1990), or more complex ones such as the shoulder (Garg and Walker, 1990) or forearm and wrist (Caliebe et al, 1991) along with concomitant muscle forces. However, computer simulation models of these movements are frequently based on planar motion, rigid body mechanics and lumped muscle action. As computer simulation experiments are often used in concert with kinesiological studies to shed light on such features as the functional

aspects of postural adjustments that are related to voluntary movements (Ramos and Stark, 1990), development of appropriate models and concomitant muscle activity are necessary evaluation tools. In addition, as our knowledge base increases at all levels of the system, from neurological to joint motion, and from the microscopic to the macroscopic, factors such as changes in tissue composition with increased or decreased stress and concomitant force transmission and power flow changes, must be incorporated into movement based models.

Adaptation and Plasticity

Interrelated to this, clearly, is the phenomenon of adaptation or plasticity in the system that comes from growth, development, experience and learning, or from injury. Returning back to the cortical level, we know from the multitude of studies which have assessed the reorganization of the sensory cortex, that not only is the cortex capable of altering its representation of the periphery from experience (Merzenich et al, 1988; Jenkins et al, 1984), or injury (Jenkins and Merzenich, 1987; Wall et al, 1986), this response is age-dependent as well (McKinley and Smith, 1990). Recent experiments have also indicated that this reorganization occurs in the motor cortex as well, not only in non-human mammals (Sanes et al, 1990, Donoghue et al, 1990), but humans who have experienced amputation. Cohen and colleagues (1991), for example, have reported that muscles proximal to the stump show enhanced representation in the cortex contralateral to the amputation. While these phenomena are indeed interesting, we, as kinesiologists, need to focus the attention onto what this reorganization can mean with respect to movement itself. Some behaviorally relevant questions might include the following: what impact does this reorganization at the neurological level have on control at the muscle/joint level? Conversely, what impact does the change in biomechanics at the limb have on representation at the neurological level? Finally, what does the effect of pain and trauma have in the remodelling of the bone, muscle and surrounding musculature, and how does this relate to adaptation?

Rehabilitation

We must also utilize a multifactorial approach in addressing questions related to rehabilitation. Although we may be primarily interested in overall outcome, or recovery of movement, we need to be concerned with adaptation and recovery at all levels. The success of a particular intervention will depend not only on what we do at a macroscopic level, but also on what is happening at the microscopic level. For example, given recent advances on connective tissue response to injury (Enwemeka 1991), or effects of disuse atrophy or strength training on the structure of the musculotendinous junction (Tidball, 1991), can we determine when recuperative exercises should begin and what an appropriate waiting period might be, based not only on range of motion, roentgenography and manual muscle testing, but tissue biopsy or biochemical analysis as well? Interventions for the repair of injured skeletal muscle can also be assessed using both molecular and standard kinesiological approaches. The combination of the two assessments may give us further insight as to the ways in which muscle responds to specific therapeutic interventions; while assessment at the molecular level can clarify the discrete biochemical changes that occur, assessment at the level of movement can tell us the compensatory adjustments and adaptations that are occurring at the global macroscopic level which may impact on the changes seen at the microscopic level. Thus, assessment at the two levels is instrumental in giving us a more complete picture of plasticity and adaptation in the living system. And, although not discussed here, we must not forget the role that psychosocial factors play in not only successful rehabilitation, but in our understanding of their effect on learning, and development of skill.

In summary, let the unique viewpoint of kinesiology take a progressive stance to incorporate all aspects of analysis - be they molecular, gross functional assessment or psychosocial - so that through an integrated approach to movement analysis we may contribute significant questions to those workers in peripherally related fields, such that our overall goal of understanding movement from a global perspective is realized.

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VOLUNTARY REDISTRIBUTION OF MUSCLE ACTIVITY IN HUMAN SHOULDER MUSCLES

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Since shoulder muscle pain is related to muscle load, it is essential to reduce the muscle activity. Earlier electromyographic studies of the shoulder muscles have revealed that the pattern of muscle activity can change without any alteration of the arm position or the load. Is this due to voluntary redistribution of the muscle activity or is it a spontanious process? If it could be demonstrated that it is possible to control shoulder muscle activities voluntarily, and the trapezius muscle in particular, this may potentially be of great importance as a prophylactic measure. For instance, the possibility to train personnel by means of feed-back technique to use their muscles more "ergonomically", should be explored. There may also be room for advisory tests in certain, specifically demanding tasks.

To study this phenomenon, six subjects were investigated. Five shoulder muscles (the supraspinatus, the infraspinatus, the middle and anterior portion of the deltoids, and the descending part of the trapezius) were examined with electromyography in four different arm positions. By using visual feed-back technique, we found that the subjects could reduce and control the EMG activity voluntarily in the trapezius muscle. The ability to reduce the trapezius activity was most significant with the arm elevated 90 degrees from a vertical line, and with a straight arm. In this posture the activity could be reduced to 53% of the initial activity. This was not true for any other muscle investigated. There was a tendency towards an increase of EMG activity in some other shoulder muscles, particularly the infraspinatus, as a reflection of the decreased muscle activity in trapezius.

The findings may be related to relaxation from an initial overstabilization of the shoulder, or redistribution of load among synergists. It is suggested that the possibility to reduce trapezius activity may be of ergonomic significance. It is also noted that EMG trapezius activity may not serve as a universal descriptor of total muscular load in the shoulder.

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A NEW ISOKINETIC LIFTING TEST: CURVE ANALYSIS AND TEST-RETEST RELIABILITY AT DIFFERENT VELOCITIES

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Knowledge of maximum individual lifting capacity obtained with fast and safe ergometric tests allows us to prevent musculoskeletal injuries in manual materials handling (2). In a previous study (3) we found that the isokinetic lifting test at 0.73 m/sec is a reliable predictor of the maximum load a person in good health can actually lift. Aims of the present study are: a) to investigate the test-retest reliability of the isokinetic lifting test at different velocities, utilizing a Lido Lift device; b) to analyse the shape of the torque/time, torque/displacement and velocity/displacement lifting curves. Data on the test-retest reliability of this new device have been recently published, but only for one velocity and with regard to the peak force analysis (1).
Method- 7 healthy subjects (age 33±4 y, height 173±5 cm, weight 72±6 kg) with no history of back problems were studied after being fully informed about the aim, procedures and possible discomfort related to the experimental protocol. The test consisted, after standard warm-up exercises, of 4 consecutive isokinetic lifting trials. The average values of the best three out of 4 repetitions were calculated. The test was randomly performed at 4 different velocities (0.20, 0.46, 0.74 and 0.89 m/sec). The subjects were instructed to lift in the vertical plane a "T" handle placed at 20 cm above the floor up to 120 cm, exerting a maximal effort without jerks, from a bent back position with fully extended knees. Torque/time, torque/displacement and velocity/displacement plots were analysed. The test-retest interval was 7 days.
Results- Paired "t-test" revealed no significant test-retest differences at different velocities. The correlation coefficient of the pooled data was r=0.89. The peak force values at increasing velocity were (kg): 36.09, 35.59, 34.40 and 32.22. The ANOVA analysis showed significant differences between the peak force values ($p<0.001$). The analysis of the curves showed a reasonably good intra-subjects reproducibility and a inter-subjects variability. In the first lifting phase, as the force progressively increases, we found in several trials two peaks separated by a notch: we considered the first peak (reached before the isokinetic phase) as an artefact due to overshoot (impact torque) (6). The heights for reaching the isokinetic peak torque were (cm): 40 at 0.20 m/sec, 49 at 0.46 m/sec, 62 at 0.74 m/sec, 63 at 0.89 m/sec.
Discussion- This study indicates that the isokinetic lifting test performed at different velocities has a good reliability and the behavior of the test curves is quite reproducible in the same subject. The variability between the subjects is probably due to antropometric features (strength, height) and to personal lifting techniques. We found in accordance with the literature that peak torque values decrease with the increase of velocity of the isokinetic lifting test (4,5) and the height for reaching the peak torque increases with lifting velocity (5).

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DIAPHRAGMATIC DYNAMICS AND LUNG MOTION DURING PHYSICAL THERAPY

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Rapid recovery of diaphragmatic mechanical activity during the postoperative period is a main prerequisite to prevent lower part lung compartments from developing atelectasis. However the contribution of diaphragmatic activity to global ventilation is virtually unknown. Furthermore, there is no quantitative diagnostic access to motility of the lungs except for radioscopy. Also, a monitoring system for diaphragmatic activity and its relation to global ventilation would be useful for the control of physical therapy both for the physiotherapist and for the patient via biofeedback. Such a monitoring system is decribed here. Diaphragmatic force development is measured by a force probe and lung volume is measured simultaneously by core- to- surface electrical impedance.

On 16 high risk patients undergoing oesophageal surgery informed consent was obtained for the implantation of a needle shaped force probe in order to measure developed force within the diaphragm. The transducer, 9 mm of length and 1.3 mm of thickness, was implanted parallel to the pleural surface into the right diaphragmatic muscle dome. Two stainless steel wire antennae (10 mm of length) served as internal poles for core- to- surface electrical impedance measurement (four electrode mode). One external electrode was fixed to the fossa supraventricularis. The other external electrode was fixed on the homolateral inguineal skin of the patient. The combined setup monitors total excursion of the lungs (electrical impedance) and the active contribution of the homolateral ventral part of the diaphragm (developed force). Both signals were on line converted, plotted and stored with a laptop computer and commercially available A/D converter boards and software.

During recovery from relaxation the developed force signal indicated the onset of diaphragmatic activity. Maximum increase in developed force was seen after coughing and bronchial toilet. When analgesia vanished, developed force diminished. The relation: amplitude of the active force signal (F) / amplitude of the impedance signal (I) decreased to less than F/I=0.5±0.1. It proved to be difficult to motivate the patient to increase diaphragmatic activity by physical therapy alone. However adequate analgesia combined with physical therapy reencouraged the patient to enhance diaphragmatic activity to F/I=0.8±0.1. Some patients "non- responding" to physical therapy improved their pattern of ventilation when seeing their own force and impedance signals on the monitor.

Combined measurements of of diaphragmatic force development and of expansion of the lungs are able to monitor even minor postoperative impairments of diaphragmatic activity and of chest wall motion. In the group of patients investigated here we found a definite, mainly pain- dependent restraint of the diaphragm. Relief from pain subtantially improved the pattern of ventilation. It proved to be difficult to motivate the patient to increase diaphragmatic activity by physical therapy alone.

MUSCLE ACTIONS DURING DAILY ACTIVITIES: RELEVANCE TO SPACE TRAVEL

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It is well known that during routine flight missions astronauts lose muscle mass and strength. This is usually greater in the legs than in the upper body, and is greater in extensor muscles than flexors. On the Skylab 2 mission, astronauts lost about 25% of their leg extensor strenth. It has been proposed that this problem may be due to a lack of eccentric muscle actions when in space (Convertino, 1990). However, before this can be documented in 0G, baseline data are needed for activities of daily living in 1G. In this study five activities (walking, sitting down, standing up from a seated position, ascending and descending stairs) were studied. Three single joint muscles were analyzed - vastus medialis, tibialis anterior, and soleus.

Electrogoniometers were affixed to the lateral aspect of the knee and ankle. Data were sampled at rate of 500Hz for 3 seconds. Post-processing was conducted according to the logic shown in Figure 1. The amount of muscle activity in each phase was taken as a percentage of the total cycle time and the values from the two cycles were then averaged together. (This method is slightly different from that of Winter & Scott (1991), who considered muscles to be "on" throughout the gait cycle, i.e., the amounts of isometric, concentric and eccentric activity summed to 100%). In our approach the duration of concentric, eccentric and isometric activity, together with the "off" periods, sum to 100% of the cycle.

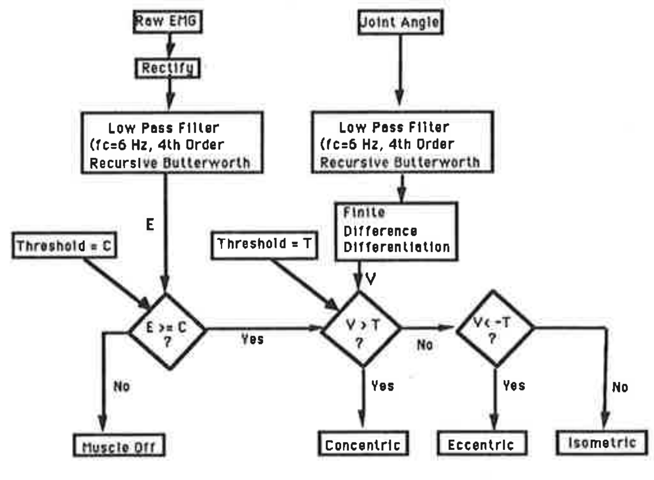

Figure 1. Determining the actions of muscles during excercise. Each type of muscle action (isometric, eccentric etc.) is expressed as a % of cycle duration.

The amounts of concentric and eccentric activity for each of the five conditions are shown in Figure 2. As may be expected, for vastus medialis, standing up and ascending stairs caused very little eccentric action, but large amounts of concentric activity. Other findings of the present study are that the amounts of eccentric or concentric activity were relatively constant for tibialis anterior (irrespective of condition). This is in contrast to soleus and vastus medialis, whose muscle actions were strongly dependent on the type of exercise - especially for sitting down and standing up. In the case of walking, the ratios of eccentric to concentric actions are in agreement with Winter and Scott (1991). These authors showed that, for overground walking, soleus exhibited about twice as much eccentric activity compared to concentric activity, whereas for vastus lateralis and tibialis anterior, the proportions of the gait cycle in which the muscles acted either eccentrically or concentrically were approximately equal.

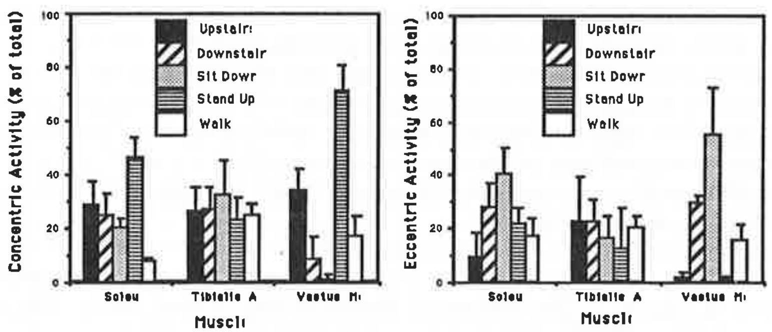

Figure 2. Amounts of concentric and eccentric activity during five selected conditions.

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LABORATORY GAIT MEASURING SYSTEM

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A simple gait measuring system has been built around a set of several pairs of footwear widely used by children from infants to adolescents. The footwear sole is equipped with four fixed conductive areas (foot-switches): on the heel, medial and lateral side, and the toe. A conductive pathway represents the second pole to all switches. Distance passed by the walking person is measured by a thread fastened onto the person's waist, and pulled over a pulley mounted on the axis of a rotatory potentiometer. Cabling connections to the appertaining electronics and a PC/AT computer consist of a small connecting box carried by the measured person, and a cable trolley covering the length of the walkway (15 m).

Data acquisition and processing software is based on the DOS 4.0, and ASYST V3.10 systems. The data acquisition program permits selection of a wide variety of parameters: number of channels to collect, number of channels to display, sampling frequency, preset or interactive control of acquisition duration, and optional transfer of the collected data to the hard disc during continuous acquisition. The basic data processing software performs analyses of the basogram and the distance-meter signals, yielding the main temporal and distance parameters of gait for separate strides (steps), and basic statistical analyses for several runs performed under the same conditions (about 30 strides). Different graphical and tabular presentations of the analyses are possible (Fig.1).

The described system represents the kernel of the gait measuring system. It is open to add optional measuring devices such as: goniometers, EMG, measuring gait assisting devices (e.g. crutches). The main objective in the development of such a system was its simplicity and low costs in comparison to more complex commercially available systems. Although it was primarily devoted to gait measurements in CP children, its use could be easily extended to the grown-up population.

UZR - Soca Ljubljana 23/06/92
Klinika za FMR
Enota za funkcionalno diagnostiko I

GAIT ANALYSIS

Name: BU2601.02 STASA
Date of measur.: 05/26/92
Comment: NORMAL VELOCITY CP STASA

Time and Distance Parameters

Number of strides: 42

	mean	std.dev.	c.var.(%)	min	max
Stride time:	1.35s	.11	8.01	1.08	1.67
Stride length:	.62m	.08	12.31	.39	.84
Velocity:	.46m/s	.06	13.48	.25	.66
Cadence:	89.58st/min	7.20	8.04	71.86	111.11
LEFT					
TIME step:	.62s	.07	11.35	.48	.81
stance:	1.01s	.06	6.37	.83	1.16
swing:	.34s	.08	23.14	.18	.52
dbl.supp.:	.39s	.02	6.11	.33	.44
LENGTH step:	.35m	.05	13.03	.23	.43
RIGHT					
TIME step:	.73s	.08	11.40	.58	.92
stance:	1.08s	.09	7.94	.91	1.27
swing:	.27s	.07	26.26	.13	.43
dbl.supp.:	.36s	.03	7.56	.29	.41
LENGTH step:	.27m	.05	17.34	.12	.38
SYMETRY					
TIME step:	.83	.11	13.21	.57	1.00
stance:	.90	.07	7.26	.70	.99
swing:	.77	.17	22.46	.37	.97
dbl.supp.:	.91	.06	6.36	.78	1.00
LENGTH step:	.75	.12	16.22	.50	.99

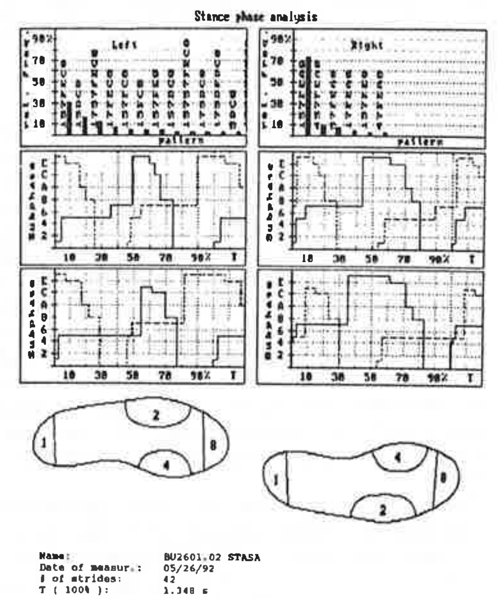

Fig.1 Example of an impaired gait analysis

A COMPARISON OF GAIT OBSERVATIONS AND GAIT MEASUREMENTS

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The physiotherapists of the Academic Hospital Nijmegen use a Gait-Analysis-List for evaluation of the gait of patients with extremity problems. In this Gait-Analysis-List the gait cycle has been divided into five phases: three in stance, and two in swing phase. For each body segment the movements can be scored during the five phases of gait into three qualities: incomplete movements, normal movements and excessive movements.
This study had two purposes, the first one is to determine the interobserver reliability of the subjective scoring. The second one is to compare the subjective scoring with quantitative measurements to study the agreement between the observations and the measurements. Quantitative measurements are performed by a new technique: a video mark computer movement analysis system (VMCMAS) which determines angles of hip, knee and ankle. In this study 12 patients with lower extremity problems have been analyzed by 9 physiotherapists and monitored by VMCMAS.
The physiotherapists scored the patients from video tape, but they were not allowed to play the tape slowly.
The interobserver reliability was assessed using the kappa coefficient. This statistical method was developed to assess the agreement among two or more raters after taking the effect of chance into account. The kappa coefficients ranged from 0.13 to 0.58. This can be classified as fair to moderate reliable.
Three consecutive steps of each patient were analyzed with VMCMAS. Each step was divided in the different five phases, such as in the Gait-Analysis-List is shown. Of every phase, the mean movement of the affected side was compared with the movement of the other side. When the difference is six degrees, the movement is defined at the affected side as incomplete or excessive. By comparing these results with the results of the physiotherapists, the percentage agreement between both methods can be determined. The percentages ranged from 50 to 90 percent. When the mean difference between the affected side and the other side is raised to 8 percent, the percentages agreement grew higher and ranged from 55 to 95 percent. When the affected side differed more than 8 degrees compared to the defined conditions, the movement is also defined as incomplete or excessive. The percentages agreement between both methods decreased and ranged from 40 to 90 percent.
The observations of the physiotherapists are fair to moderate reliable. The observations are based too much on the left - right symmetry. The percentages agreement between the observations and the measurement are higher when the difference between the affected side and the other side are higher. The observations would improve by more obvious definitions of the scoring-items.

NECK MOVEMENTS IN HUMANS: KINEMATIC ANALYSIS AND CLINICAL CORRELATES

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Introduction
The evaluation of neck movements in man has always been an challenge to clinicians and researchers (Mimura et al.,1989; Yoganandan et al.,1991) who have tried with invasive and non-invasive techniques to gain an accurate idea of spinal function. To assess the kinematic of neck movements a new system, which yields a 3D analysis evaluation of movements, has been used. The data have been compared with the clinical examination.

Material and Method
Fifteen healthy control subjects (age range: 19-50 years; average 31±8; 8 M and 7 F) underwent a kinematic and clinical evaluation (visual and goniometric) of range of motion (ROM) during flexion, right and left rotation, right and left flexion and extension, of the neck. Exclusion criteria were: history of head and neck trauma, and of head and neck pain. Nine patients were retested one week apart. The ELITE system (version 5.4; BTS, Milan, Italy; Ferrigno et al. 1985) developed using criteria from the computer vision technology, was used for the ROM analysis. This is a vision system with a two level architecture; the first level, implemented by hardware, receives the images from a set of TV cameras and recognizes in real time (50 Hz) the markers on the scenes. By a cross correlation filtering algorythm, the markers are recognized on the basis of their shape. The second level, implemented by software, performs the matching of surveyed markers' coordinates to the marker arrangement predefined in a model. Our model was made gluing hemispherical passive markers on the following places: C7, C5, C3, External Occipital Protuberance, Vertex (10 cm. from EOP), and right and left shoulder. A specially designed program tracks down each marker, frame by frame, reconstructing the coordinates. After this procedure a 3D reconstruction is carried out by means of a generalized triangulation algorythm starting from the images picked up by a couple of TV cameras. The clinical evaluation, carried out by two different examiners during passive movements, was based on a goniometric evaluation and a 4-point scale (4 = no reduction of the excursion; 3 = 25% reduction; 2 = 50% reduction; 1 = 75% reduction; 0 = 100% reduction).

Results
A significant positive correlation was found between the ROM evaluated with ELITE system on flexion and extension with clinical data obtained by the first observer, while this was not the case in lateral flexion and rotation and for the second observer in general. No positive correlation was found between kinematic analysis and universal goniometer evaluation. As we expected a positive correlation was found between visual and goniometric evaluation since the two methods compare passive movements. The test-retest difference in absolute value of ROM evaluated with the Elite system, the clinical evaluation and the goniometer has shown a good reproducibility of neck movements, the former being the expression of active and the latter of passive movements. There was a significant positive correlation between two separate evaluation with the three methods, the kinematic analysis having the highest coefficient of correlation. Elite: $r = 0.906$; visual examination: $r = 0.797$; goniometer: $r = 0.820$; $p < 0.005$).

Conclusion

To our knowledge the comparison of different methods for cervical spine ROM has been made only in patients (Youdas et al.,1991). In our hands, the kinematic analysis during active movements of the neck is not completely comparable to the passive movements evaluation clinically speaking. Besides, the kinematic analysis of ROM with the Elite system has shown a good reproducibility (Nappi et al.,1991) and it may be considered a useful procedure for the evaluation of active neck movements. We suppose that with this accurate technique, since we analyzed active movements, there may be factors interfering with the results (anatomic locations of markers, patients cooperation, etc.). Unilateral and/or bilateral painful conditions of the neck-head complex (cervicogenic headache, migraine, tension headache, whiplash etc.) may be an interesting model for the evaluation of head pain effect on neck motion.

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LENGTH-TENSION MODELS OF MULTICOMPARTMENTAL MUSCLES IN ISOMETRIC AND ISOTONIC CONDITIONS

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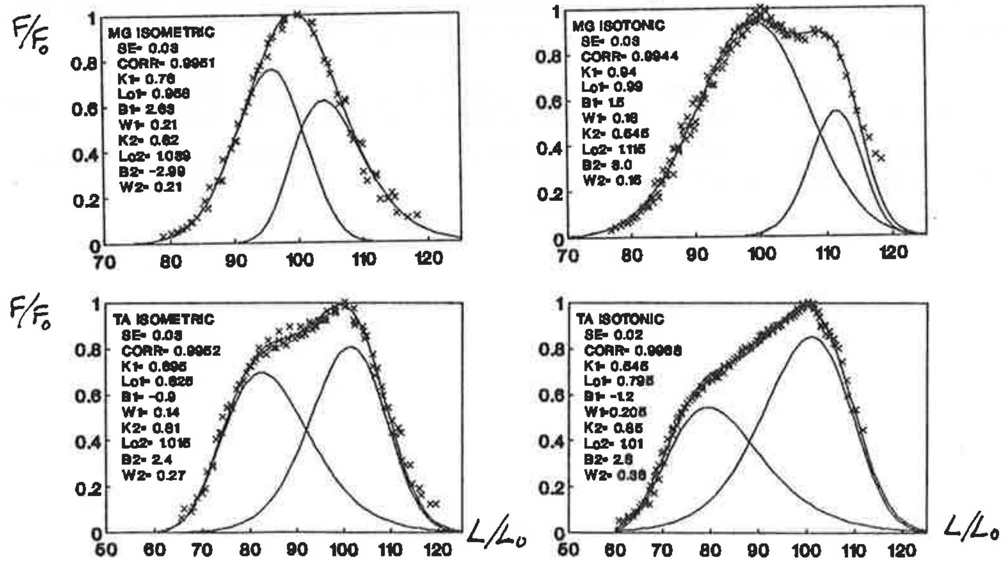
The length-tension property of muscle has often been regarded as one of the main components of its dynamic behavior. A model of the length-tension relation by Otten has been successfully used to predict the isometric tension throughout the elongation range of several muscles in the cat, but such attempts were unsuccessful in the Tibialis Anterior and Medial Gastrocnemius (Gareis et al., J. Biomech. In Press). Studies by Huijing et al. (Acta. Anat., 1988) and Stephens et al. (J. Morph., 1975) have suggested that these muscles may contain several compartments of muscle fibers, each with its distinct architecture and optimum length, providing a greater effective elongation range for these muscles. In light of these reports, a bicompartmental model of the active force in the T.A. and M.G. of the cat was developed, using a modified Otten model to describe the additive contribution of each muscle compartment. The model is of the following form:

$$F= \sum_{j=1}^{n} K_j \exp\left[-\left(\frac{(\epsilon_j+1)^{\beta_j}-1}{w_j}\right)^2\right]$$

Where ϵ_j is compartment strain, w_j is a width factor, β_j is a skewness factor, and K_j the compartment maximal force.

In addition, these length-tension models were validated by experimental data in isometric condition and also by using a novel isotonic or load moving method. This method can predict the muscular force when the muscle is allowed to shorten, and is more relevant to the dynamics of load moving muscle.

When compared to the standard isometric technique, the load moving method yielded a significant reduction in force, a shift in the optimum length, and an increase in the elongation range. These changes were verified and quantified by the bicompartmental length-tension model. Normalized examples are shown in **Figure 1**.



PREOPERATIVE ISOKINETIC EVALUATION IN SHOULDERS WITH INSTABILITY

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Introduction: Postoperative evaluation of unstable shoulders often shows a limitation of range of motion and a reduction of strength in some groups of muscles, especially in the external rotators. These deficiencies cuold be better evaluated with the isokinetic method, wich also, and most importantly, helps to understand if they already existed before the surgical operation or not.

Methods: We studied 15 patients with unstable shoulder by using a Cybex 350 isokinetic dynamometer and evaluated internal and external rotators with 30° abduction and 15° flexion arm. The test has be made during the preoperative period, with a bilateral comparison uninvolved/involved side. Each group of muscles has performed a test for strength (3 repetitions at 60°/sec) and a test for resistance (20 repetitions at 180°/sec).

Results and discussion; All examinated patients have shown a deficit of external rotators muscles, both in the strength and in the resistance test: the mean deficit was 19% (from 8% to 31%). A deficit of range of motion was also pointed out. These deficits, particularly the muscular one, are generally related to the surgical techniques (arthroscopic transglenoideal suture, Bankart and Latarjet-Bristow operation) and to the post-operative immobilisation; on the contrary, their presence before the surgical operation, pointed out using an isokinetic dynamometer, lets us think they must be related to a secondary injury of the musculotendineous structures, wich is probably caused by traction forces performed by the unstable humeral head.

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A RELATIONSHIP BETWEEN THE TRUNK AND THE ARM SWING IN HUMAN WALKING

Y.Ohsato, N.Matsusaka, R.Suzuki JAPAN

<PURPOSE>

The purpose of this study is to analyze the movements of the shoulder girdle, pelvic girdle and arm in walking. We hope also to clarify the relationship between these elements, and the relationship of the ground reaction forces to these movements.

<SUBJECTS AND METHODS>

Twelve normal males ages ranging from 22 to 33 years were examined in this study. The subjects were asked to walk at least 10 times in both free and fast walking on a 7 m walkway, in the center of which were two large force plates.

The angular changes of the horizontal rotation of the body and the arm swing were measured by rotational angle goniometers employing a gyrosensor.

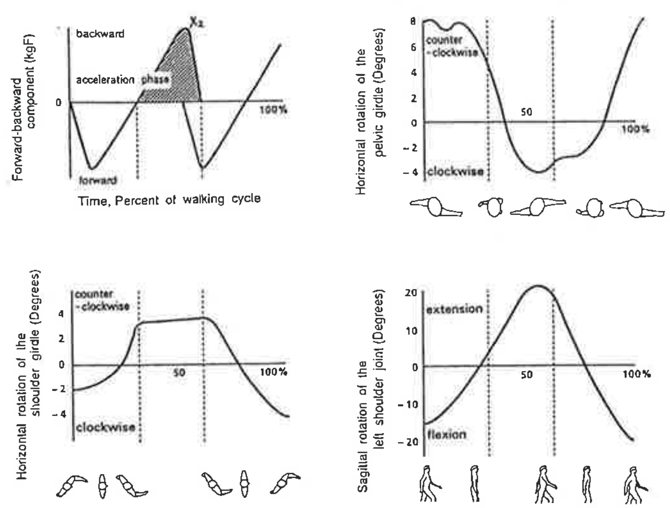
Three goniometers were fixed using special bandages: one in the interscapular region, one in the center of the sacral region and the remaining one on the left arm of the subject.

<RESULTS>

A representative pattern of the forward-backward component of the ground reaction forces, the shoulder girdle rotation, the pelvic girdle rotation and the left shoulder joint rotation during one stride of a 23 year-old male are shown in Figure.

The rotation of the shoulder girdle stayed almost stationary with the acceleration phase of the forward-backward component, and no significant difference was observed between the mean amplitude of the rotational angle of the shoulder girdle during free walking and that of fast walking.

However, the pelvic girdle and the left shoulder joint made these maximum rotation during the stationary phase of the shoulder girdle. Moreover, the mean amplitude of the rotational angles of the pelvic girdle and the shoulder joint in fast walking were significantly larger than these in free walking.



There were positive correlations between the maximum acceleration forces (X) in the forward-backward component of the ground reaction forces and the amplitudes of both the pelvic girdle and the left shoulder joint in all subjects. On the other hand, there was no significant correlation between (X) and the amplitude of the shoulder girdle in 10 of 12 subjects.

<DISCUSSION>

It is suggested that the arm swing is affected by the acceleration force of the contralateral lower extremity through counter-rotation of the trunk, keeping the muscular activities aside which also has a role in the arm swing.

Furthermore, it may be that the stability of the upper body is maintained by the shoulder girdle rotation becoming almost stationary in the acceleration phase and its amplitude not increasing with walking speed.

A CONTRIBUTION TO BIOMECHANICS OF THE LUMBAR SPINE

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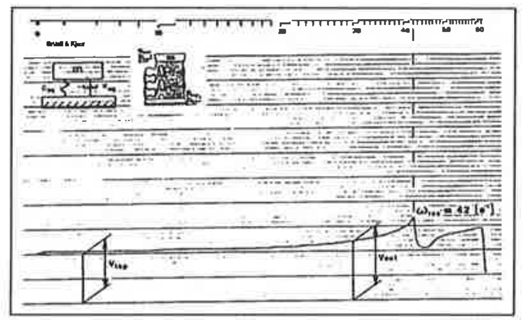
The theory of human rest and motion, as wheel as the theory of motion of all the living beings, is mostly based on experiments which include measurement in the evaluation of position, speed, acceleration, force, ect during various forms of activities. By those we primarily mean man's walk as a unique phenomenon of bodily propulsion, various sports activities (running, jumping ect) and movements a man does in various jobs (e.g. riveting, writing etc) . The most complex movements of the human locomotion system consist of various simple movements of its single parts and joints. Man's skeleton can be considered as a group consisting of one closed (the spine and the chest) and five open kinematic systems (the head, the arms, the legs).

Schultz and coll. (1982) have drawn our attention to the fact that the bodies of the vertebrae together with the intervertebral joint during the transmission of complex lads suppress axial dynamic loads.

In order to describe the nature of human body vibration, and in it the vibration of the spine we have chosen the example of forced vibrations generated by the kinematic excitement of the base. It must be said that one of the fundamental properties of biological materials is that we are functionally adaptable and that we are observing the dynamic segment, which is a part of a complex system such as the base. It must be said that one of the fundamental properties of biological materials is that they are functionally adaptable and that we are observing separately the dynamic segment, which is a part of a complex system such as the spinal column. By means of the vibrator "type 4802" we give the

stimulus to the dynamic segment. At their other side of the dynamic segment where the weight is we measure the speed response by means of the acceleration-meter. In the measurement of axial vibrations of the dynamic segment of the spine it has been token that the mass with which the dynamic segment has been loaded was 20 times bigger than the dynamic segment which is negligible.

Furthermore, analysing the measurement results, on a model had the equivalent constant of stiffness the 8.82 N/mm and the equivalent suppression of 0.196 Ns/mm at the resonant frequency of 42 s^{-1}. For the axial system with one degree of freedom its natural frequency results arehigh. As the examined dynamic segment is essentially shorter than the whole spine (axis of the skeleton), the natural frequency of the dynamic segment will be much wigher than the frequency at the axis of the skeleton. This has been confirmed by experiments.



→ **Fig. 1**: Output speed of the dynamic segment. Excitement speed v= 63,2 mm/s. Loading of the dynamic segment m = 5Kg

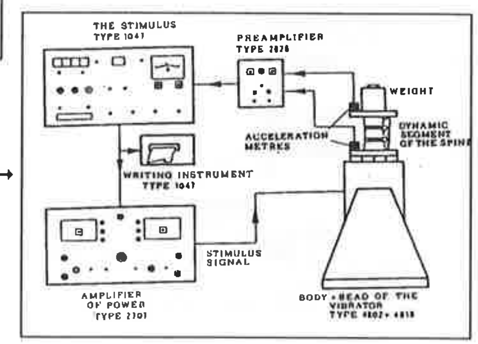


Fig. 2: Instrument for measuring the dynamic segment vibrations →

IDENTIFICATION OF COORDINATION INTERVALS IN THE UPPER EXTREMITY MUSCLES

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The aim of this paper is to define and determine coordination intervals (CI) in the upper extremity muscles fulfilling motor and stabilizing functions. The CI was defined as the time elapsing between the moment at which the resultant muscle force is applied to the surrouding by the ending of the biokinematic chain and the front of the functional current in the muscle actuating this biomechanism characterised by many degrees of freedom. The coordination interval is negative (NCI) when the beginning of the myopotential recording precedes the beginning of the force course, otherwise it is positive.

12 adult men, students of physical education, took part in the examination. Measuring instrument was in the form of a physical pendulum supplied with EMG apparatus. 3 endings were used, they were in the form of handles with mobility of 0; 1 and 2 degrees of freedom (DOF) in relation to the pendulum stem with the moment of inertia 574 kgm^2. The task of the tested subject was to push the pendulum with the hand using maximum force in such a way as to give it possibly high kinetic energy irrespective of the number of DOF of the ending. The force exerted on the pendulum ending of adjustable mobility and myopotentials of selected muscle groups were synchronically measured. Electromyography included: m. deltoideus - Dl, m. triceps brachii - Tr and palmar muscles (wrist flexors) - Fl and dorsal muscles of the antebrachium (wrist extensors) - Ex.

Negative (NCI) and positive (PCI) coordination intervals were presented in the form of histogram in Fig. 1. Occurence of PCI from 67 to 101 ms makes it possible to notice that if the pendulum ending does not require stabilization, the beginning of the stabilizing muscle functioning comes in the phase when the force exerted on the external object reaches high values. However, when force is exerted on the unstable ending, only NCI values (from -81 to -11 ms) were recorded, which appeared in a characteristic and stable sequence (Fig. 1). Their values prove that in the discussed case the moment of exerting force to the external object was anticipated providently by the stabilizing muscles Ex.

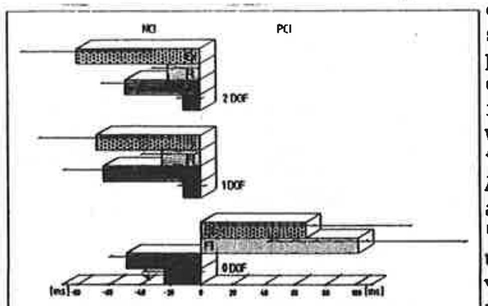

Fig 1. Negative Coordination Intervals and Positive Coordination Intervals: sequence of recruiting the muscles to take part in a motor task.

ACKONWLEDGEMENTS: The paper is an excerpt from an aricle: "Coordination intervals in the upper extremity muscles involved in motor and stabilization activities", submitted for publication to the Journal of Biomechanics.

MATHEMATICAL MODEL FOR GALL BLADDER DYNAMICS

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Data and models are used in systems analysis approach in Health Care System for decision making in medical diagnosis as well as clinical investigation. Gall stone is a major disease of the gall bladder, which is caused due to abnormal functioning of contraction of the gall bladder. Computer modelling may be helpful for dynamic analysis of gall bladder system, which may further be useful for applying biochemical and bioelectric control at appropriate time.

The present paper deals with the development of mathematical model for computer stimulation of bile in the gall bladder.

COÖRDINATION IN RISING FROM A CHAIR, USING DIFFERENT STRATEGIES

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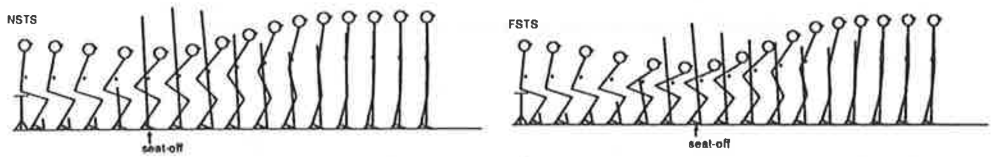
Rehabilitation medicine is directed to the restoration of motor-functions, aiming at the improvement of functional activities, such as walking, negogiating stairs and rising from a chair. In order describe the relation between joint- and muscle-function and the functional behavior in rising from a chair, we studied this latter activity, using integrated biomechanical and electromyographic analyses.
Ten young healthy subjects were measured. The saggital kinematics of the movement was measured using high speed film. Foot reaction force was measured using a force plate. Muscle activity of 9 muscles (GM, RF, VM, VL, ST, BFl, GA, SO, TA) of the right leg were recorded using surface EMG. From a standaardized sitting position subjects were asked to rise using two different strategies, in a standaardized tempo. Both Normal Sit To Stand (NTST) and the first Forward bending (FSTS) strategy were measured 5 times. Also for each muscle group Standard Isometric Contractions (SIC) were performed, recording reference levels of EMG.
Offline analysis included calculation of net-joint moments using a linked segment model. Smoothed (4 Hz) rectified EMG was used to reflect muscle activity, expressed in %SIC. All signals were ensemble averaged synchronized by seat-off.



The results show that no differences occur between the kinematics of knee and ankle, but just an, expected, higher flexion of the hip at seat-off. The higher net moment in the knee during NSTS is shifted to equal higher moments in the hip and ankle during FSTS. The slight differences in the EMG-levels of mono articular m. Gluteus Max., m. Vastus Med. & Lat. do not explain the difference in net joint moments. However, the EMG-level of the biarticular m. Semi Tend. shows a disproportionate increase, thus mainly being responsable for the compensation of the shift in net moments from knee towards hip. This result is in accordance with recent insights in the particular role of biarticular muscles. At the other hand, the moment shift from knee towards ankle cannot be associated with a specific increase of EMG-level of the m. gastrocnemius; the m. Soleus shows an equal increase.
This experiment shows the contribution of quantified surface EMG in an attempt to solve the indeterminance problem. As such it is necessary information to translate the (in this case more or less obvious) biomechanical results into muscle function. A practical conclusion is that a lower net moment around the knee in FSTS does not automatically mean that this stategy reduces the load on the knee extensor muscles.
It is thanks to the almost equal kinematics of NTST and FSTS that comparisons of this kind can be made. More complex comparisons can only be made using a sophisticated EMG>muscle>force model.

MODIFICATION OF MUSCLE FATIGUE PHENOMENA IN PARKINSON DISEASE

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INTRODUCTION

Structural and functional changes can occur in the characteristics of the motor units in Parkinson disease.

Some authors Edstrom L., [1970] analysing muscle biopsy material from these patients, showed different degrees of atrophy in white fibres, while red fibres usually maintained normal diameters and sometimes appeared hypertrophic. These changes could be explained by selective disuse of high - threshold phasic motor units and increased usage of low - threshold tonic motor units owing to akinesia and rigidity.

The purpose of the present study is to further investigate the effects of parkinsonism on some fuctional properties of muscle fibres by means of electrophysiological techniques: we refer to the possibility of ascertaining, during segmental isometric contraction, some central and peripheral aspects related to muscular fatigue in these kind of pathology, using power spectrum analiysis of surface EMG.

SUBJECTS AND METHODS

Subjects affected by Parkinson disease:7 cases, 4 males and 3 females, aged 55-67 yrs, with disease duration of 4-13 yrs., rated II-III according to Clinical Hoehn and Yahr scale were studied.

All patients assumed L-Dopa + D.I.(range 335-750 mg),(tab.1).

Tab.1

CASE	NAME	SEX	AGE Yr.	HOEHN & YAHR	L-DOPA	ILLNESS DUR.Yr.
1	C.O.	F	69	III°	335 mg	13
2	P.C.	F	58	II°	500 mg	7
3	M.A.	M	68	II°	750 mg	6
4	D.G.	M	62	III°	750 mg	8
5	V.P.	F	61	II°	375 mg	10
6	D.C.P.	M	55	III°	335 mg	4
7	V.V.	M	55	II°	250 mg	7

Data obtained in parkinsonian subjects were compared with those of 7 age-matched control subjects, aged 54-72 years.

Each performed an isometric elbow flexion at 70% of maximal voluntary contraction (m.v.c.) considered as the mean force value measured during 5 maximal contractions, each lasting 5 seconds.

Force was recorded by a strain-gauge connected to an oscilloscope for on line monitoring and to an A/D converter for storage on hard disk.

During recording subjects were sat in a chair, with their trunk upright, their upper arm abduced at 90°, resting on a table and fixed to the strain-gauge. The elbow was flexed at 90° and fixed to the same table. The forearm was maintained in an intermediate position between pronation and supination. Patients received visual feed-back of force. Contraction time was established as 60''.

EMG activity was recorded from biceps brachii(BB) and brachioradialis(BR) by silver electrodes placed distally to the end plate zone and 30mm apart. The signal was filtered at 1-512 Hz,digitized with a sampling rate of 1024 Hz and stored on hard disk. Analysis was performed off-line. Segments of EMG signal, lasting 500 msec. were sampled, and for each of them root mean square (RMS), mediane frequency of power spectrum (MDF) and products of RMS and square root of MDF (corrected RMS) were calculated. The last value represents an index of the generalized firing rate as described by De Luca [1979] and Kranz et al [1985].

RESULTS

Control subjects(fig.1): maximal force range was 14-29 Kg, MDF initial range values resulted 75-95 Hz in the BB, 70-100 Hz in the BR. MDF values during the exsercise decreased in 5 out of 7 subjects for the BB, in 6 for the BR. Average MDF values in the group decreased in both muscles. Corrected RMS for the BB increased in 2, proved constant in 4, decreased in 1 subject; for the BR increased in 5, appeared constant in 1, decreased in 1 subject.

In parkinsonians (tab.2), strength was in the normal range in 4 subjects, reduced in 3. All the subjects but 2 succeded in maintaining the isometric contraction without significant oscillations during the 60" of the test. One subject presented wide oscillations of strength after 15" of contraction, another gradual decrease from 15 to 11 Kg.

CASE	STR	MDF(i)		MDF(f)		C.RMS		ILLNESS
	Kg	BB	BR	BB	BR	BB	BR	DURATION
1	5	70	75	70	70	=	=	13 yrs
2	9	70	80	70	80	=	=	7 "
3	10	60	60	60	60	=	o	6 "
4	14	70	80	70	80	<	>o	8 "
5	15	65	65	65	60	=	=	10 "
6	20	65	75	55	65	=	>	4 "
7	30	90	90	85	85	=	=	4 "

Tab.2: = : constant; o: oscillating; >: increasing;<: decreasing; i: initial; f: final; C.RMS:correct. root mean square; MDF: median frequency.

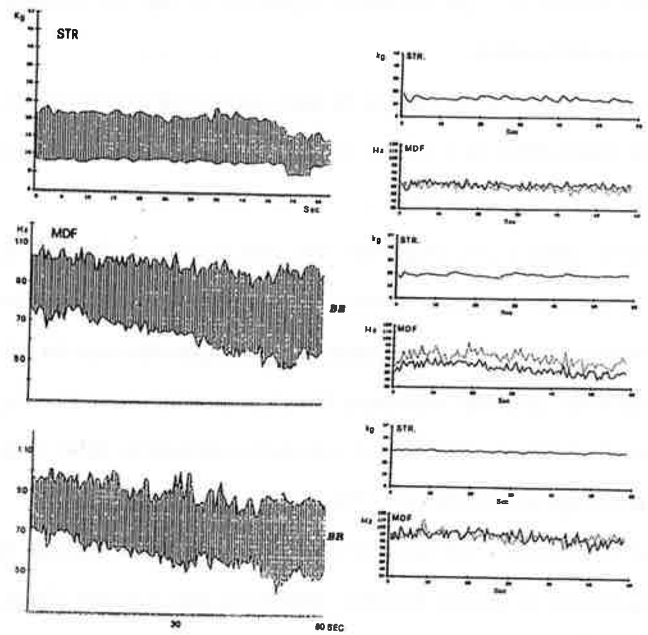

Fig.1
Trend of strength (STR) and median frequency (MDF) during the exercise:left side range values of control group; right side three example in subjects 5, 6,7, are reported.In ordinate STR expressed in Kg and MDF in Hz; in abscissa time in seconds.

Initial values of the BB muscle MDF resulted clearly reduced in 3 patients, slightly reduced in 3 other, normal in 1; the same parameter for the BR muscle appeared reduced in 2 patients, normal in 5. During the exercise the MDF values resulted constant in 6 patients both for the BB and BR muscle, with no significant decrement resulting, three example are reported in fig.1. One patient with considerable strength oscillations showed parallel modification of values, but with final MDF values superimposed with the initial ones.

The corrected RMS showed constant values in 6 and decreasing in 1 for the BB, constant values in 5 subjects, increasing values in 2 for the BR.

DISCUSSION

Different factors can modify the muscle structure and function in the affected extremity in parkinsonism, namely whether the type of muscle is postural or not, the duration of the disorder and the degree of disuse and disability.

Histochemical studies of skeletal muscle in man inactivated experimentally through point fixation [Karpati and Engel,1968] or clinically in conjuction with plaster [Patel et al,1969] have shown atrophy of both types of muscle fibres.

Rigidity undoubtly play a very important role both directly or through some mechanical aspecific process. Passive stretching for instance is known to have an effect on the size of muscle fibres and the co-contraction or lack of relaxation of the antagonistic muscles, during movement that would stretch agonistic muscles, counteract the atrophy which would be expected to result from the low degree of tension obtained during voluntary contraction. Other concomitant factors can be vasomotors disturbances, and pressure effects upon peripheral nerves.

The use of some electrophysiological techniques as here reported offer the possibility to analyse these modifications of muscle structure (diameter) and function (fatigue phenomena). This possibility is easily understandable considering that spectral parameters are directly dependent on the myolectric conduction velocity, which in turn is influenced by the muscle fiber

diameter. The ratio between the mean frequency value of spectral parameters and the muscle conduction velocity are almost constant

in the force range of 20-80 % of massimal strength [Broman et al.,1985].

During fatiguing contraction, both the parameters of MDF and conduction velocity (CV) decrease, even if this decrement is greater for MDF than CV, probabily for the effects on myoelectral parameters,of changes in the firing patterns of the active motor units which on the contrary do not influence CV values [Basmajian and De Luca,1985].

In parkinsonian patients a reduction of mean values in the area of white fibres has been shown by Edstrom [1970], by means of histological techniques. Compared with the control material in which small white fibres were completely lacking, the material from patients with parkinsonism exhibited up to 60 % of small white fibres. The maximal atrophy values were found in patients with a longer history of the disease and with very severe akinesia. On the contrary the red fibres exhibited both normal size or also tendency to hypertrophy,expecially in patients characterized by pronounced rigidity,whilst when akinesia was prevalent and rigidity slight, red fibres could also show low mean values of cross-sectional area. This differential effect on red and white fibres of the disease, therefore seems to be dependent on certain qualities of the mobility disorder such as the degree of increase of tone or of akinesia.

The electrophysiological techniques here reported allow us to interpret our data in a similar way with respect to the previous histological reports. The low MDF values observed in some subjects both for BB and BR may be related to a diameter reduction of white fibres,which would depend on a selective disuse of the high treshold motor units. Another characteristic of the MDF trend during the contraction in the parkinson group, is the lack of significant decrement both for the BB and BR, even for those cases where high strength has been exerted. This phenomena may be justified by two different processes: first the lower MDF initial value consequence of the lack of normal sized white fibres, and secondarily the increased chronic usage of the low threshold tonic units consequent to the rigidity.

In conclusion these data show the utility of this kind of EMG power spectrum analysis in assessing muscular structural and functional modifications, in parkinsonian lesions. These modifications could have some role in the functional deficits of these patients.

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TOWARDS A CRITERION FOR MINIMAL MUSCLE FATIGUE IN INTERMITTENTLY STIMULATED QUADRICEPS DURING CYCLICAL MOVEMENTS OF THE LOWER LEG

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Muscle fatigue is one of the main limiting factors when using Functional Electrical Stimulation of muscle (FES) for restoration of lower extremity functions as prolonged standing and ambulation. Therefore, both the muscle stimulation method as well as the control of stimulation should be optimized for minimal fatigue. In this study we concentrate on a fatigue criterion for optimization of control for a given stimulation method. As a test case we consider intermittent stimulation of quadriceps during cyclical movements of the lower leg. From this study, we hope to derive a criterion, which is relevant for FES-induced ambulation with minimal fatigue of the stimulated muscles. In this paper, an inventory is made of the stimulation and movement parameters dependencies of fatigue, described sofar. Furthermore, a way of using these relations in the optimal control of stimulation is proposed.

Muscle fatigue can be characterized as the time-dependency of the torque-output (maximal torque, averaged torque or torque-time integral) of the muscle during a fixed stimulation protocol. For intermittent as well as continuous stimulation this time-dependency appears to have an exponential decay to a steady state fatigued value ([1],[2],[3]). Both the time constant of the decay τ and the value of the steady state fatigued torque relative to the unfatigue torque (relative steady state torque M_r) appear to depend on stimulation parameters as well as on parameters of the cyclical movement. Several of these relations have been reported for isometric or isokinetic contractions [1]-[3]: e.g. τ and M_r appeared not to depend on recruitment level for continuous stimulation (under isometric condition) [1]; when considering maximal torque per cycle they do depend on duty cycle during intermittent stimulation (under isometric and isokinetic conditions) [1-2], and on the velocity of movement [2]. In most cases, τ and M_r appeared to be correlated. Therefore, maximizing M_r tends to maximize τ.

For optimal control of a cyclical movement, the stimulation and movement parameters should be chosen such that τ and M_r are maximized, while satisfying the functional movement conditions. If τ and M_r are relatively high, the fatigue adaptation rate can be small. If Mr approaches 1, fatigue is no limitation to perform the function as long as necessary.

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FORCE-EMG RELATIONSHIP IN STIMULATED PARALYZED MUSCLES DURING FATIGUE

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The activation of paralyzed muscle by surface FES is accompanied by the appearance of the Compound Muscle Action Potential (CMAP), which can be detected by surface EMG. The CMAP is preceded by the stimulus artifact, which takes the form of a repetitive signal. As usage of a stimulator of the constant-current type necessitated suppression of the stimulus artifact, we designed a stimulator artifact suppressor. The existence of any residual artifact after suppression was studied and characterized by means of experiments performed on quadriceps muscles, which did not respond to electrical stimulation. This was made on fresh cadavers as well as on one flaccid paraplegic subject, in whom no muscle contraction was evoked as a result of stimulation. The results obtained revealed a residual artifact with an intensity of less than 3 mV and a recovery time of less than 3 ms. The residual artifact obtained was found invariable during stimulation, enabling us, in the fatigue experiments on activated muscles, to subtract this signal from the measured EMG, in order to obtain an artifact free signal.

The situation, in which the only existing muscle forces are those of the activated quadriceps, renders a mechanically determinate system and allows the non-invasive determination of these forces from the externally measured torques. The time course of the quadriceps force obtained in this study indicated a loss of force-generating capacity, starting in the very early stages of activation.

The EMG signals obtained were sampled at 5000 samples/sec, simultaneously with the mechanical signals. In the group of four paraplegic patients studied, activation of the quadriceps muscles resulted in decreasing the amplitudes of the M-wave signal as fatigue progressed. Since a fatigue experiment lasted about three minutes, and due to the high sampling frequency of the EMG signals, the latter were sampled intermittently, i.e. in segments of 100 msec and in a sampling cycle of 1 sec. Under these conditions, between two and three M-waves were obtained in each segment. The peak-to-peak (PTP) amplitude at a given time was taken as the average of the PTP's obtained within each sampling segment. The normalized PTP and force were plotted together and an exponential fit was made between the two parameters to describe their parallel decay.

Our results of force-EMG correlation revealed a correspondence between these two parameters in the first 60%-70% portion of the fatiguing process, as expressed by the force. It is believed that this relationship obtained may be used practically in the future for non-invasively monitoring the force developed within the muscle at functional levels and, accordingly, for controlling the stimulus parameters required.

CONTRACTILE PROPERTIES OF STRIATED MUSCLES BEFORE AND AFTER ISOMETRIC FATIGUING EFFORT IN MITOCHONDRIAL CYTOPATHIES WITH COX-DEFICIENCY

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Fatigue-induced changes in twitch response of striated muscle have been estensively investigated in normals and neurological disorders such as neuromuscular transmission failure and dystrophic myopathies.
In this study we examined the twitch response of tibialis anterior muscle before and after prolonged isometric effort in ten normals and five patients affected by mytochondrial cytopathy with partial c-COX-deficiency.
Conventional electromyographic and electroneurographic studies detected myopathic findings in only two patients while the remaining patients presented normal electrophysiological parameters. Furthermore subtetanic and tetanic repetitive stimulations of peroneal nerve did not show differences in M-wave between patients and normals. The twitch response of tibialis anterior muscle was investigated according to Vandervort et al. (Exp. Neurol., 1976). In particular, baseline and potentiated twitch responses were examined by evaluating contraction time (CT), 1/2 relaxation time (RT) and twitch torque (TT), in all subjects. Furthermore, twitch changes provoked by maximum isometric effort lasting 60 sec were evaluated.
Baseline and potentiated twitch responses of all patients showed significant 1/2 RT slowing and TT reduction. Fatiguing efforts significantly increased differences between patients and normals in both 1/2 RT and TT values. These results suggest that contractile more than electrical abnormalities of striated muscles are initially present in mitochondrial cytopathies with c-COX deficiency. Reduced ATP availability seems to be the main factor responsible for the increase in twitch response abnormalities subsequent to fatigue in mitochondrial cytopathies.

ADJUSTMENT PROCESS WITH FATIGUE DURING BRAKING OF FAST VOLUNTARY MOVEMENTS

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Single-joint fast movements are characterized by a triphasic electromyogram (EMG). The agonist muscle provides the driving force for setting the limb in motion and then, in sequence, an antagonist and second agonist burst occur. Zouhri and Le Bozec (1988) have shown that the capacity of the braking process during elbow extension after a local muscular fatigue in antagonist muscles led to changes especially in these muscles when subjects have developed a "motor program" for this particular movement and were trained to produce this basic task. The process of transforming the pattern into an appropriate pattern of muscle activity after fatigue during fast movements with low amplitudes took place progressively. Therefore, we can consider that for movements with large amplitude and different duration, the nervous system can use sensory inputs to control parameters of motor activity, and then to correct errors in trajectory. In this study, this hypothesis will be tested during elbow extension movements with large (80°) and small (20°) amplitudes.

Subjects performed extension movements of the right elbow in an horizontal plane. Surface EMGs from triceps brachii (TB), biceps brachii (BB) were recorded simultaneously with the mechanical variables. In order to obtain a constant state of fatigue, an experimental protocol (Le Bozec and Rougier, 1991) was used. Analysis of onset of muscular activities which were defined in relation to the onset of movement and to integrated EMG (IEMG) was made 1) without fatigue during the last ten movements, 2) with fatigue during the first movement.

Results show that there is no difference between the IEMG or the onset of TB activity before or during fatigue for the two tested conditions (20° or 80°). With fatigue, IEMG and onset of activity BB values measured during the first 20° amplitude movement, are similar to those obtained before fatigue. With fatigue, movements overshot the target. With fatigue, during 80° amplitude movements, IEMG BB values increase of about 10 to 15%. Onsets of BB activity are different for the subjects and occur early in the braking process. The duration of the EMG BB burst increases. The precision of movement does not change. From these results, we concluded that antagonist muscle fatigue leads to amplitude perturbations during aimed fast movements with small amplitude. These data are in favour of the preprogrammed nature of the antagonist burst during smaller movements. After fatigue, the precision was not changed with movements with large amplitude. This implies that adjustment process by the motor system shows that the central control structures could possess the predictive ability to estimate the contributions of visual and kinesthetic influences and how they change during the movement with large amplitude.

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FATIGUE-ASSOCIATED EMG-BEHAVIOUR: DIFFERENCES BETWEEN INTRINSIC HAND MUSCLES

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In man, measurements concerning the relationship between EMG and force reactions have often been performed in the ulnar-nerve hand muscles adductor pollicis (AP; thumb adductor) and first dorsal interosseus (FDI; abductor of the index finger). In histochemical studies, these two muscles have been shown to differ in composition: in AP, type I fibres are dominating (about 80%) whereas FDI has a more evenly balanced composition (about 57% type I; Johnson et al., J.Neurol.Sci., 18 (1973) 111-129). We have compared the fatigue-related EMG behaviour of AP and FDI during an electrically stimulated fatigue test (normal subjects aged 19-31 year; 8 male and 8 female). In this test, the muscles were activated by transcutaneous electrical stimulation of the ulnar nerve at supramaximal intensity (pulse duration 0.1 ms). Bursts of 10 pulses at 30-Hz were given once a second during 5 min. EMG (monopolar) and force recording were obtained from both muscles (FDI and AP) simultaneously. During the test there was a marked decline in evoked thumb adduction force (mainly AP; down to about 48%); abduction forces of the index finger, as caused by electrical ulnar-nerve stimulation, are of complex origin and will be discussed elsewhere (Zijdewind and Kernell, in preparation). With regard to the EMG recordings, the present report concerns measurements of compound action potentials (M-waves) with regard to: (i) peak-to-peak amplitude ('M- size') and (ii) area of the negative peak ('M-area'), all analyzed for the first M-wave of each burst. In the Table, ratios (%) are given for measurements obtained at the end of a test vs. those at the onset. When average EMG measurements were considered for all subjects together, no statistically significant differences were found between the two muscles. However, the picture changed drastically as the results were considered separately for male and female subjects: differences were then observed (i) between AP and FDI muscles of the same group of individuals (paired t tests), and (ii) between males and females with regard to reactions of the AP muscle (t tests; ** $P<0.01$, * $P<0.05$, ns= not significant $P>0.05$). When compared at corresponding times, there were no significant differences between males and females with respect to the drop of thumb adduction force (AP force) during the fatigue test.

TABLE. Average fatigue-associated EMG-reactions for AP and FDI muscles in different groups of subjects.

	M-size			*M-area*		
	AP	*AP vs. FDI*	*FDI*	*AP*	*AP vs. FDI*	*FDI*
All subjects	94	ns	90	97	ns	90
Male (M)	107	**	89	110	*	99
M vs. F	**		ns	*		ns
Female (F)	81	*	90	82	ns	81

The results summarized in the Table suggest: (a) that AP and FDI indeed tend to differ in their fatigue-associated EMG reactions; (b) that recordings from the AP muscle tend to display markedly different reactions in males and females. Further analysis of the data showed a significant correlation ($r=0.992$; $n=9$; 4 male and 5 female) between the relative AP M-wave amplitude at the end of the fatigue test and handlength, as measured from the crease at the base of the hand to the tip of the middle finger. As the female subjects generally had smaller hands, the observed differences in M-wave behaviour might (partly) have been caused by handsize-related factors. In a smaller hand, the relative pick-up area of the electrodes is greater than in a bigger hand. The EMG-depression in AP M-wave amplitude of the female subjects was perhaps reflecting an EMG contamination from other, more fatiguable ulnar-innervated hand muscles (e.g. first lumbrical; first palmar interosseus).

DIFFERENCES IN RESPONSE BETWEEN DIFFERENT ELECTROMYOGRAPHIC FATIGUE INDICES - A THEORETICAL STUDY YIELDING A SYNCHRONIZATION INDEX

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The progressive spectral compression of the electromyographic signal during a sustained contraction is well documented. An established method to mirror this compression is to calculate either the median frequency (MF) or the mean power frequency (MPF) from a surface EMG (SEMG) power spectrum. Also the zero crossing rate (ZC) has been used as a fatigue index.

One of the major physiological factors influencing the SEMG spectrum during fatigue is action potential velocity decrease. It has been shown that the alteration of all three indices mentioned above are proportional to the action potential velocity decrease. This can be expressed mathematically as

$$MF/MF_0 = MPF/MPF_0 = ZC/ZC_0 = v/v_0$$

where the index 0 refers to a well rested reference and v is the average action potential velocity.

There is today evidence that the SEMG power spectrum is altered at fatigue not only by action potential velocity decrease but also by firing statistics alterations (presumably synchronization). These alterations are seen in the power spectrum as an increase in the 10-40 Hz band.

The analytic expression for MF and MPF involves spectral moments of order zero and one respectively. The analytic expression for the expected number of zero crossings per time unit, E[ZC], involves a spectral moment of order two. Generally, the relative response to firing statistics alterations decrease with increasing spectral moment order. These circumstances imply that MF is most sensitive to firing statistics alterations while ZC (or E[ZC] calculated from the power spectrum) is least sensitive with MPF in between. This effect is superposed on the action potential velocity response. There are several examples from the literature illustrating these principles.

As a result of these considerations, an index of synchronization I_s is suggested as

$$I_s = \frac{E[ZC]/E[ZC]_0}{MF/MF_0}$$

where 0 refers to a well rested reference. Alternatively, the $E[ZC]/E[ZC]_0$ factor could be substituted by ZC/ZC_0 or direct measurements of v/v_0 depending on available facilities. Increase in the 10-40 Hz band related to firing statistics alterations will imply an index increasing from unity. If firing statistics components are negligible this index is insensitive to action potential velocity alterations.

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THE EMG MEAN POWER FREQUENCY AS AN ESTIMATOR OF LOCALIZED MUSCLE FATIGUE: PROBLEMS AND CONTROVERSIES

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Ever since Kogi and Hakamada in 1962 described the shift of the electromyographic power spectrum towards lower frequencies in sustained static muscle contractions, spectral parameters have been used as estimators of localized muscle fatigue. However, many basic questions about these parameters are still to be answered. This study is concerned with different validity and reliability aspects of the electromyogram mean power frequency (MPF) as an estimator of localized muscle fatigue in the trapezius muscle.

Systematic variation: We found a systematic and statistically significant variation of normalized MPF in the surface electromyogram caused by factors other than muscle fatigue: shoulder joint angle, hand load and shoulder joint torque. The largest variation, ±8%, was related to joint angle. Thus, there must be a systematic change of more than this value for the MPF to be indicative of localized muscle fatigue. The most probable explanation of this variation was found to be a geometric displacement of the electrodes relative to the underlying muscle tissue.

Random variation: We also found a considerable random variation of single MPF estimates: ±0-26% (or ±11-20% if only within subjects variation was considered). If such single estimates are used, for example, as reference values, steps must be taken to reduce this random variation. At very low signal levels we found severe non-linearities due to noise. This indicates, that any system for electromyographic spectral analysis must have a noise-handling routine, e.g. a lower amplitude limit for signal analysis.

Initial reference value: When different routines to obtain an initial reference value (calibration routines and regression routines) were compared, we found a large between subjects variation, a moderate between routines variation and a variation within subjects with a residual standard deviation of 2-5 Hz (4-10 Hz 95%-interval). We recommend a standard position to obtain a reference value: The straight arm elevated to 90 degrees of abduction in the scapular plane; 2 kg hand load; 10 seconds EMG recording. To reduce the random variation we recommend the calculation of the mean of repeated measurements or the application of a regression routine with calculation of the y- (MPF-) intercept at time zero.

Subjective and objective muscle fatigue: The correlation between subjectively experienced and objectively recorded EMG signs of localized muscle fatigue in the trapezius muscle was examined to test the hypothesis that MPF and a psychophysical fatigue score (CR score) provide essentially the same information. At high load level there was a significant correlation (r= -0.46) between MPF and the CR score, but at low load level there was no correlation. There was a rise of the CR score with increasing load dose, more pronounced at the high load level.

High and low load fatigue: At high load we found a decrease of MPF with increasing load dose, but at low load the MPF did not change despite significant subjective fatigue. It was concluded, that MPF did not work as a valid estimator of localized muscle fatigue at low load levels, common in working life. Cautiousness is recommended if MPF is to be used in static low load situations.

EMG signs of fatigue in work-related myalgia: Patients with unilateral trapezius myalgia were exposed to static muscle load at different load levels. On both sides there was a rise of RMS and a decrease of MPF with increasing time and load, i.e. classical EMG signs of muscle fatigue. However, the changes were less pronounced on the affected side, probably due to pain inhibition, impaired circulation and biochemical changes. At low load level there was no change of MPF despite subjective fatigue and pain.

To sum up, we have found many basic facts, which have to be considered if electromyographic parameters are to be used as estimators of localized muscle fatigue.

FATIGUE OF INTERMITTENTLY STIMULATED HUMAN QUADRICEPS DURING CYCLICAL LOWER LEG MOVEMENTS

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A major issue in the control of functional electrical stimulation (FES) of paralysed muscles is the decay of muscle force as a result of fatigue under sustained (continuous and intermittent) stimulation. Quadriceps fatigue under isometric condition can be described by an exponential decay. In this study the torque of intermittently stimulated paralysed human knee extensor muscles during (isokinetic) cyclical lower leg movements has been investigated.

A protocol was designed to compare overall loss of tetanic torque at the knee joint during sustained intermittent stimulation at different isokinetic velocities. The angle and velocity ranges were limited by the anatomical restrictions of the lower leg, the restrictions imposed by the experimental set-up (a KINCOM 125ES (Kinetic Communicator Exercise System) dynamometer bench) and the desire to maintain a constant cycle time, comparable to a walking cycle. The influence of duty cycle and stimulation frequency, at isokinetic joint movement, was also investigated. Identification trials, determining the torque-angle (isometric) and torque-angular velocity (isokinetic, measured at 40 deg. of knee flexion) relations, were performed. Pulsewidth and amplitude were set to obtain maximal recruitment. The interpulse intervals (IPI) used were 20 and 40 msec, both ensuring a fused contraction. Force, angular position, and velocity were sampled at 100 Hz. These signals and stimulus data were stored on disk for off-line analysis.

The subjects who participated in this study were complete T5-T6 level spinal cord injured patients. All had normal excitable quadriceps muscle and had been enrolled in the FES training program of the rehabilitation center *'t Roessingh* (Enschede, The Netherlands).

From the resulting torque, obtained by subtracting the averaged passive torque from the measured torque at the knee, the maximum and torque-time integral (TTI) per swing were calculated. The TTI was obtained by summating over a constant time period of the swing where the contraction takes place, including activation and relaxation phases, with constant velocity. The typical exponential decay of isometric quadriceps torque, reaching asymptotic values, resembles the overall loss of tetanic torque and TTI during sustained intermittent stimulation at isokinetic condition as found in this study. Additionally, the results indicate a significant dependence of the rate of decay on the contraction velocity, which has not been reported before. Higher velocities result in a faster decay of maximal torque and TTI.

Electrically stimulated muscle is a nonlinear dynamic system, exhibiting nonlinear dependence on position and velocity, which make it extremely difficult to control. Our identification results resemble the output of muscle model structures reported in animal experiments. The typical Gaussian-type dependence of the generated torque on the angle was also found. Hill's equation, favoured by many researchers for curve-fitting the torque-velocity relation for concentric contractions, is also representative for our results.

The dependence of the fatigue curve of transcutaneously stimulated human quadriceps on the isokinetic knee joint velocity and the applied stimulation parameters (duty cycle, IPI) is an important factor in the design of optimal control systems for FES which pursue minimization of muscle fatigue. The results may contribute to the derivation of an optimization criterion, describing muscle fatigue as a function of both joint movement and stimulation parameters.

EFFECTS OF ELECTRICAL STIMULATION FOR MUSCLE STRENGTHENING ON ELECTRICAL MANIFESTATIONS OF MUSCLE FATIGUE

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The aim of this work is to present a case study about electrical stimulation applied for muscle strengthening on a paraplegic patient with a complete T7 lesion. In particular, we emphasize the importance of surface myoelectric signal analysis for assessing the effects of muscle training on the electrical manifestation of muscle fatigue.

Before starting the muscle strengthening protocol, the patient showed poor excitability of knee extensors and very low force production. Muscle fatigue was studied during electrically elicited contractions of thigh muscles. With the patient laying supine on an examination table, Vastus Lateralis was stimulated maximally by means of surface electrodes. Stimulation frequency was set to 35 Hz and the contraction lasted 30 s. Electrically elicited myoelectric signals were detected by means of a surface electrode placed on the stimulated muscle. Following a technique previously developed, we studied muscle fatigue by computing and plotting the time course of the EMG power spectrum mean frequency, EMG average rectified value, and muscle fiber conduction velocity. The results obtained showed a low initial value of conduction velocity (3.05 m/s) associated with a relatively low value of power spectrum mean frequency (64.2 Hz). By observing the time course of EMG parameters, it was also evident that the muscle was only able to sustain the contraction for approximately 21s. After a resting period of 15 minutes, stimulation was administered to the quadriceps by means of larger electrodes, with the leg swinging at the end of the examination table. The torque generated could barely produce a knee rotation of approximately 4-5 degrees.

After this evaluation (May, 1991) the patient underwent regular training sessions repeated everyday. We started with mild isometric contractions of the knee extensors, and we increased progressively the stimulation intensity. Two weeks later, stimulation was administered to the patient with the leg swinging at the end of the examination table, thus allowing for a free rotation of the knee joint. Two months later the patient was able to fully extend his legs cyclically, during training sessions lasting 20-30 minutes. In order to force the muscles to produce a higher torque, increasing weights were added to the ankles. At the beginning of December 1991 the patient was able to fully extend his legs with 2 kg added to each ankle, thus showing an unquestionable increment of the force produced by the quadriceps. Evaluation of muscle fatigue was repeated with the same stimulation parameters previously described, and the results obtained showed that: a) the initial value of muscle fiber conduction velocity (5.30 m/s) was considerably higher (+ 74%) than that obtained before training, b) mean frequency also increased remarkably (+66%), c) the time course of the observed parameters was much more similar to that of a normal subject than the previous one, and d) the muscle was able to sustain the contraction for 30 s. Furthermore, it must be noted that the thigh circumferences increased of approximately 25 mm after training.

In conclusion, the results obtained during this study show clearly that the evaluation of muscle fatigue is well suited for documenting changes that take place in muscles during strengthening programs, allowing the physician to gain a deeper understanding of the effects of electrical stimulation on muscles.

THE INFLUENCE OF MUSCLE LENGTH ON MUSCLE FIBRE CONDUCTION VELOCITY (MFCV) AND MUSCLE FATIGUE

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Introduction. Theoretically, it has been found that the muscle fibre conduction velocity increase when the diameter increase. During dynamic muscle load the muscle length change continuously, but the importance of length changes for the development of fatigue is not known. The electromyogram (EMG) can be used to clarify some of these aspects. The present study investigated the influence of muscle (vastus lateralis) length on MFCV and on muscle fatigue.

Methods. The EMG responses were evoked by electrical stimulation of the motor point and recorded by a tripolar surface electrode array aligned along the muscle fibre direction. The MFCV (determined by cross-correlation) was measured for 5, 45, 90 and 120 degrees knee flexion at three different background extension torques (0, 25, 50% maximum voluntary contraction). In the second experiment the EMG activity to a static fatiguing contraction (80% MVC) was measured at 45 and 90 degrees knee flexion and the MFCV calculated.

Results. The MFCV declined for increasing muscle length, and increased for increasing background torque from 0 to 90 degrees knee flexion. The MFCV can change approximately 2 m/s dependent on the background force and muscle length. During fatigue the MFCV decline rate (0.047 m/s/s) was significantly higher for the 90 degrees knee angle compared to 45 degrees (0.024 m/s/s/).

Conclusion: The muscle length is an important parameter for the propagation velocity of action potentials and for the development of static muscle fatigue.

PATTERNS OF SHOULDER MUSCLE FATIGUE IN A REPETITIVE ARM WORK WITH AND WITHOUT ORGANIZED PAUSE ACTIVITIES

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In order to break monotony in repetitive work tasks and to activate muscles exposed to long term static load, pauses or pause gymnastics have been introduced into the work. One crucial issue in the introduction of pauses is, however, that the productivity has to be maintained. The aim of this study was to investigate whether muscle fatigue measured with electromyography develops in shoulder and neck muscles during a continuous work and during work with organized pauses.

Materials and Methods

Twelve healthy female subjects (mean age 25.5 years) were performing repetitive arm work for two hours in a controlled laboratory setting. The work task consisted of pinching a small cylinder (weight 15 g) with the right hand and to release it through a hole in the table. In a randomized order one hour was continuous and the second with organized pauses. The pauses were introduced every sixth minute lasting one minute each, i.e. ten minutes during one hour. The pause activity consisted of lifting a box (2 kg) 5-6 times and of rest between the lifts. The work pace was set to 2466 cycles per hour which means 41 cycles per minute for continuous work and 49 for work with pause activities. Thus the productivity was kept the same during the two work types.

Surface electromyography was recorded on the right side from the lateral and cervical portions of the descending trapezius muscle and from the infraspinatus muscle. The mean power frequency, MPF, and root mean square amplitudes, RMS, were computed in parallel processes for analysis of fatigue for the entire work period and separately in five minute periods for analysis of fatigue patterns. Increase in RMS with a simultaneous decrease in MPF were considered as electrophysiological signs of muscle fatigue.

Results

Muscle fatigue developed over the entire work in the lateral portion of the descending trapezius muscle in eight subjects during continuous as well as during work with organized pauses. For the cervical portion of the descending trapezius muscle fatigue developed during continuous work in eight subjects and with pauses in three. In the infraspinatus muscle, muscle fatigue was found in three subjects during continuous work and in none during work with organized pauses. Fatigue patterns in five minute intervals were found both during continuous work and during work with organized pauses. The number of fatigue patterns differed between muscles and between subjects and there were fewer patterns during work with pauses. The ratings of perceived exertion increased rapidly during the first fifteen minutes during both work types. Greater variations in perceived exertion could be seen during work with organized pause activities than during continuous work.

Conclusions

Pause activities and work pace are important factors for the development of electromyographic signs of shoulder muscle fatigue and perceived exertion in a seated work with repetitive work tasks.

MUSCLE FIBRE AND MOTOR UNIT ELECTROPHYSIOLOGY AND THE SURFACE EMG DURING FATIGUE: A SIMULATION STUDY

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It is known that intracellular action potentials (IAPs) of muscle fibres are influenced by local muscular fatigue (Lännergren and Westerblad, 1987). It is further known that surface EMG (SEMG) characteristics are largely dependent on these local electrophysiological changes (Lateva, 1988).

The study presented here aimed at a model-based quantification of the influence of changes at single fibre level and at motor unit level on major SEMG characteristics (root mean squared amplitude (**RMS**) and median frequency (**Fmed**)). In the model the IAP wave shapes are characterized by three parameters, namely their mean IAP amplitude (**A**), their mean duration (**T**) and the mean fibre conduction velocity (MFCV here: **U**). A motor unit action potential is modelled as the sum of the extracellular single fibre action potentials dispersed in arrival time at the electrode.

The results can be expressed in the following statements on the SEMG amplitude and frequency characteristics:

(1) As known, volume conductor differences (e.g. the distance between a motor unit and the electrode and the electrical conductivities) dominate the absolute value of **Fmed** and **RMS**. Volume conduction does not interfere, however, with relative changes in Fmed and RMS when they are induced by IAP changes.

(2) **RMS** is almost proportionally dependent on the spatial extension of the IAP (**L=T.U**) and on **A**. It is inversely proportional to the square root of **U** itself. In combination:

$$\mathbf{RMS = C_1.A.T.U^{1/2}}.$$

(3) **Fmed** is about proportionally dependent on **U**, independent of **A**, and almost independent of **T**:

$$\mathbf{Fmed = C_2.U}.$$

The influence of IAP characteristics on SEMG changes can be described by the above simple relations quite accurately (for RMS ref. Fig. 1). An important assumption has to be mentioned. For the above relations to be valid a motor unit must be homogeneous with respect to the relative fatigue resistance of its constituent fibres. Violations of this homogeneity do influence **Fmed** and especially **RMS** (Fig. 2). Quantitative basic experimental knowledge on homogeneity and on the relation between the conduction velocity U, which can be measured under in situ conditions, and the other IAP parameters (A, T) is lacking. This knowledge would allow the practical implementation of model results as presented here, thereby largely supporting the interpretation of SEMG in fatigue.

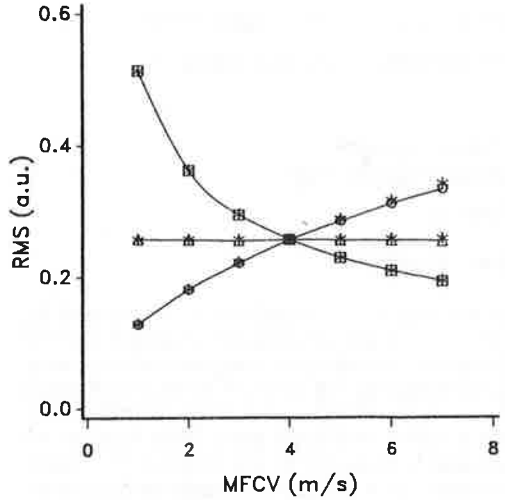

Fig. 1 The calculated relation between **RMS** and **U** (MFCV). The IAP duration **T** is changed in relation to the **U** in different ways (squares: **T=4/U**, triangles: **T=(4/U)**$^{1/2}$, circles: **T=1** ms). Asterisks show the behaviour of the equation under (2). Note the close fit with the model results.

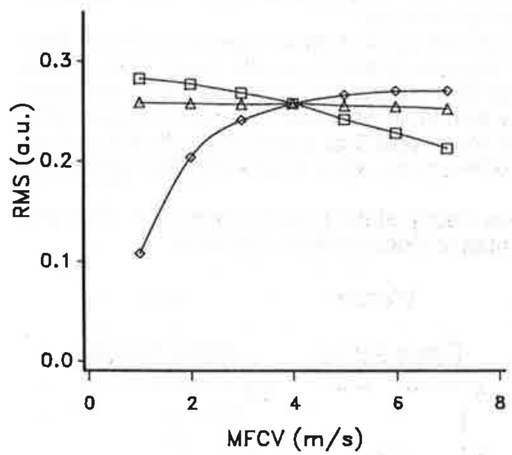

Fig. 2 The calculated relation between **RMS** and **U** (MFCV). Triangles as in Fig. 1. Squares: as triangles, assuming that within a motor unit the thicker fibres are less resistant to fatigue than the thinner fibres. Diamonds: vice versa.

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WOMEN HAVE LOWER TRAPEZIUS MUSCLE CAPACITY THAN MEN

A Study Using Continuous Laser Doppler Flowmetry (LDF) and Surface Electromyography (EMG)

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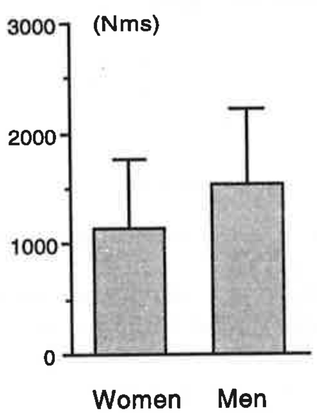
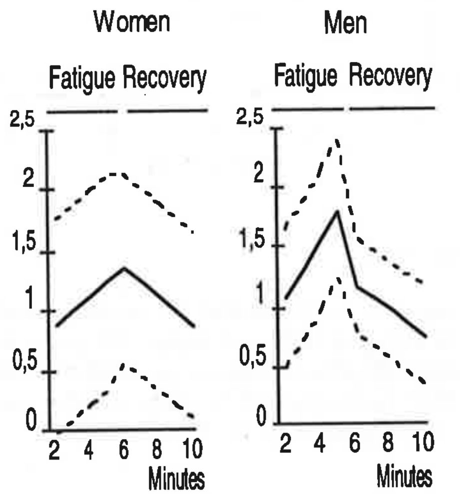
The microcirculation in the upper portion of the trapezius muscle was measured by continuous LDF during two 10-minute long series of alternating one-minute periods of static contraction and rest, followed by a 10-minute recording of fatigue and recovery. Stepwise increased contraction was induced by keeping the straight arms elevated at 30, 60, 90 and 135 degrees, respectively, that was repeated with a 1 kg (women) or 2 kg (men) load carried in each hand. The arms with hand load were then kept at 45 degrees until fatigue. The blood flow was studied in relation to the degree of muscle activity in consecutive series of 16 healthy women aged 43 (20-62) years and 13 men aged 37 (20-57) years. Signal processing was done on-line by using a 386 SX computer. Root mean square converted LDF- and EMG- signals were normalized by using the average value of the serial examinations as a reference value which reduced the variations. Spectral analyses of EMG were performed.
Women had significantly less endurance measured as performed load dose (Nms) than men. At muscle fatigue shown by an EMG frequency shift towards lower frequencies (MPF and MDF), a rise in blood flow occurred first during the rest periods and later also during the high-force contractions. Recovery was accomplished by a marked rise in the LDF/EMG quotient due to increased muscle blood flow that correlated with the degree of fatigue, i.e. the lowering of MPF. Age differences were moderate and similar for women and men.
Women developed more fatigue despite less heavy static muscle work than men and showed a less ability than men to increase muscle blood perfusion at work.

Fatigue Experiment



Endurance as performed load dose in women and men.
Nms=Newton-meter-seconds.



Trapezius muscle blood perfusion during fatigue and recovery in women and men.

PARAVERTEBRAL MUSCLES DURING PROLONGED SITTING POSTURE

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Introduction
Although the precise physiopathology of low back pain is still incompletely understood [1], the relevance of paravertebral muscles distress in this syndrome is well known. This is particularly true in the case of forcedly prolonged bodily postures, making this issue one of particular importance in the ergonomics of work and daily life activities. A major contribution to this problem may come from studies on muscle fatigability [2], and in particular from recent studies that show the distinctive fatigability of paravertebral muscles in back pain patients, as measured by means of EMG techniques during effort tasks [3].

Material and Methods
In agreement with this approach, we performed examinations of the fatigability of some paravertebral muscles during a constant postural task, namely the prolonged sitting posture, similar to the condition of prolonged car driving or sitting at a work station. Fifteen adult voluntary subjects were requested to stay one hour quietly sitting on a high chair without back support, with their legs freely hanging without contact to the floor. Six preamplified surface leads (DEM, Leini (Turin, Italy)) were applied to each subject, the first pair over the paravertebral area at the level of T7-8, about 2 cm from the middle line, a location corresponding to lower fibers of the Trapezius (to evaluate also the possible participation of higher dorsal muscles to the postural effort); the second pair was placed at L1-2 level, about 4 cm from the middle line, in correspondence with the Latissimus Dorsi; the third pair was placed at the sacro-lumbar junction, 2 cm from the middle line, just over the terminal part of the Sacrospinalis muscle. The leads were directly connected to a personal computer via an A/D converter for recording and subsequent elaboration. EMG signals were sampled at a sampling rate of 1kHz for a 1 second period every ten seconds: hence, 360 periods were sampled in one hour. For each period the mean amplitude of the EMG signal and the mean frequencey and energy content of the EMG power spectrum were calculated and plotted versus time.

Results
Some subjects showed a clear cut, progressive downward shift of the mean EMG frequency throughout the examination time, together with a simultaneous increase of the signal amplitude and energy content; these variations, clearly suggestive of muscular fatigue, involved one or more recording sites, generally with asymmetrical distribution. This pattern was consistently observed in the six older subjects with history of low back pain, although the site of fatigued muscles did not necessarily match the distribution of pain (Fig.1) Five young subjects showed nearly invariant frequency patterns (Fig.2), and four, with history of backache, showed only a slight decrease of mean frequency in one or more sites (Fig.3).

Conclusions
These preliminary results should be considered only as scattered observations on the behaviour of some back muscles during a particular task. They need to be validated and characterized on wider populations, with systematic evaluation of the various muscular groups of the trunk. Nevertheless, they seem to confirm, on a purely qualitative basis, the

relationship between back muscles fatigability and back distress, in "daily life" conditions. Further studies are needed to investigate also: I. to what extent back muscles' fatigability is related to some other (for example, discal) abnormalities, or alternatively if it may play a primitive role in the genesis of back pain; II. the meaning and role of asymmetrical muscular behaviours leading to different levels of fatigue during postural tasks.

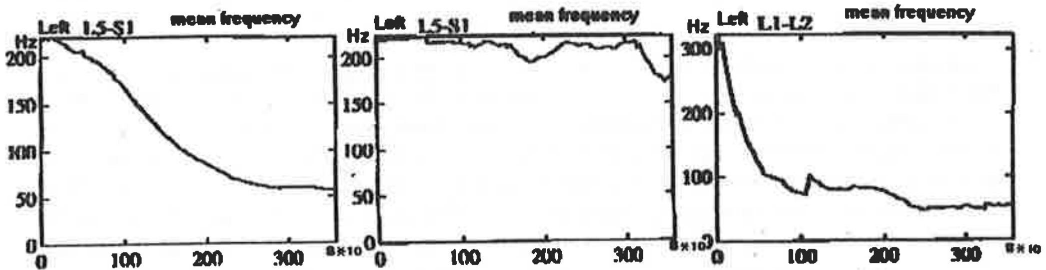

Fig 1. 48 yeaq male   Fig 2. 24 years, male   Fig 3. 38 years, male

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AUTOREGRESSIVE MODELLING, POWER SPECTRAL ESTIMATE OF ELECTROMYOGRAPHIC SIGNALS AND PARAMETER EXTRACTION FOR MUSCULAR FATIGUE ANALYSIS

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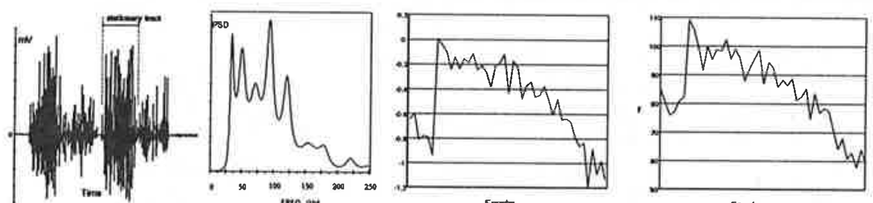
Muscular fatigue appears in skeletal muscles as the inability to maintain a predetermined load, for example a weigth. Particularly interesting is fatigue analysis in respiratory muscles, due to their regular, continuous and essential work, the same as the heart, during the whole life of the subject. Fatigue, as happens for skeletal muscles, affects diaphragm working under a resistive load. In this case the resistive load consists of a partial occlusion of inspiratory ways. A classical and direct approach in studying muscular fatigue is the analysis of the electromyographic (EMG) signals. In this work signal analysis was carried out mainly in the frequency domain due to the experimental results which have related the beginning of muscular fatigue with the shift of the main spectral components towards lower frequencies [3] [4].

METHODOLOGY: Two series of experiences, one for the human biceps, the other for the diaphragm in pig, were carried out, both analysed using the same methodology. During the experiments a resistive load was imposed until fatigue appeared. EMG signals were aquired with pre-amplified electrodes, amplified, recorded, digitalized with a 12 bits A/D converter and analysed with PC using dedicated software. The stationary tracts of EMG signal has been considered as the output of a discrete time stationary random process described by an autoregressive model (AR) given by :

$$y(k) = -\sum_{i=1}^{M} a_i y(k-i) + e(k)$$

where M is the AR model order and the driving process e(k) is a white noise stationary stochastic process with variance σ^2 [2]. A maximum entropy spectral estimate was then carried out using the previously identified AR model (M=20) [1] [5]. Centroid, already used in literature as a fatigue index, and higher order moments were then computed on the spectrum to evaluate the spectral shift.

RESULTS: In all the experiences the behaviour of the first AR model coefficient (a_1), during the epochs of the experiment, was found to be related to the centroid. This result suggests the use of the parameter a_1 as a fatigue index, computable directly from the EMG data without estimating any spectrum. The centroid and a_1 shift has shown that diaphragmatic fatigue in pig occurred only 2 hours after resistive inspiratory load application, while in experiences carried out on human subjects diaphragmatic fatigue appeared after 5'-10' [3].



EMG signal   *M.E. Spectrum*   *Parameter a_1*   *Centroid*

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THE ROLE OF SYNCHRONIZATION IN LOCALIZED MUSCLE FATIGUE

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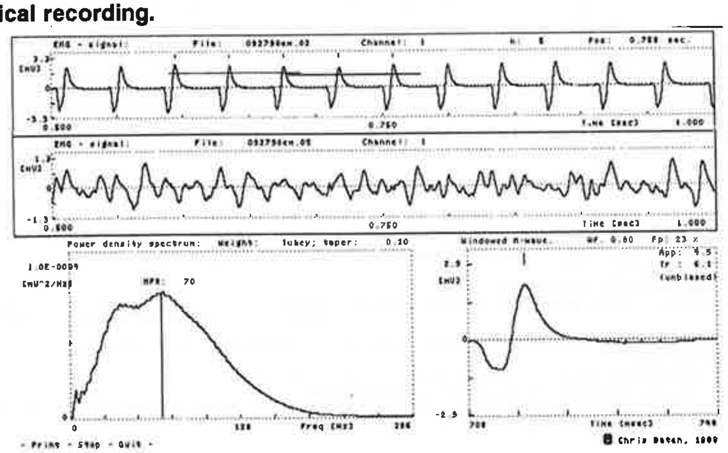
1: Central question.
Does synchronization of motor unit firing patterns play a significant role in the phenomena observed during voluntary sustained contractions which are referred to as localized muscle fatigue [1]?

2: Theoretical background.
In voluntary unfatigueing contractions the median frequency (MF) is reported to be related to [2,3,5]:
1: changes in MFCV,
2: changes in synchronization level,
3: changes in mean firing rate (FR),
4: changes in recruitment level and
5: changes in temporal dispersion among motor unit fibers.
-Theoretical work and a simulation study [4,5] rule out possibility 3 as being significant.
-4 and 5 occur only shortly before total exhaustion [own observations confirmed in these experiments]

3: Experimental design.
To induce localized muscle fatigue 5 subjects in 10 experiments sustained a isometric voluntary contraction at a 80% MVC level. To seperate the contribution of changes in synchronization level from other mechanisms of fatigue the median frequency (MF) values of two types of EMG recording were compared, being:
- MFV: the MF of the EMG signal from the voluntary contraction,
- MFS: the MF of the EMG signal from stimulated contractions which shortly interrupt the sustained voluntary contraction.

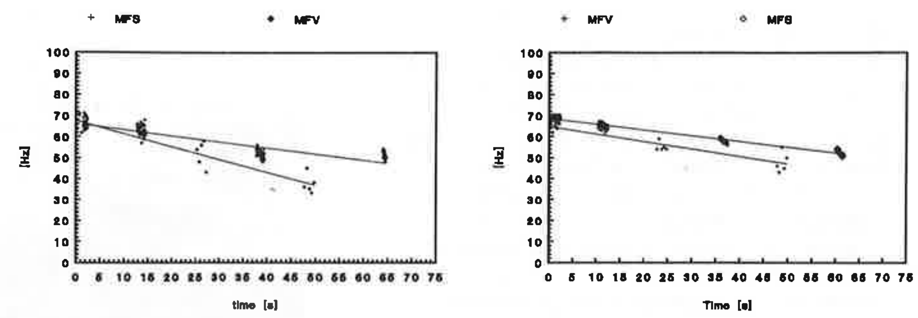
Both MFV and MFS are equally sensitive to changes in MFCV (1) but only MFV is sensitive to changes in synchronization level (2).

4: Typical recording.



Upper trace: EMG stimulation (NB. small stimulation artefacts, stable recruitment
Lower trace: EMG voluntary contraction

5: Typical analysis result.



Left: MFV and MFS trends with influence of synchronization level changes (no initial synchronization)
Right: MFV and MFS trends with no significant influence of synchronization level changes (with initial synchronization)

6: Summarized results.
Experimental assumptions.
- Both peak-peak-amplyitude and shape of the M waves did not change during the sustained contractions until shortly before total exhaustion.

Initial MF values.
- In 5 out of 10 experiments the initial value of MFV was equal to the one of MFS (Left graph).
- In 5 out of 10 experiments the initial value of MFV was lower than the one of MFS (Right graph).

Decrease in MF values.
- In 6 out of 10 experiments the MFV decreased faster than MFS (Left graph).
- In 4 out of 10 experiments no significant difference in MF decrease was found (Right graph).

7: Conclusions.
About experimental assumptions
- In the stimulated contractions changes in recruitment level and in temporal dispersion took place only shortly before total exhaustion.

About initially present synchronization.
- In 5 out of 10 experiments no synchronization was present in the voluntary signal at the start of the experiment (Left graph).
- In 5 out of 10 experiments synchronization was already present in the voluntary signal at the start of the experiment (Right graph).

About contribution of synchronization level change to spectral shift.
- In 6 out of 10 experiments changes in synchronization level contributed significantly to the spectral shift (Left graph).
- In 4 out of 10 experiments no significant contribution of changes in synchronization level was found (Right).

Methods details

I.: Experimental set-up.

-Experiments on human Tibialis Anterior muscle
-Leg fixation at 100 degrees dorsal flexion
-Force level controlled by visual feedback
-Bipolar EMG electrode, d = 14mm, positioned between most distal motor point and tendon
-Stimulation with round electrodes (ø 30 mm)
kathode: on most proximal motor point
anode: on bursa infrapatellaris profunda
-Stimulation rate: 20 Hz.
-Inflatable cuff just below knee to maintain ischemia during interruptions (MFS estimation)
-Computer controlled recording, stimulation and experimental timing

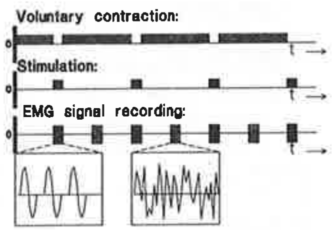
II.: Recruited motor unit pool

- Voluntary contraction: at 80% MVC all motor units are assumed to be recruited
- Stimulated contraction: using the PPA trajectory for growing stimulation amplitude the level of (supra)maximal recruitment was determined

III.: Natural development of fatigue.

Preparatory experiments showed that artificial sustain of the ischemic condition during the interrupotions of the voluntary contraction (for MFS estimation) was neccsary and sufficient to prevent disturbance of the natural development of fatigue.

IV.: Experimental timing.



- Voluntary sustained contraction of 75 sec or upto exhaustion
- MFS estimation: 2.5 sec. interruptions and recordings at t = 10 s, 35 s, 60 s, etc.
- MFV estimation: 2.5 sec. recordings at t = 22.5 s, 47.5 s, 72.5 s etc.

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SUPERNORMAL CONDUCTION VELOCITY AFTER SUSTAINED MAXIMAL ISOMETRIC CONTRACTION IN HUMAN MUSCLE

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<u>Introduction</u> In studies of local muscle fatigue and recovery under aerobic conditions a normalization of EMG parameters appears within 5 to 7 minutes. However, some authors mention an overshoot of the median frequency (Fmed) in the recovery phase some minutes after fatiguing contraction. This phenomenon prompted us to a more detailed study of the recovery phase after maximal contraction by means of surface and invasive EMG.
<u>Methods</u> Surface EMG measurements were performed on the biceps brachii muscle of 4 volunteers to record the changes in muscle fibre conduction velocity (MFCV), Fmed, integrated EMG (IEMG) and force during 1 min maximal voluntary contraction (MVC) and a recovery period of 60 min during short lasting isometric contractions at 20% and 50 % MVC. In an additional experiment we studied the changes in MFCV before and after 1 min MVC and during 15 min recovery, by directly stimulating and registrating the muscle fibres with needle electrodes in both biceps brachii and brachioradialis muscle.
<u>Results</u> Surface EMG: The remaining force after 60 sec MVC was 46% of the value at the start of the experiment (mean slope -2.1 N.s^{-1}). The MFCV and Fmed also showed a rapid decline. MFCV slope: -0.035 m.s^{-2} and Fmed slope: -1.01 Hz.s^{-1}. During recovery the main finding was a persistent supernormal MFCV after a normalization phase of 5 minutes, reaching a steady state after 10-12 min. Mean MFCV increase at 20% MVC after 60 min recovery 0.58 m.s^{-1} (15%), at 50% MVC 0.32 m.s^{-1} (8%). The changes in Fmed paralleled the MFCV increase. Post fatigue IEMG values were increased at all contraction levels, despite a normalization of force generating capacity. This resulted in a long-standing decrease in 'neuro muscular efficiency' (IEMG / force).
Invasive EMG: During fatigue the mean MFCV decreased 0.59 m.s^{-1} (17%), during recovery the mean overshoot was 0.50 m.s^{-1} (15%). During the recovery phase a rapid return to prefatigue levels was seen after 4 to 5 minutes, followed by an overshoot which stabilized after 10 - 12 minutes. The MFCV values were significantly higher compared to the prefatigue situation.
<u>Conclusions</u> During recovery we found after 1 min MVC a 'supernormal' MFCV, reaching a steady state after 10-12 min. Post fatigue IEMG values were increased at all contraction levels which resulted in a decrease in 'neuro muscular efficiency' (IEMG / force). We suggest that the higher EMG amplitudes are mainly based on the MFCV increase. The relative MFCV increase between fastest and slowest fibres measured was equivalent indicating an equal effect on type I and II fibre types. A possible explanation is a muscle fibre swelling on the basis of an increase in intracellular water content, in combination with changes in membrane properties.

RELATIONSHIP BETWEEN ENDURANCE TIME AND CHANGES IN THE EMG POWER SPECTRUM OF THE ERECTOR SPINAE DURING ISOMETRIC CONTRACTION TO FATIGUE

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Excessive fatiguability of the back extensor muscles is a frequent dysfunction associated with chronic low back pain (1). Consequently, techniques for objectively monitoring improvements in the endurance of the erector spinae, in response to treatment or rehabilitation, are continually being sought. Unfortunately, many of the clinical procedures currently in use are highly dependent upon subject motivation for their validity. We have attempted to evaluate one such test with respect to the simultaneous changes in the EMG power spectrum of the erector spinae muscles.

49 young healthy human volunteers, aged 19-33, agreed to participate in the study. Whilst lying prone on a couch, with the lower body supported and stabilised from the hips down, subjects were required to maintain the unsupported upper body in a horizontal position until (due to fatigue) they were no longer able to overcome the force of gravity. Throughout the duration of the test, surface electromyograms were recorded from the erector spinae (ES) at the levels of the 10th thoracic (T10) and 3rd lumbar (L3) vertebrae, from which the median frequency (F_{med}) was determined. Regression analysis was used to examine the relationship between time to fatigue and the rate of decline of the median frequency.

The test induced a significant ($p=0.0001$) reduction in the median frequency at both ES recording sites (thoracic, $-30 \pm 12\%$ and lumbar, $-35 \pm 13\ \%$), but in a number of subjects the change was far more prominent at one level than the other. A significant ($p<0.001$) relationship was observed between the time to fatigue and the relative rate of decline of F_{med} at (i) the T10 level ($r=0.44$), (ii) the L3 level ($r=0.56$) and (iii) both regions (mean value) ($r=0.63$) of the erector spinae. However, the best predictor of time to fatigue, was given by the greatest % decline in F_{med} obtained at *either* of the two levels ($r=0.71$; Fig. 1). In the latter case, the decline in F_{med} accounted for 51% of the variance in the test time to fatigue.

The data appear to suggest that fatigue in a specific region of the ES (as determined by F_{med} changes) may result in the individual being unable to generate the required force for continuation of the test, such that the limiting factor for performance may be the endurance of the most fatiguable region of the ES. This, coupled with the variability in the median frequency changes at different regions of the ES, demonstrates the importance of recording at more than one level. The rate of decline of F_{med} of the ES appears to be a good predictor of endurance time, but the latter is undoubtedly also influenced by the motivation of the subject. Patients suffering from back pain may be reluctant to continue the test to their limit of fatigue, and it is suggested that a better indicator of back muscle endurance may be obtained by monitoring the rate of change of F_{med}, from a number of regions of the ES, over a fixed sub-maximal time period. In this manner, valuable information about the fatiguability of the individual can be obtained, which is not dependent on him/her reaching any endurance limit.

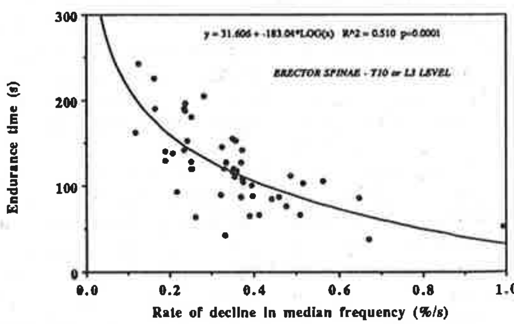

Fig.1 Relationship between endurance time and rate of decline of the median frequency of the erector spinae during isometric contraction to fatigue.

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A KEY PROBLEM IN BIOMECHANICS OF SPORTS: THE DETERMINATION OF INTERNAL FORCES

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1. INTRODUCTION

Internal forces like muscle and joint forces play a major role in human movement. They are closely related with performance and mechanical loads on the movement apparatus. The maximization of performance as well as the danger of excessive loads during physical activities emphasize the need for a detailed quantitative picture of muscular performance under different and sometimes extreme dynamic conditions. This is of value in particular for top level sports. Moreover the process of functional adaptation of the biological structures is connected with muscular activity and therefore an important basis of the training in sports.

2. METHODOLOGICAL APPROACHES AND RESULTS

Because of intolerable feedback muscle and joint forces acting within the human body have not been measured directly. Exceptions are endoprostheses equipped with force and pressure transducers resp. The determination of these forces has to be carried out indirectly, by means of mechanical models of the body and appropriate measurements of external mechanical quantities. Biological systems are constructed redundantly, that means there are much more muscles than needed to perform a certain movement. Consequently the equations of motion describing the relations between forces and movement, represent a mathematically indeterminate problem. Possible solutions can be achieved either by drastic simplifications (e.g. reduction of the number of unknown forces) or by the stipulation of principles according to which the movement is controlled (optimization criteria, various constraints). Whichever appraoch is used, the result can only give an estimate of the desired quantities.

2.1 NET MOMENTS

A simple model is commonly used to determine net joint moments and global internal forces in the lower extremity. It is based on the assumption that external moments are balanced by internal moments. This is the first step reflecting the "bio" parameters: it takes into account the lever arms of the musculature, which are important structural characteristics of the body. The following examples may demonstrate how useful joint moments can be in the judgement of movement techniques.

In training very often certain exercisis are selected to simulate the technical and physical requirements of competitive movements. In fig. 1 an example from *weightlifting* is depicted: the comparison between snatch (competitive movement) and snatch pull (training exercise). Looking at the kinematic patterns of selected body points, no substantial differences can be detected. The external dynamics, represented by the vertical component of the ground reaction force, are also nearly identical for both movements. Only the resultant net moment shows the considerable differences in muscular loading of the knee extensors. In this as in many other cases neither the kinematics nor the force platform data alone give enough information about the structure of the movement.

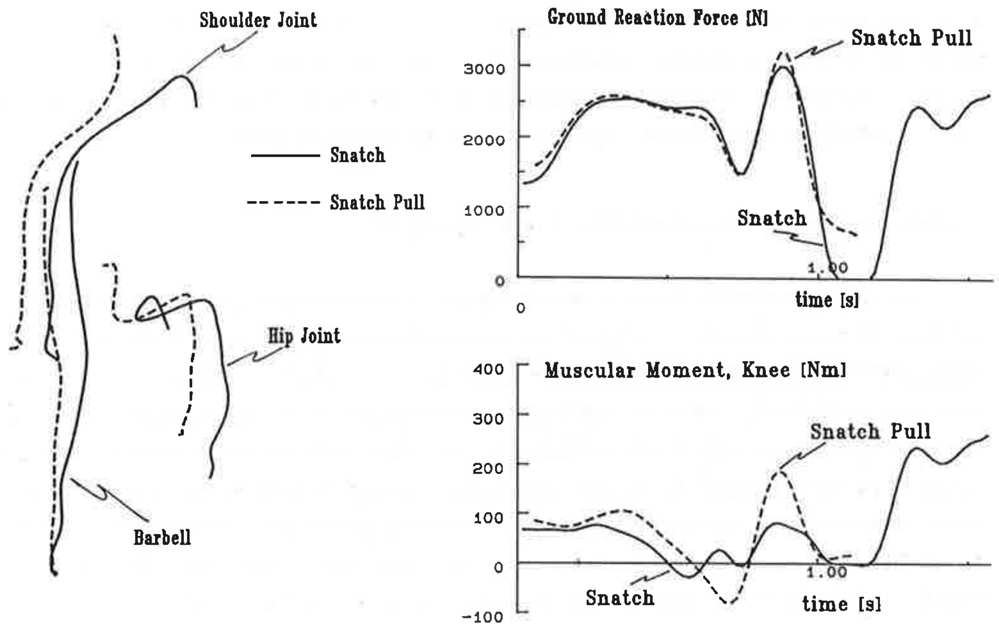

Fig. 1: Kinematic (left side) and kinetic parameters in weightlifting. Comparison between snatch and snatch pull.

In *running* the movement of the foot related to the shank during the contact phase is considered to be an indicator of pathological loading of the achilles tendon and the medial/lateral ligaments of the ankle joint. In particular excessive pronation in the beginning of the stance phase is regarded to be responsible for several trouble. A simple consideration may be that pathological

loads can only be present when the structures are under a certain tension. In this case predominantly tensions in the achilles tendon and the medial ligaments/muscle components come into question. The figures 2a/b show the results of an analysis of runners for one single subject.

From the net moment in the sagittal plane (Fig. 2a) it becomes clear, that it concerns a rearfoot runner, since in the very beginning the foot flexor muscles are active, thus controlling the extension of the foot. What is more important: In the phase of maximum pronation (30 - 40 ms after heel strike) the moment in the sagittal plane indicating a loading of the achilles tendon is close to zero when the pronation angle is maximum. The same is true for the moments in the frontal plane (Fig 2b), possibly indicating a loading of the medial ligaments and muscle components. This is not a *general* pattern, it is individually different and most probably depends on the flexibility of the foot joints. However, the result underlines that it is not advisible to judge a movement alone on the basis of kinematic informations.

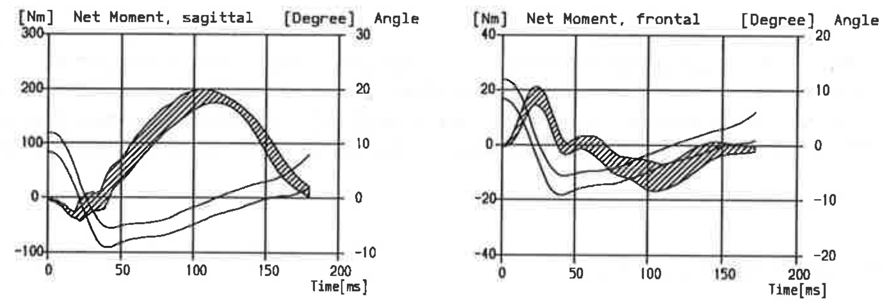

Fig. 2a/b: Net moments and pronation/supination angle of the foor during the stance phase in running. Shaded areas: moments with respect to the ankle joint, empty areas: angle. Positive angles indicate supination, negative angles pronation. 2a: sagittal plane, moments are pos. when loading foot extensors, neg. when loading foot flexors. 2b: frontal plane, moments are pos. when loading the medial, neg. when loading the lateral structures.

As a result of dynamic calculations the net moments acting in the joints are representing the load on global structures of the movement apparatus. Not considering the antagonistic muscle actions, joint and muscle forces can be calculated. Fig. 3 gives some examples of the the net moments acting in the knee joint in the sagittal and frontal plane resp. These figures give a rough estimate of the minimum moments required to perform the motor task. These results are very useful in comparative investigations, where the influence of movement technique, physical properties of the body and of external factors can be studied.

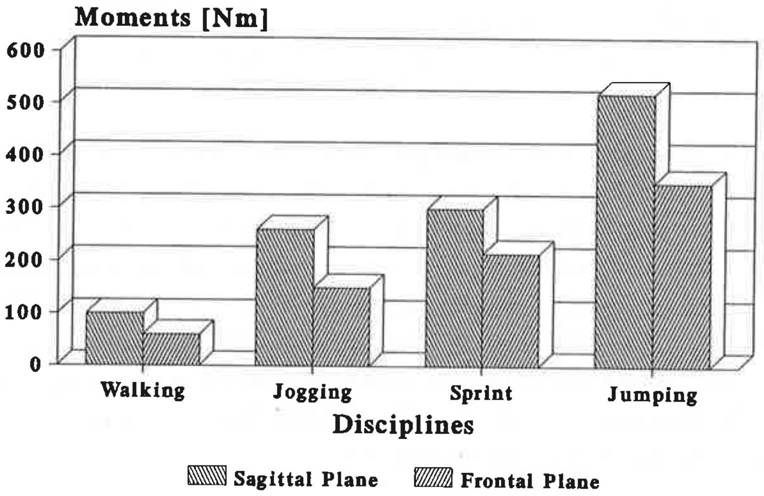

Fig. 3: Maximum values of net moments in the knee joint during the stance phase of various sports movements. The moments in the sagittal plane are related to the knee extensors. The moments in the frontal plane are compensated predominantly by the lateral and medial load bearing structures of the knee joint. These values largely depend on varus/valgus positions of the knee joint.

2.2 OPTIMIZATION CALCULUS

Despite the usefulnes of this "net moment model", due to a number of reasons it can yield more or less erroneous results: uncontrolled antagonistic muscular activity, unknown force distribution between two-joint muscles or muscles acting in the same function, loading of ligaments and - very important - a rather rough estimation of the macroscopic anatomical structures of the body. Appropriate parameter variations using the same model tell us something about the wide range of numerical results. In order to overcome the indeterminacy of the biomechanical system, mathematical optimization procedures have been used. These methods are useful tools for the analysis of complex systems, which are functioning according to certain principles. The crucial question hereby is which criteria should be selected as objective functions.

Following the model of BRAND et al. (1982) a complex approach, including 47 muscles of the lower extremity and further structual data, was worked out for analysing the effects of different objective functions, experimental setups and joint constructions.

Mainly due to the availability of optimization routines linear objective functions have been used at first. Different investigators used linear approaches for determing muscle forces in the upper and the lower extremities. But the theory of linear optimization proves that its characteristic is quasi in contradiction to the redundancy of biological systems. Linear optimization tries to share the load among the fewest possible number of structures, whereas in man a lot of synergistic muscles are activated during locomotion, which can be shown easyly by EMG recordings (s. Fig. 4 left side).

Further developments of nonlinear optimization algorithms yielded to their application also in biomechanics. The characteristic of quadratic or higher polynomial objective functions is to share the "costs" among synergistic elements - a major progress in the development of more sophisticated models of the musculo-skeletal system. The first, native objective function is the sum of the squares of the muscle forces. But then the model will handle the muscles independent from their strength and this will result in very high muscle stresses in some cases. Therefore it is necessary to incorporate the cross sections of the muscles. With such an approach the muscle force predictions are in much better consistence with the EMG patterns (s. Fig. 4 right side).

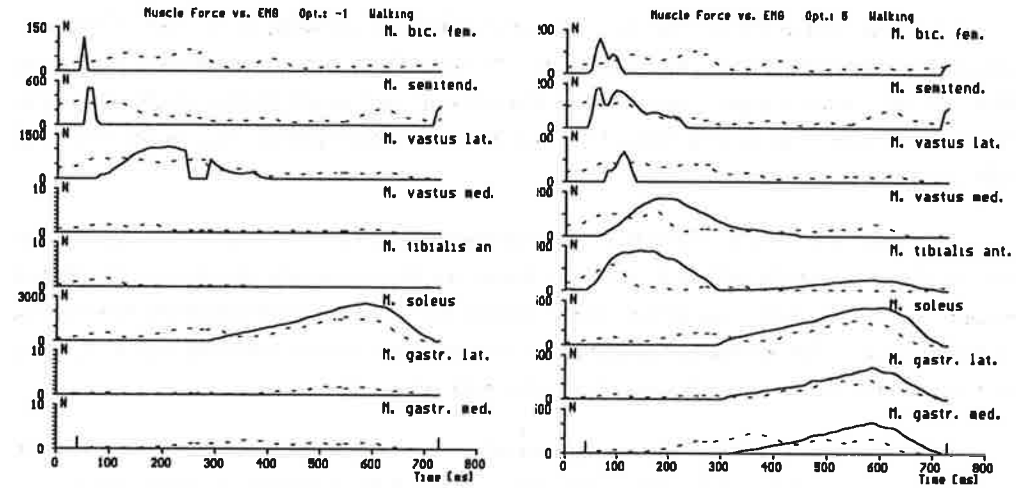

Fig. 4: Comparision of the muscle forces (solid) resulting from a linear optimization approach (min Σ muscle forces) (left side) and a quadratic optimization approach (min Σ (muscle stresses)²) (right side) with high-low pass filtered EMG (dashed). The EMG data were individually scaled and can not be compared with others (from GLITSCH 1992).

Besides the objective function, the results of an optimization approach are mainly affected by its dimension and the representation of the joints. So, the hip joint forces increase by almost 100% using a threedimensional approach instead of a twodimensional (s. Fig. 5).

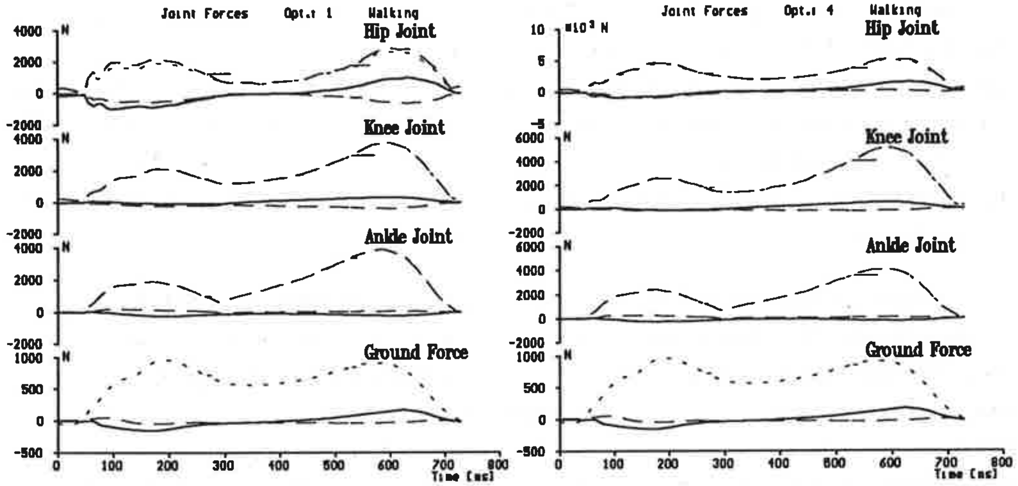

Fig. 5: Joint forces of one trial using a twodimensional (left) and threedimensional (right), quadratic optimization approach (min Σ (muscle forces)²) approach. (——) horizontal (· · ·) vertical, (- -) lateral, (—— ——) resultant component (from GLITSCH 1992).

A further step in finding objective functions of biological relevance is the study of Pauwels' results. He studied the construction principles of the human body from an engineering standpoint. His work represents a major contribution to biomechanics and would deserve much more attention than it received up to now. One of his well known key experiments concerns the functional role of the ilio-tibial tractus.

The main results of his theoretical and experimental studies concerning the mechanical properties of bones and the effects of muscular forces can be summarized as follows: The human skeleton is an ideal light construction. Shape, density and structure of the bones are reducing the bending stresses. The geometrical configuration of muscles and their coordinated activities are reducing bending stresses of the bones and reducing peak stresses in the joints.

It seems obvious that these constructional principles yield a sound basis to deduce biologically meaningful optimization criteria: the minimization of total stresses in active and passive structures of the locomotor system in a time average. The relation between structure and function is so close that just from the structure the functional requirements can be deduced. Supplementary to the recent research work of Glitsch, Pauwels criterion is used by Siebertz as the objective function to compute internal forces using optimization techniques.

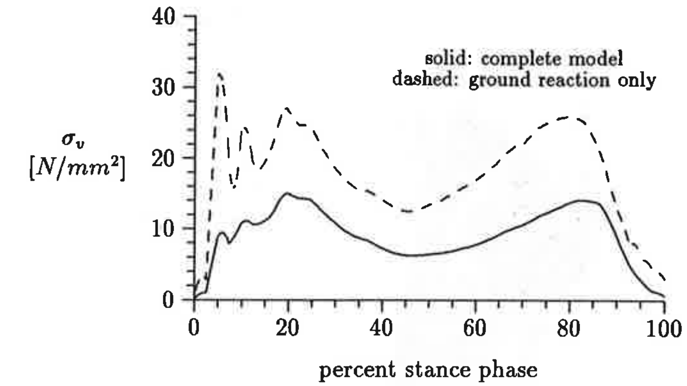

Fig. 6: The average equivalent stress in the femur during normal walking. Whereas the solid line represents the result of the complete model, the dashed line only displayes the influence of the ground reaction. If there were no muscles, but some kind of little engines in the joints to keep the moments balanced, the dashed line would represent the resulting average equivalent stress.

The fully 3-D stress analysis in the bone is based on Bernoulli's beam theory. Bending stress, compressive stress, shear stress due to shear force and shear stress due to torsion are simultaniously calculated for 540 points which form a dense net along the bone's surface. The multi-axis stresses are converted to single-axis equivalent stresses according to the theory of maximum strain energy. Despite the unbeatable accuracy, the Finite Element Method cannot be used in this context because it is to slow on the available computers. Since we need all partial derivatives at any time step, one gait analysis with 100 time steps and 45 muscles would at least require 4500 FEM runs. Using the beam theory a complete calculation of internal loads can be performed in less than five minutes on our Silicon Graphics workstation, which corresponds to approx. 600 mio. floating point operations.

The target function used here is the strain energy in femur and tibia. This implies that the problem can be expressed in terms of quadratic optimization, which is numerically stable. All results obtained up to now are matching Pauwels conclusions. This is remarkable since the mechanical model is three dimensional and includes 45 muscles, a complexity that could not be handled at Pauwels times.

The realization of this approach of course requires more precise informations about the external kinematics and kinetics of the movement and in particular about the 3-dimensional form and topology of the individual body segments. In collaboration with the University of Mainz and the Institute of Radiology in Muehlheim /Ruhr we were able to scan two complete legs from the toe to the pelvis with Magnetic Resonanz Imaging.

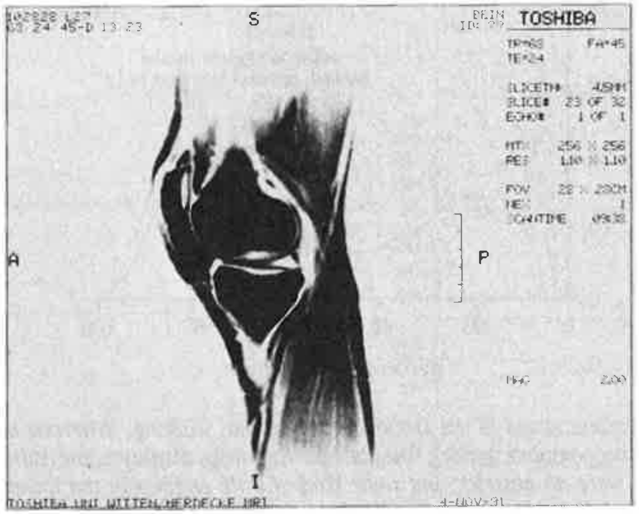

Fig. 7: Sample recording in the sagittal plane of the knee-joint using MRI. There are about 300 recordings necessary to scan one total leg.

In contrast to the common practice to measure the anatomical data on cadaver legs and scale the obtained values, we now have the unique chance to measure all required quantities directly on the living test subject, noninvasively and without any known risc. The above mentioned two legs belong to two different subjects with about the same height and the same weight. It has been found, that there are significant differences in the anatomical positions which can not been extracted from scaling procedures. Any lack of accuracy concerning the anatomical data can endanger the whole calculation and might influence the results more than the objective function itself.

The new methods for body structure measurements available now with computer tomography (CT) and magnetic resonance imaging (MRI) will give considerable support to this approach.

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UTILIZATION OF MUSCLE ELASTICITY IN MEN AND PREPUBERTAL BOYS

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The ability to utilize stored elastic energy in the leg extensor muscles is one factor determining the efficiency of movement in various forms of locomotion. The purpose of this study was to use a standardized test to investigate if this ability differs between prepubertal boys and adults, and if so, whether this difference could be related to variations in myoelectric activity patterns, as has been indicated in studies on running (Thorstensson 1986) and hopping (Moritani et al. 1989).

Seven adult males (age 25-38 years) and seven nine-year-old boys were studied. A paradigm of repeated knee flexions and extensions in a standing position was used, performed either with a direct rebound, making storage and utilization of elastic energy possible, or with a rest interval in the flexed position (Margaria et al. 1963). Mechanical work performed was calculated from recordings of vertical force via a force platform. Aerobic energy demand was estimated from oxygen uptake during steady state. Myoelectric activity was recorded with bipolar surface electrodes from the lateral portion of the quadriceps femoris muscle group.

The apparent mechanical efficiency was lower ($P<0.05$) for the boys than the adults in both experimental situations (rebound: 22.5 *vs* 27.6 %, no-rebound 17.8 *vs* 22.8 %). The values for the tests with rebound were significantly higher ($P<0.01$) than with no-rebound in both groups. Extra loading corresponding to 20% of body weight, applied symmetrically onto the upper body of the boys, did not change the efficiency. There was no difference between the men and boys in the general pattern of knee extensor muscle activation in relation to the phases of the respective movement or in the relationship between the amount of activity in the negative (eccentric) and positive (concentric) motion sequences. In the no-rebound tests, the integrated EMG activity was always clearly higher in the positive than in the negative phase of the movement, whereas the reverse was true for the rebound trials.

Thus, it appears that the ability to utilize elastic energy, as evidenced by a decreased aerobic cost in the rebound situation, was fully developed in the nine-year-old boys, and that the reasons for the generally lower work efficiency in the boys have to be sought among other factors.

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MEASUREMENTS OF PLANTAR PRESSURE DISTRIBUTIONS DURING THE SHOT PUTT AND DISCUS AND JAVELIN THROWS

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The study of the plantar pressure distribution on the feet of athletes has been claimed to offer the potential of optimising technique, improving footwear and reducing injury to the foot as well as allowing improved modelling of the forces in the foot. Despite these claims, there is little research which has reported and quantified plantar pressures during track and field athletic activities. The objectives of this study were to measure the pressure distributions on the plantar surface of the foot during three throwing events in track and field athletics. As both peak pressure and the pressure-time integral have been considered to have implications for soft tissue damage, this paper will concentrate on those measurements.

Two subjects were used in the study, carried out in May 1991 at the University of Innsbruck, one a good standard decathlete who performed the javelin throw and one specialist shot putter and discus thrower. Following habituation and warm-up, pressures were recorded for at least three javelin and discus throws by the decathlete sampling both EMED insoles (at 40 Hz) and for similar numbers of shot and discus throws by the specialist thrower with only one insole being fitted and sampled to obtain the higher sampling frequency (100 Hz). The magnitudes and locations of the peak pressures and the maximum pressure-time integrals for all trials are shown in table 1.

Table 1. Peak Pressures (PP) in kPa and Maximum Pressure Integrals (MPI) in Pa.s.

Javelin	Final Front (Right) Foot Contact		Shot	Left Foot Landing at Front of circle	
Throw	PP-site	MPI-site	Throw	PP-site	MPI - site
1	470 - M1	104 - M1	1	580 - M1	200 - Hallux
2	850 - M1	191 - M1	2	480 - M1	250 - M2-5
3	820 - M1	170 - Hallux (164 M1)	3	570 - M1	210 - Hallux

Discus	Left Foot Drive into Entry	Left Foot Landing at Front of Circle	
Throw	PP - site	PP- site	MPI - site
1	500 - M2-5	460 - M1	135 - Hallux
2	620 - M2-5	500 - M1;M2-5	136 - Hallux
3	580 - M1;M2-5	490 - M1	139 - Hallux

The greatest peak pressures were recorded for front foot impact in the javelin delivery stride, which would be expected as this follows a run up rather than a movement across the circle. The differences between the first and the other two throws was caused by a flat foot landing for this throw compared with the other two in which initial contact was on the medial border. This showed that this athlete had not developed a consistent landing pattern. In all the shot putts, as with all javelin throws, the peak pressures were recorded on the first metatarsal head (M1). The lower variability in these results probably reflects a more established motor pattern for the specialist thrower. The site of the maximum pressure integral changed between putts but without pronounced differences between the first metatarsal, metatarsals 2-5 (M2-5) and the hallux. In the discus throw reasonable intra-throw consistency was again seen for this specialist thrower with the greatest pressures occurring on one or both of the metatarsal regions at push off from the rear of the circle not on landing. The maximum pressure integrals, on the hallux, were very similar.

Although the peak pressures measured in this study were smaller than those reported by other authors for the trimple jump and running, such pressure distributions measurements offer considerable potential for enhanced biomechanical analysis of these track and field athletic events both in terms of technique assessment and also in evaluating injury risk during repetitive training throws.

SURFACE EMG OF AN ARM ERGOMETER, A KAYAK ERGOMETER AND KAYAK ON THE WATER

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Robertson (1985), Wilson et al. (1988) and Robertson et al. (1988) studied the muscular function in ergometer rowing in an effort to determine the most efficient technique of rowing. Inherent to their set-up approach it is assumed that such an ergometer is a specific simulation of the actual movement. Such an assumtion can be hazardeous since Clarys & Olbrechts (1983) and Clarys (1985) investigated for similar reasons - but in Swimming - the muscular activity of 5 "ergometer-type" dry land exercise devices in comparison with actual swimming. They found no EMG similarities or "specifity" at all.
In view of a quality evaluation of alternative (winter) dry land training for Kayak athletes this study has investigated the MREMG and IEMG of 6 shoulder & upper limb muscles during standardized submaximal & maximal work-outs on an Arm Ergometer, a Kayak Ergometer and in Kayak under real circumstances (N=7 ; mean age 22.3 years).
The analog raw EMG was recorded with a seven channel FM recorder (TEAC HR30) using pre-amplified bipolar surface electrodes supplied with a precision instrumentation amplifier (AD 524, Analog Devices Norwood, USA). These active electrodes were fixed on the midpoint of the contractred muscles. The hardware for the EMG data acquisition was designed for multidisciplinary purposes. The athlete was not to be disturbed during the movement. The system has a freedom of action (continuous measurement over several minutes). Several muscles as well as synchronization signals were monitored simultaneously. Influences of skin resistance phenomena were eliminated by means of high input impedance amplifiers (Clarys-Publie, 1987). The raw EMG was full wave rectified and enveloped using a moving average principle and normalized to the highest peak amplitude procedure per subject and integrated (Winter et al. 1980). Further analyses procedures were carried out with the Electromyographic Signal Processing and Analysis System - E.S.P.A.S. (Cabri, 1989). Qualitative pattern specificity characteristics were analysed with the IDANCO-EMG pattern evaluation system (Clarys-Cabri, 1988).
Statistical treatment of the IEMG data showed significant differences between all 3 test situation and for all muscles indicating a higher intensity level in the kayak ergometer as compared to the arm ergometer, both in maximal and submaximal work outs.
In view of these findings it is important to realize that the Kayak ergometer V02 max demands are equaly and significantly higher than the arm ergometer. While the qualitative treatment of the data with the IDANCO system indicates a higher degree of specificity for the MREMG patterns of the kayak ergometer (5° to 66%) as compared with the EMG collected in field circumstances. These data suggest that the kayak ergometer may be considered as a better alternative training system for kayak athletes.

EFFECTS OF SPRING LOADING ON KNEE EXTENSORS' EMG ACTIVITY

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Spring loading is a method sometimes used for strength training in sport and physiotherapy. Springs of various stiffness have also been used in studying the tremor characteristics of the neuromuscular system. Because the effects of spring loading on electromyographic activity (EMG) are not well known, the present study was designed to investigate EMG activity during contractions performed against a rigid isometric load (inelastic) and against five different spring loadings (isometric tremor).

On a dynamometer thirteen subjects performed isometric knee extensions with the dominant leg against a rigid resistance and against five different springs (780, 830, 935, 1090 and 1524 $N \cdot m^{-1}$) in turn at levels of 50 % and 30 % of MVC maintaining a 90° angle at the hip, knee and ankle joints.

Surface EMG activity was registered from the m. vastus lateralis and vastus medialis for the determination of full-wave rectified and averaged EMG values (AEMG) and of the mean power frequency (MPF) of the power spectral density function. The AEMG values of the submaximal contractions were expressed in relation to the EMGs measured for the MVC as follows

$$\mathrm{relAEMG} = \frac{\mathrm{AEMG}}{F} * \frac{F_{MVC}}{\mathrm{AEMG}_{MVC}}$$

The results showed the relAEMG values of the two knee extensor muscles at levels of 30 % and 50 % of MVC to be statistically significantly greater during the spring loading than during the rigid isometric loading. The spring stiffness affected ($p<0.05$) the relAEMG activity; the greatest value for both muscles and both contraction levels were measured for the spring of 1090 $N \cdot m^{-1}$ stiffness. The MPF values did not differ statistically significantly between the loading conditions. Enhancement reflex activation in connection with changes in the motor unit firing characteristics are suggested as influencing the higher relAEMG values for the spring than for the rigid isometric loading.

MUSCULAR CONTROL, EMG POWER SPECTRA AND BIO-MECHANICAL INDICATORS IN WEIGHTLIFTING

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INTRODUCTION

The relationship between EMG power spectra (PS) and the particularities of the activation of motor units (MU) for isometric and dynamic muscle contraction represents a topic of growing interest (Solomonow, Bilodeau). We combined EMG in the analysis of the biomechanical structure of the movement and in dynamometric monoarticular and specific testing protocols in the training of elite weightlifters.

METHODS

A special Kistler platform and an integrated automatic video-digitizing system were used. Computerized isokinetic dynamometers were also used to determine torque-angle profile of the knee, hip and elbow joints under eccentric, concentric and isometric conditions.

For EMG purpose, surface electrodes (O= 5 mm; interelectrode distance 45 mm) were placed over the belly of each muscle (VL, BF, TR, BB, ES). Bipolar myoelectrical potentials were recorded with the passive electrode placed between the two actives; signals were preamplified and band -pass filtered (CMRR≥ 70 db, BP= 10 Hz- 1 KHz; Zin=1,5 MΩ, gain= 1000). EMG signals and the vertical GRF were digitized on-line with a sampling frequency of 1000 Hz. For time structure analysis EMG signals were full-wave rectified and band-pass filtered (20-70 Hz) to obtain envelop curve patterns of each muscle. Power spectra analysis (median frequency MF) were performed using 1024 and 512 data points.

Six to ten competitive weightlifters participated in the investigations. Subjects executed typical competitive movements and also assistant exercises. An isometric specific test (maximal effort) was introduced in which the subject kept the same starting position as in the competitive exercise.

RESULTS

1) The activity of knee extensors and flexors during the second knee bend and the following second pull is different among the athletes. 2) In the second pull, a lenghtened duration of BB (or an early activation) is often combined to a shortening of the duration of VM. 3) The inter activity pattern of VM, BF, and BB seem to be representative of the main invariant characteristics of the movement sequences. 4) Difference in the inter activity pattern are not necessarily matching similar variation in the GRF. This is essential for the usefullness of GRF as interpretative and diagnostic indicator in training. 5) There is a correlation between increasing load conditions and the duration of activity and rest of the muscle tested. However there are different trends according to the type of exercise and to the individual control strategies. 6) In the power spectra, between isometric and isokinetic test conditions and maximal specific effort, there are consistent but also discontinuous trends. MF indicates that some subjects have reserves in recruiting MU (significant increase in MF for the VM during concentric tests after 4 weeks).

THE ELECTROMYOGRAM DURING ISOKINETIC TEST

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Our work concerns with the study of the EMG activities in pertinent skeletal muscles during an isokinetic test (constant speed) of the knee joint. During isokinetic test the subject exerts his maximal strength while progressing at costant speed throughout a limited range of motion. The movement is a succession of periodically reversing rotations round the joint axis of rotation (flexion/extension). Electromyographic signals were recorded by two pairs of surface electrodes placed along the length of the rectus femoris, vastus medialis, vastus lateralis, semitendinous and of the biceps femoris. The signals were acquired at three different speed (90, 120, 180°s-1). In fig.1 the poligraphic record of the raw electromyogram is shown. To have a condensed parametric representation of the information of the EMG we have utilized the autoregressive (AR) and autoregressive-movingaverage (ARMA) models . The electromyogram, which exibits weak stationarity over short time interval, can be represented by a sixth-order AR model or by a (2,2) ARMA model. The values of the AR model coefficents present large variations whereas the ARMA model coefficients are more constant. In the zero-pole representation, the ARMA model exhibits a fixed pole configuration, which depend mainly by the muscle. The pole configuration of AR model is variable and does not show any correlation with the muscle. In the recent advance of our work we have applied a recursive algorithm to compute the ARMA model. The zero-pole representation of the recursive model shows a different behaviour of the poles and the zeros. The poles are fixed during the active phase of the test the zeros modify their configuration. These behaviours are in agree with the modification of the spectrum array computed during this active phase.

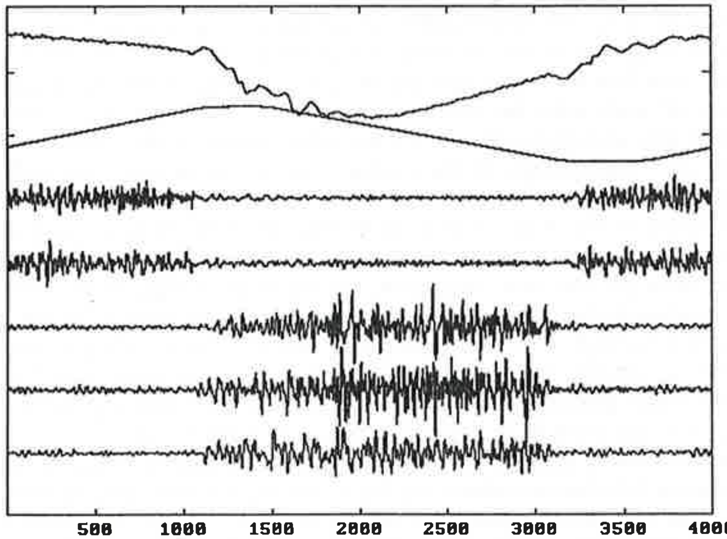

Fig.1 Isokinetic poligrafy. Force, position, biceps femoris, semitendinous, vastus medialis, rectus femoris and vastus lateralis.

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THE SUITABILITY OF THE EMED INSOLE FOR THE MEASUREMENT OF PRESSURE DISTRIBUTIONS ON THE PLANTAR SURFACE OF THE FOOT DURING THE ATHLETIC THROWING AND JUMPING EVENTS

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The objectives of this study were to assess the suitability of the EMED pressure insole for the measurement of the pressure distributions on the plantar surface of the foot during a range of throwing and jumping athletic events and to validate the overall vertical forces measured by an EMED insole against those recorded simultaneously from a Kistler force platform.

The study reported here was carried out during May 1991 at the University of Innsbruck. The 2 mm thick EMED insole consists of 85 capacitive sensors, one per 2 cm^2, homogeneously distributed, with a maximum sampling rate for the total sensors on one insole of 100 Hz or of 40 Hz if a pair of insoles are sampled. Four subjects were used in the study: one good standard decathlete, two specialist jumpers and one specialist thrower. In the first set of trials, a pair of EMED insoles was carefully fitted into the soles of the decathlete's shoes, after calibration, followed by those of one of the specialist jumpers. After habituation and warm-up, pressures were recorded for at least three javelin and discus throws and long jumps by the decathlete and several triple jumps by one jumper sampling both insoles (at 40 Hz). Simultaneously, a Kistler Model 9281B force platform was used to record the three components of ground reaction forces, at a sampling rate of around 200 Hz, for the landing of one foot for jump take off or last foot contact for the throw. In the second set of trials with the other two athletes, the thrower performed similar numbers of shot and discus throws and the other jumper made similar numbers of high, long and trimple jumps. In all of these, only one insole was fitted and sampled to obtain the higher sampling frequency.

Comparisons of the total vertical force from the EMED insole and the force platform showed the two to follow very similar trends with identical numbers of peaks and throughs, the mean discrepancy in the times of occurrence of which was only 15 ms, close to the accuracy with which measurements could be taken from the final graphical displays. The mean discrepancy between the forces registered at these turning points (where the discrepancies were at their greatest) was 170 N; the differences were probably due mainly to the different sampling rates and the summing of forces from 85 individual transducers in the insole.

The results showed that the EMED insole gave a reasonable temporal resolution of the pressure distributions when sampled at 100 Hz, but that sampling both insoles at 40 Hz was too slow for events of such a dynamic nature. No problems were experienced when using the insole for the three throwing events and the high jump. However, the first pair of insoles was severly damaged by the attempt to measure pressure distributions for the triple jump, whilst one of the replacement insoles was

damaged by either long or triple jumping. The main problem would appear to be the high shear forces involved in these events, possibly compounded by the presence of substantial spikes in the footwear.

In conclusion, the study showed the EMED insole to measure forces in good agreement with those from a Kistler force platform. The pressure distribution results have considerable potential for biomechanical studies of these track and field athletic events, especially because both peak pressure and the pressure-time integral have been considered to have implications for soft tissue damage. However, it is felt that the manufacturers would need to increase the sampling rate with both insoles to around 100 Hz to increase the temporal resolution, as the information from both feet is of importance in most athletic events (as in many other sports). Furthermore, some attention should also be paid by the manufacturers to improving the ability of the insole to withstand large shear forces without failure, perhaps by increased padding on the insole.

A STANDARDIZED METHOD TO ASSESS VERTEBRAL PAIN IN SPORT ACTIVITIES

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Vertebral pain is one of the major disorders among the general population and also represent a very high social-cost. Distribution of this disorder, in working communities has been evidenced by several epidemiological studies. As for as concerns sport activities, it has been put in that vertebral disorders were present not only in amateurs but also in professional athletes. The different methodologies used in these studies, introduce difficulties in comparison and in the adoption of these methods in preventive and rehabilitation practice. The aim of the present study is to verify the possibility of applying a standardized metodology for assessment of vertebral pain in sport activities. Preliminary data, collected on 17-29 year old athletes engaged in different sports, using a standardized self-reported questionnaire are presented. Our sample consisting of athletes engaged in swimming, rhythmic gymnastics, body-building, basket ball end foot-ball shows in half of subjects back pain. A decrease in performance and an absence from training session was reported by some athletes. In few istances pain was treated.

STABILOMETRIC EVALUATION OF ANKLE FUNCTIONAL INSTABILITY IN A CLINICAL SETTING

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1 - Introduction

In literature there are studies about prevention of ankle sprain, but objective methods to evaluate subjects at risk were not frequently used, and always in laboratory settings [1,2,3]. Obtaining an effective method in screening in a clinical setting which subjects engaged in sport activities might develop functional ankle instability was the aim of our study.

2 - Methods

15 male basketball players (30 ankles) were examined standing on one foot (arm around chest, eyes open) for 60 seconds. Their ankles were divided in 3 populations according to history and clinical examination: normal (9 ankles): no history of sprain (Group N); remote pathology (9 ankles): previous history of sprain with complete remission (Group PR); pathology (12 ankles): suspect history of ankle functional instability (Group P).

We used the "Balance Platform" system made by Cosmogamma. 3 mechanical-electrical force transducers under the platform calculate via computer the movements of Center of Pressure (ground projection of Gravity Center of the body) using ground force reactions on feet. Software offers data base management and video and printed output. Examinations were carried out always by the same physician, minimizing visual and earing disturbances. There was one fixation point, a vertical line, at two meter distance.

3 - Results and Discussion

Groups in our study did not differ for age and weight, but they did for height (P vs PR and P vs N). Statistical analysis permitted to verify that group P and N were different in almost all variables examined ($P< 0,05$); we obtained the same results comparing Group P and PR, but only in those variables that a previous study demonstrated to be reliable (Space, Speed and Sagittal Speed, but not Frontal Speed); Group PR and N diverged just for Frontal Oscillations. We computed with Discriminant Analysis 3 functions (General, Sagittal and Frontal components), but only the first one was really able to discriminate between populations: ankles in Group N remained always under the zero-line which represented a limit to distinguish from Group P; Group PR showed to be composed of ankles either completely normal or presumably pathological

4 - Conclusion

If a screening has to be rapid, reliable, of low cost and able to discover disorders that otherwise could remain unknown [4], then this method could be applicable in athletes at high risk of ankle sprain to diminish its incidence and to prevent possible complications that could lead to abandon sport activities sometimes professional.

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AN OPTIMIZING MODEL OF MECHANICAL POWER OUTPUT DEVELOPMENT FOR 13-15 YEAR OLD BOYS

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Introduction. The aim of this study is the presentation of the way to the mathematical model of the training process of 13-15 years old boys and optimization of this process. The performance index is maximal value of mechanical power output (springness) in a given final phase of the training. The 7-dimensional state vector contains:springness, velocity of straightness of foots, velocity, mechanical work and power, time of supercompensation and index of maximal mechanical power while the percentage of mechanical work and power are control parameters.

Methods. By observation of 3 groups of 30 boys in period of 3 years (all results were assemble every quarter) with different values of controls, using some statistical methods, methods of computer's graphies and the least-square approximation methods the mathematical model of the problem was constructed in form of the following bilinear control system:

$$x = Ax + Bux + b + h(t,u), \quad x(to) = xo$$

(A,B are some constant matrices and b is a vector) with the performance index J(u) = x7(t1), which must be maximized.

Results and discussion. At first, the existence of optimal control is proved. Next the optimal control is determined using the Krylov-Chernousko algoritm with the help of IBM PC minicomputer for few boys. As it was computed, the value of the performance index obtained by application of the optimal control were about 20-30% greater than those from empirical data. In conclusion, the application of our model together with methods of optimal control theory shows some colossal reserves which, in practice, was difficult to detect in another way.

COMPUTED EVALUATION OF ISOKINETIC ENDURANCE IN TWO DIFFERENT SPORTS

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Introduction

The assessment of muscle fatigue is crucial in evaluation of athletes as well as in the sport Medicine field. Several procedures has been reported in this connection. Among these methods, muscle assessment by means of isokinetic procedure analysis seems to be particularly interesting, due to both subjects compliance and reliability, the latter having already been proved in a number of experimental studies. On the basis of instructions provided by the isokinetic equipment's manufacturer, such a procedure has been further improved in our Center by means of sophisticated computed analysis of diagrams, recorded during a prolonged muscular work.

Material and Methods

30 young men involved in semiprofessional sport practice were assessed (age range 16-22). The series included two main groups: 15 footballers and 15 cyclists. Each subject was submitted to an isokinetic endurance test consisting in a series of knee flexionextension cycles at 180 degrees/sec for 1 minute. The analogic output corresponding to knee torque and range of motion, recorded by the isokinetic dynamometer, are converted into a digital mode. A properly designed PC program enables the analysis of the following parameters: Initial strength peak value (on the basis of the first five cycles only); Number of cycles and; Time necessary for a strength decrease of 20 - 30 and - 50% respectively with respect to the initial strength peak ; Slope of the line intersecting the endurance curve (according to the "minimum square" principle), the former representing the percentage of strength decrease recorded after each cycle; Total work performed by each subject and number of cycles necessary for a work decrease of 20 - 30 and - 50% respectively.

Results

The analysis of results shows that: When considering identical strength peaks in both groups, strength decrease (slope) was faster in footballers than in cyclist, since the number of cycles and the work necessary for a corresponding strength decrease was averagely greater in the latter. Such a difference was particularly significant for the knee extensor muscles.

Discussion

The major concerns provided by data analysis is represented by the different fatigue pathway observed in footballers and in cyclists respectively. Therefore, such an outcome confirms theoretical expectations. As a matter of fact, quadriceps performances are peculiar in cyclists and require a significantly higher percent of type I fibres than in footballers, as already shown in former studies. Longitudinal investigation on subject belonging to both groups of athletes confirmed that the present procedure is also reliable to assess increases in muscle endurance induced by specific training, in absence of muscular strength improvements.

AUTHOR INDEX